# Force Tracking With Feed-Forward Motion Estimation for Beating Heart Surgery

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Abstract—The manipulation of fast-moving, delicate tissues in beating heart procedures presents a considerable challenge to the surgeon. A robotic force tracking system can assist the surgeon by applying precise contact forces to the beating heart during surgical manipulation. Standard force control approaches cannot safely attain the required bandwidth for this application due to vibratory modes within the robot structure. These vibrations are a limitation even for single degree-of-freedom systems that drive long surgical instruments. These bandwidth limitations can be overcome by the incorporation of feed-forward motion terms in the control law. For intracardiac procedures, the required motion estimates can be derived from 3-D ultrasound imaging. Dynamic analysis shows that a force controller with feed-forward motion terms can provide safe and accurate force tracking for contact with structures within the beating heart. In vivo validation confirms that this approach confers a 50% reduction in force fluctuations when compared with a standard force controller and a 75% reduction in fluctuations when compared with manual attempts to maintain the same force.

Index Terms—Beating heart surgery, force tracking, medical robotics, motion compensation, 3-D ultrasound.

I. INTRODUCTION

In BEATING HEART procedures, the surgeon operates on the heart while it continues to pump. These procedures eliminate the need for cardiopulmonary bypass and its associated morbidities [1] and allow the surgeon to evaluate the procedure under physiological loading conditions. The latter is particularly useful in the repair of cardiac structures like the mitral valve that undergo substantial mechanical loads during the heart cycle [2]. However, surgical manipulation of the beating heart is challenging because heart motion exceeds the human tracking bandwidth of approximately 1 Hz [3]. The mitral valve annulus, for instance, traverses most of its 10–20 mm trajectory and undergoes three direction changes in approximately 100 ms [4], which makes the application of precise forces for surgical tasks, like mitral valve annuloplasty, difficult. Indeed, recent animal trials indicate that beating heart repair of the mitral valve cannot be performed reliably due to its fast motion [5].

A force controlled robotic surgical system could benefit the surgeon by applying precise forces to the heart as it moves. Previous work on surgical force control has largely focused on force feedback for teleoperation of surgical instruments and robots, as reviewed in [6]. Force feedback has demonstrated a number of performance benefits in the execution of remote surgical tasks [7], [8] and can enhance safety when it is used to implement virtual workspace limits [9]. In this setting, the primary role of the force controller is to provide haptic information to the user while the user commands the robot to interact with the surgical target.

In contrast, beating heart applications require the robot controller to autonomously maintain the prescribed forces of the instrument against the target tissue despite its fast motion. One major concern is safety, given the well-documented occurrence of instability in force control [10]–[13]. A robotic system for beating heart surgery must be damped and stable to ensure that it will not overshoot or oscillate in response to changes in the desired force trajectory or sudden target motions. Furthermore, the system must have sufficient bandwidth to reject the disturbance that is caused by heart motion. Previous research indicates that standard force control strategies can only achieve stability for low closed-loop bandwidths due to vibratory modes in the robot structure [11]–[13]. These findings were obtained in the context of large industrial robots that interact with stiff targets. To ensure adequate robot performance and safety, it is essential to determine whether the same limitations exist in beating heart surgery where the target is soft but rapidly moving.

In this paper, we study force control in the context of beating heart surgery and find that the standard force controller does, indeed, suffer from bandwidth restrictions due to the vibratory modes that are present in long surgical instruments. However, by incorporating feed-forward tissue motion information into the controller, safe and accurate force tracking can be achieved at low bandwidth. In preliminary work, we experimentally demonstrated the efficacy of the approach [14], and in the following sections, we provide a detailed analysis of the feed-forward force controller, as well as in vitro and in vivo validation. In the first part of this paper, we show that simultaneously achieving an adequately damped system with
good disturbance rejection is challenging because it requires a closed-loop bandwidth that would excite undesired vibratory modes in the robot. Subsequently, we describe a force-tracking system that bypasses these bandwidth limitations by using feed-forward heart motion information that is derived from 3-D ultrasound to augment the controller. The system, shown in Fig. 1, is adapted for beating heart mitral valve annuloplasty. It uses a 1 degree-of-freedom actuated instrument, termed the motion compensation instrument (MCI; see Fig. 2), that can follow the rapid, nearly uniaxial motion of the mitral valve annulus [4]. We validate our system and demonstrate its utility to the surgeon in an in vivo experiment in a large animal model.

