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3D Force Control for Robotic-Assisted Beating Heart Surgery Based on Viscoelastic Tissue Model

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***Abstract*— Current cardiac surgery faces the challenging problem of heart beating motion even with the help of mechanical stabilizer which makes delicate operation on the heart surface difficult. Motion compensation methods for robotic-assisted beating heart surgery have been proposed recently in literature, but research on force control for such kind of surgery has hardly been reported. Moreover, the viscoelasticity property of the interaction between organ tissue and robotic instrument further complicates the force control design which is much easier in other applications by assuming the interaction model to be elastic (industry, stiff object manipulation, etc.). In this work, we present a three-dimensional force control method for robotic-assisted beating heart surgery taking into consideration of the viscoelastic interaction property. Performance studies based on our D2M2 robot and 3D heart beating motion information obtained through Da Vinci^{TM} system are provided.**

I. Introduction

Minimally invasive surgery (MIS) has been widely adopted in medical intervention around the world due to its various advantages over traditional open surgery. Robotic-assisted surgery further sharpen the edge of MIS by overcoming human physical constraints in operation like tremor, accuracy, motion bandwidth etc. and proving centralized operating console and many auxiliary instruments which greatly increase the surgeon's dexterity and reduce the surgeon's burden during surgery.

For robotic-assisted surgery, beating heart surgery is still challenging due to the heart's fast and relatively large motion [1], [2]. Even though mechanical stabilizer could be used to constrain this motion, the residual motion is still significant [3] and the