II. RIGID BODY ANALYSIS

To gain some insight into the use of force control in beating heart surgery, we first consider the case of a perfectly rigid robotic instrument. The robot is modeled as a mass \( m \) and damper \( b \) that is subjected to a commanded actuator force \( f_a \) and environment contact force \( f_c \). The damper \( b \) captures the effects of friction in the robot, friction at the insertion point to the heart, and fluid motion. Approximation of the environment as a spring of stiffness \( k_c \) yields the system dynamics

\[
m\ddot{x} + b\dot{x} = f_a - k_c(x - x_e),
\]

where \( x \) is the instrument tip position, and \( x_e \) is the desired tissue target position (i.e., its position if it were not deformed by contact). The model in (1) assumes rigid contact between the instrument and compliant target and is illustrated in Fig. 3.

For simplicity, we neglect force sensor compliance in the model because it is significantly stiffer than the tissue environment. Let us now consider a standard force regulator control law given by

\[
f_a = f_d + K_f(f_d - f_c) - K_v\dot{x},
\]

where \( K_f \) and \( K_v \) are controller gains, and \( f_d \) is the desired force [16]. Combining (1) and (2), and applying the Laplace transform gives the closed-loop contact force relationship

\[
F_c(s) = T(s)F_d(s) + Z(s)X_c(s)
\]

where the force tracking transfer function \( T(s) \) and robot impedance transfer function \( Z(s) \) are as follows:

\[
T(s) \triangleq \frac{F_c(s)}{F_d(s)} = \frac{(k_c/m)(1 + K_f)}{C(s)},
\]

\[
Z(s) \triangleq \frac{F_c(s)}{X_c(s)} = -\frac{k_c(s + (K_v + b)/m)}{C(s)},
\]

\[
C(s) = s^2 + \frac{K_v + b}{m}s + \frac{k_c}{m}(1 + K_f).
\]

Equation (3) explicitly shows that target motion \( x_e \) is a disturbance that perturbs \( f_c \) from \( f_d \).

Controller gains \( K_f \) and \( K_v \) are chosen to ensure system stability, sufficient damping, and good rejection of \( x_e \). The last is achieved by designing \( Z(s) \) to have small magnitude in the bandwidth of \( X_c(s) \). For the mitral valve annulus, which is bandlimited to approximately 15 Hz [4], this is equivalent to setting the impedance corner frequency \( f_c \) greater than or equal to 15 Hz. Fig. 4 depicts typical mitral valve annulus motion [4] and its effect on the contact force for various \( f_c \) based on simulations of (3)–(5), with \( f_d = 0 \). Parameter values of \( m = 0.27 \text{ kg}, b = 18.0 \text{ Ns/m}, \text{ and } k_c = 133 \text{ N/m} \) are assumed based on system identification of the MCI and preliminary estimates of the mitral valve annulus stiffness. As we will show shortly, obtaining a large impedance corner frequency \( f_c \) is synonymous with an increase in the natural frequency \( f_n \) of the closed-loop system, which is also equivalent to an increase in \( K_f \).

The natural frequency \( f_n \) and damping \( \zeta \) of the system can be expressed as follows:

\[
f_n = \frac{1}{2\pi} \sqrt{\frac{k_c}{m}(1 + K_f)},
\]

\[
\zeta = \frac{K_v + b}{4\pi mf_n}.
\]
To avoid potentially dangerous overshoot, we set the system to be critically damped ($\zeta = 1.0$). By manipulation of (5), (7), and (8), it can be shown that the natural frequency $f_n$ of a critically damped system is a function of the impedance corner frequency $f_z$, which is given by

$$f_n = \left(\frac{\sqrt{208} - 14}{6}\right)^{-1/2} f_z \approx 3.7698 f_z. \tag{9}$$

Hence, while a position regulator can follow a trajectory that is bandlimited to 15 Hz with about the same closed-loop natural frequency, a force regulator must have a natural frequency of approximately 57 Hz to follow the same motion. This indicates that the force regulator inherently requires high bandwidth to compensate for target motion. From (7)–(9), we calculate gain settings of $K_f = 255.2$ and $K_v = 173.8$ for $f_z = 10$ Hz, and $K_f = 2845.7$ and $K_v = 621.4$ for $f_z = 50$ Hz. Mitral valve annulus motion data is from [4].

**III. BANDWIDTH CONSTRAINTS DUE TO ROBOT DYNAMICS**

In this section, we study the effect of high gains on the robot performance. Extensive prior work on force control has delineated a number of instabilities that can arise when attempting to control forces using multiple degree-of-freedom robot arms in contact with hard surfaces [11]–[13]. In particular, force controllers can excite structural modes in the manipulator, thereby leading to high-amplitude force transients at the end effector. These mechanisms do not pertain to this surgical application, where the end effectors tend to be long, rod-like instruments that reach patient anatomy through small ports, and tissues are highly compliant.

It is well known, however, that the axial motion of such long rods excite transverse vibrations [17]. The dynamics that describe this motion are nonlinear and have time-varying parameters; however, for the purposes of developing effective force controllers, it is not necessary to model these dynamics: We only need to determine the frequency at which they become significant so that reasonable bandwidth restrictions can be imposed on the rigid body model and closed-loop system analysis of the previous section. While structural resonance can be used to enhance the performance in some situations, it is typically avoided because inadvertent excitation of the resonance can destabilize the controller, reduce the positional accuracy of the instrument, and cause undue wear to the robot. In the present application, resonance can further cause injury to the patient due to the transmission of vibrational energy to the tissue that is in contact with the robot.

In the following sections, we analytically and empirically demonstrate that vibration precludes the use of the high-gain force regulator suggested in Fig. 5. We first demonstrate that vibration occurs at relatively low frequency for surgical robots with long instruments. Subsequently, we empirically demonstrate that these vibrations are significant in our system and can lead to instability as $K_f$ is increased until the natural frequency approaches the resonance of the instrument.

**A. Gain Limit to Avoid Vibration**

Let us consider a cylindrical rod of length $l$ and radius $r$ that undergoes axial motion while being compressed by a force $f_c$ (see Fig. 6). At low velocities and at compressive loads that are much smaller than the Euler buckling load ($f_c \ll E\pi^3 r^4/4l^2$), the fundamental mode of transverse vibration is well approximated by

$$f_1 \approx \frac{3.5156}{4\pi} \sqrt{\frac{E}{\rho}} \left(\frac{r}{l^2}\right), \tag{10}$$

where $E$ and $\rho$ are, respectively, the Young’s modulus and density of the material that makes up the rod [18]. At large velocities, the fundamental frequency is time-varying [17]. At large compressive loads, the fundamental frequency decreases [19]. We omit both of these phenomena for the ease of analysis.

The closed-loop natural frequency of the system should be set lower than the first resonance ($f_n < f_1$) in order to avoid the effects of vibration. Combining (7) and (10) provides a limit on the proportional gain

$$K_f^\text{limit} = \frac{m}{K_c} \left(\frac{3.5156^2 E r^2}{4\rho l^4}\right) - 1. \tag{11}$$

The value of $K_f$ should be chosen substantially lower than $K_f^\text{limit}$ so that the gain of the closed-loop system is small through the spectral extent of the resonance.

The MCI is mounted with a stainless steel, 14 gauge blunt needle ($E = 200$ GPa, $\rho = 7900$ kg/m$^3$, and $r = 1.1$ mm) with an inner stainless steel push rod that is used for anchor

![Fig. 4. The effect of mitral valve annulus motion on contact forces. (a) Human mitral valve annulus trajectory and (b) corresponding force disturbances for $f_z$ of 5, 15, and 50 Hz. The mitral annulus has a motion bandwidth of approximately 15 Hz [4]. Gain settings are $K_f = 27.5$ and $K_v = 45.9$ for $f_z = 5$ Hz, $K_f = 255.2$ and $K_v = 173.8$ for $f_z = 10$ Hz, and $K_f = 2845.7$ and $K_v = 621.4$ for $f_z = 50$ Hz. Mitral valve annulus motion data is from [4].](image)
The desired force is \( f_n \). For simplicity, we approximate the entire structure as ranged from 0 to 55 in steps of five and 

\[ \text{Forcing to predict the fundamental resonance frequency and gain limit} \]

The MCI was positioned horizontally and clamped along its body to restrict vibrations to the instrument shaft only. The instrument tip was imaged by a digital camera (EOS 20D, Canon, Tokyo, Japan) with 200 ms exposure time. Vibration amplitudes were measured by a scale placed in the image and oriented to the plane of motion.

\[ \text{Excitation of the observed resonance would not be safe for beating heart surgery. For context, the mitral valve annulus is a ring of smooth tissue with an approximate width of a few millimeters. While in contact with the annulus, vibration could cause the instrument to slip into the mitral valve or adjacent cardiac structures. Contact with tissue could dampen the vibration, but this would also transfer its energy to the patient anatomy and could cause injury.} \]

2) Force Regulator Instability at High Gain: In this experiment, we study the effect of vibration on stability by testing increasing values of \( K_f \) that approach the predicted \( K_f^{\text{limit}} = 70.1 \) from Section III-A. This is equivalent to testing the system at increasing natural frequencies that approach the experimentally observed 27 Hz resonance of the MCI.

As in the previous experiment, the MCI was positioned horizontally on a flat surface and clamped along its body to restrict vibrations to the instrument shaft only. The instrument tip was placed in series with a steel leaf spring, with a stiffness matched to the approximate stiffness of the mitral valve annulus \( (k_c = 133 \text{ N/m}) \). The desired force \( f_n \) was a unit step and the MCI was controlled by the force regulator law in (2). Gain values for \( K_f \) ranged from 0 to 55 in steps of five and \( K_v \) was fixed at 50. These gain values are marked in Fig. 5 to illustrate their effects of discretization are negligible: The plant had a natural frequency of 3.5 Hz and the closed-loop system reached a maximum natural frequency of 26.4 Hz. Ten trials were performed for each gain setting for a total of 120 trials. Controller performance was judged to be stable or unstable for each trial, with the latter condition defined as exhibiting nondecaying oscillations.

\[ \text{Fig. 5. (a) Impedance corner frequency} f_1 \text{ and (b) damping} \zeta \text{ over different values of} K_f \text{ and} K_v \text{ using a rigid body model of the MCI. A tradeoff exists between damping and disturbance rejection. The natural frequency} f_n \text{, which is related to} K_f \text{ through (7), is also shown. Dots indicate the gain test points used for laboratory tests, as described in Section III-B2.} \]

\[ \text{Fig. 6. Axially oscillating cantilevered rod of length} l \text{ and radius} r \text{. Lateral motion at the base of the rod} (x) \text{ excites transverse vibrations} (y) \text{ [17].} \]

\[ \text{Fig. 7 depicts the transverse vibrations that were observed at} 10, 27, \text{ and} 30 \text{ Hz. Large transverse motions are apparent at} 27 \text{ Hz. Fig. 8 shows the magnitude of the transverse displacement, normalized by the magnitude of the axial displacement over frequency. The resonance frequency occurs at} 27 \text{ Hz, which is in close agreement with the predicted value from Section III-A. The resonance peaks to a value of} 9.8 \text{ dB and its effect becomes small at approximately} 22.2 \text{ Hz. Vibration for inputs with frequencies below} 10 \text{ Hz and from} 50 \text{ to} 100 \text{ Hz were negligible.} \]

\[ \text{Excitation of the observed resonance would not be safe for beating heart surgery. For context, the mitral valve annulus is a ring of smooth tissue with an approximate width of a few millimeters. While in contact with the annulus, vibration could cause the instrument to slip into the mitral valve or adjacent cardiac structures. Contact with tissue could dampen the vibration, but this would also transfer its energy to the patient anatomy and could cause injury.} \]

1) Characterization of the Transverse Vibration Over Frequency: The MCI was positioned horizontally and clamped along its base to restrict vibrations to the instrument shaft only. The instrument was commanded to follow axial, sinusoidal motion inputs at frequencies between 1 and 100 Hz. The axial position of the actuator was measured by a high linearity potentiometer (CLP13-50, P3 America, San Diego, CA) at 1 kHz. Transverse vibration of the instrument tip was imaged by a digital camera (EOS 20D, Canon, Tokyo, Japan) with 200 ms exposure time. Vibration amplitudes were measured by a scale placed in the image and oriented to the plane of motion.

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\[ \text{Pre} \text{liminary animal testing found this to be the minimum length is necessary for the instrument to access the mitral valve during beating heart procedures in a porcine model. The instrument approach was from the left atrial appendage through the second intercostal space in a left thoracotomy.} \]
in excess of 50% overshoot for more than 1 s after the unit step input.

Fig. 9 shows the percentage of stable trials over $K_f$. As expected, increasing the gain reduces damping and eventually leads to the unstable behavior. Instability first arises for 10% of the trials at $K_f = 40$, which corresponds to a closed-loop natural frequency of $f_n = 22.6$ Hz, and is nearly coincident with the location of the 0 dB corner frequency for the observed resonance (see Fig. 8). Increasing $K_f$ to 50 and above leads to 100% of the trials being unstable. This is less than the predicted upper bound of $K_f^{\text{limit}} = 70.1$ but is nonetheless expected because of the spectral width of the resonance seen in Fig. 8. Overall, these results suggest that exciting vibrational modes, even in a single degree-of-freedom robot, can lead to an unstable, unsafe controller.

IV. FORCE CONTROL WITH FEED-FORWARD TARGET MOTION

The previous sections indicate that vibrational modes in surgical instruments prevent the high-gain settings that were required for a force regulator to obtain both damping and good heart motion rejection. Rather than using a pure force error-feedback control strategy, an alternative strategy employs feed-forward target motion information in the controller. Previous work has shown that this approach can improve force tracking when dealing with moving or uneven surfaces [21]. This approach is well-suited to our application because accurate predictions of heart motion can be obtained by exploiting its periodicity [4, 22, 23, 24].

Let us consider the control law

$$f_a = f_d + K_f (f_d - f_e) + K_v (\dot{x}_e - \dot{x}) + b \ddot{x}_e + m \dddot{x}_e,$$

which is (2) augmented with feed-forward estimates of the target velocity $\dot{x}_e$ and acceleration $\dddot{x}_e$. The contact force relationship in (3) becomes

$$F_e(s) = T(s)F_d(s) - Z(s)(X_e(s) - \dot{X}_e(s)),$$

where $T(s)$ and $Z(s)$ are as previously defined in (4) and (5), respectively. Observe that the use of feed-forward terms $\dot{x}_e$ and $\dddot{x}_e$ enable the cancellation of the motion disturbance $x_e$ without the need to greatly increase the natural frequency of the system. The controller can then be designed with a low closed-loop natural frequency to avoid the effects of vibration and other high-order dynamics that lead to reduced damping and instability. The feed-forward bandwidth is set equal to the bandwidth of the heart motion disturbance, which is lower than the resonance frequency of the robot.

V. TISSUE MOTION ESTIMATION WITH THREE-DIMENSIONAL ULTRASOUND

Because our surgical application is performed on the mitral valve annulus inside the beating heart, a real-time imaging technology that can image tissue through blood is required for guidance. We employ 3-D ultrasound because it is currently the only technology that meets these criteria while providing volumetric information. To obtain the motion terms that are needed in the feed-forward controller, we first determine the position of the tissue in the ultrasound volume using the real-time tissue segmentation algorithm from [20]. The algorithm takes advantage of the high spatial coherence of the instrument, which appears as a bright and straight object in the volume, to designate the tissue target. Fig. 10 depicts the method to track a mitral annulus point in a beating porcine heart using this method.
et al. ζ = 4 are the harmonic amplitudes, and c = 8 ± 10 Hz (K f = 4.1 and K v = 10.5) based on parameter values m = 0.27 kg, b = 18.0 Ns/m, and k e = 133.0 N/m. The gains were left intentionally low to guarantee controller stability in the unstructured environment of the operating room, where off-axis loading could result in instrument bending, as well as to account for uncertainty and variability in heart stiffness. The elastic properties of the heart can vary by a factor of three from patient to patient in normal human hearts, and hearts afflicted with congestive cardiomyopathy are on average five times stiffer than the average healthy heart [25].

The force tracking system uses a dual CPU AMD Opteron 285 2.6 GHz PC with 4 GB RAM to process the ultrasound data and control the MCI. The 3-D ultrasound machine streams volumes at 28 Hz to the PC over a 1 Gb LAN using TCP/IP. A program written in C++ retrieves the ultrasound volumes and loads them onto a GPU (7800GT, nVidia Corporation, Santa Clara, CA) for real-time beating heart tissue segmentation. This provides tissue position measurements that are used to train the extended Kalman filter. After 5 s of initialization, the filter outputs estimates of the tissue velocity and acceleration. These are used in tandem with force measurements from a custom, tip-mounted optical force sensor (0.17 N RMS accuracy and 0–4 N range [14], [15]) according to the control law in (12) in a 1 kHz servo loop. As previously stated, the controller was approximated as continuous because the effects of 1 kHz discretization on this 15 Hz bandlimited system is negligible. The MCI is powered by a linear power amplifier (BOP36-1.5 M, Kepco, Flushing, NY).

B. Results

Fig. 12 provides example force traces for the task executed manually, with the force regulator, and with the feed-forward force controller. Table I lists the force standard deviations for each trial. Averaged across all trials, manual contact with the annulus yielded force standard deviations of 0.48 ± 0.06 N (mean ± standard error). The force regulator reduced these deviations to 0.22 ± 0.01 N with clear statistical significance in a two-sided t-test (p = 0.012). The feed-forward force controller...
The force regulator and feed-forward force controller also reduced peak-to-peak forces. Manual use of the instrument gave swings in the contact force of $2.57 \pm 0.29$ N. The force regulator and feed-forward force controller reduced these forces to $1.16 \pm 0.10$ and $0.65 \pm 0.04$ N, respectively. Statistical significance was found between all conditions at $p < 0.05$.

VII. DISCUSSION

The results from our in vivo experiment underscore the benefit of a force controlled robot in beating heart procedures. Without force control, placement of the instrument against the mitral valve annulus gave peak-to-peak force swings of $2.57$ N, which is unacceptable compared to the desired $2.5$ N force set point. The standard force regulator reduced this fluctuation by 50% and the feed-forward controller reduced it by another 50%. In the case of the feed-forward controller, the precision of the contact forces was $0.11$ N. In all of the force controlled experiments, the surgeon expressed greater confidence in instrument manipulation against the beating mitral valve annulus, with the feed-forward controller subjectively better than the standard force regulator. These findings suggest that robotic force control may be an effective aid to the surgeon for beating heart mitral annuloplasty. We note, however, that a potential limitation of this study is that manual tasks were done with a (nonactuated) MCI, which is heavier than typical surgical tools.

The in vivo results also verify that safe, precise robotic force tracking is feasible inside the beating heart through the use of feed-forward target motion information in the controller. This approach enables the robotic system to operate at the motion bandwidth of the heart while simultaneously ensuring damping and providing good disturbance rejection. In contrast, a purely force feedback controller would require a bandwidth that is approximately 3.8 times higher than the heart motion bandwidth to have the same performance. Our analysis and laboratory experiments indicate that force control at such high bandwidth excites transverse vibrations in the robot that could lead to a variety of dangerous outcomes, including controller instability.

The difficulty in achieving a fast and stable force controller has been examined extensively by other researchers, typically in the industrial setting where large, multiple degree-of-freedom robots interact with stiff, nearly motionless surfaces. Their analyses found a number of fundamental sources for instability at high bandwidth, such as sampling time [10], actuator bandwidth limitations [12], force measurement filtering [12], actuator and transmission dynamics [13], and flexible modes in the robot arm [11], [13]. Recent in vivo force-control experiments using a multiple degree-of-freedom endoscopic robot in contact with liver indicate that arm dynamics can limit the controller bandwidth to the extent that it is not able to adequately reject slow respiratory motion [26]. For our system and application, the dominant source of instability is from the flexible modes in the surgical instrument. This is somewhat surprising, given the simplicity of our robot: basically a small, 1 degree-of-freedom actuator mounted with a stiff, nonarticulated rod as an end effector. However, our analysis and experiments confirm that these resonances should and do occur in our robot and are also likely reduced the deviations to approximately 25% of the manual case ($0.11 \pm 0.02$ N, $p = 0.017$). Statistical significance was also found between the force regulator and feed-forward controller conditions ($p = 0.009$). These results are summarized in Fig. 13. The third trial for the feed-forward force controller was omitted from statistical analysis because the animal showed reduced cardiac viability at the end of the experiment. The implications of this result and possible improvements to the surgical procedure are discussed in greater detail in Section VII.
to occur in instruments that are mounted to the standard surgical robots.

To overcome the bandwidth limitations imposed by vibration, we developed a system that exploits the quasi-periodicity in heart motion to generate feed-forward motion terms for the force controller. Similar approaches have been used for robotic position tracking of the beating heart. Independently, Ginhoux et al. [23] and Bebek and Cavusoglu [24] have demonstrated that the use of model predictive control can increase the effective tracking bandwidth and positioning accuracy of their multiple degree-of-freedom robots for coronary artery bypass graft procedures. Our previous work has focused on heart motion prediction to compensate for the time delays and noise inherent in 3-D ultrasound-guided, robotic intracardiac procedures [4], [20]. All these groups have demonstrated the in vivo feasibility of accurately positioning a robotic instrument relative to a beating heart surgical target. In this paper, we address the successive problem of tracking the heart while applying precise contact forces for surgical manipulation. We pursue this from the perspective of force control, and to the knowledge of the authors, our in vivo experiment is the first demonstration of such an approach within the beating heart. Approximately 10 s of contact is needed to securely place surgical anchors into the annulus. Our in vivo experiments demonstrate that the system can control to precise forces for up to at least 20 s.

A limitation of the current system is that the image-based updates of target position are stopped after contact is made because the interaction of the robot with the tissue causes the tissue to deviate from its desired trajectory. In cases of high heart rate variability or arrhythmia, the system would not perform well because tissue motion would not follow the feed-forward model predictions provided by the extended Kalman filter. This poor performance was observed in the third in vivo trial of the feed-forward force controller (see Table I). Although not pursued in this paper, a clinical workaround may exist for operating on a heart with low motion periodicity. The spontaneous beating of the heart can be slowed through the administration of drugs, and then the heart can be electrically paced at a fixed frequency [27]. This may mitigate both heart rate drift and arrhythmia, and we believe that it would have improved performance for the third in vivo trial. This proposed clinical approach is expected to work for most patients experiencing arrhythmia; however, as in all surgical procedures, some patients may be ineligible for this beating heart procedure based on the severity of preexisting conditions.

Finally, we note that there are alternatives to our approach of feeding-forward target motion in order to avoid vibration at high bandwidth. For instance, one could attempt to structurally reinforce the surgical instrument to shift the resonance to higher frequencies, use preshaped command inputs to avoid excitation of the resonance [28], actively control vibration [29], redesign the robot to have a macro–mini actuation scheme so that fast actions are located closer to the instrument tip [30], or use iterative learning control [31]. One could also place a three-axis position sensor on the tip of the instrument, build a nonlinear model to describe the axial-to-transverse coupling dynamics [17], and then attempt the control in a nonlinear controller.
Shelten G. Yuen received the B.Sc. degree in electrical engineering from the University of California, Davis, in 2001 and the S.M. degree in applied mathematics and the Ph.D. degree in engineering sciences from Harvard University, Cambridge, MA, in 2007 and 2009, respectively. From 2001 to 2005, he was with the Agilent Technologies, Inc., Santa Rosa, CA, and with Lincoln Laboratory, Massachusetts Institute of Technology, Lexington. He is currently a Research Scientist with Fitbit, Inc., San Francisco, CA. His research interests include applications of estimation, control, and robotics in medicine and health.

Douglas P. Perrin received the Ph.D. degree in computer science from the University of Minnesota, Minneapolis, in 2002. He is an Affiliate with the School of Engineering and Applied Science, Harvard University, Boston, MA, where his postdoctoral research includes surgical robotics, haptics, and medical imaging. He is also currently an Instructor in surgery with the Harvard Medical School, Boston, MA, where he is a member of the del Nido Research Group, Department of Cardiovascular Surgery, Children’s Hospital Boston. His research interests include image-guided beating heart surgery, ultrasound image enhancement, and patient-specific models for surgical plating.

Robert D. Howe (S’88–M’89–SM’07) received the B.Sc. degree in physics from Reed College, Portland, OR, and the Ph.D. degree in mechanical engineering from Stanford University, Stanford, CA, in 1990. He is currently an Abbott and James Lawrence Professor of engineering, an Associate Dean for Academic Programs, and an Area Dean for bioengineering with the Harvard School of Engineering and Applied Sciences, Harvard University, Cambridge, MA, where he is engaged in directing the Harvard BioRobotics Laboratory, which investigates the roles of sensing and mechanical design in motor control, in both humans and robots. His research interests include manipulation, the sense of touch, haptic interfaces, and robot-assisted and image-guided surgery.

Pedro J. del Nido received the M.D. degree from the Medical School, University of Wisconsin, Madison. He completed his training with Boston University Medical Center, Boston, MA, Toronto General Hospital, Toronto, ON, Canada, and the Hospital for Sick Children, Toronto. He is currently with the Department of Cardiovascular Surgery, Children’s Hospital Boston, Harvard Medical School, Boston. His research is directed toward understanding the metabolic and structural changes due to left ventricular hypertrophy. His clinical goal is to apply minimally invasive robotic surgery and other cutting-edge techniques to enhance cardiac surgery.

Nikolay V. Vasilyev received the M.D. degree from Moscow Medical Academy, Moscow, Russia. He completed training in cardiovascular surgery with the A.N. Bakoulev Center for Cardiovascular Surgery, Moscow. He is currently a Staff Scientist with the Department of Cardiac Surgery, Children’s Hospital Boston, Harvard Medical School, Boston, MA, where he is also an Instructor in surgery with the Division of Surgery. His research interests include the development of beating heart intracardiac procedures, in particular, new imaging techniques, computer modeling and simulation, and device design.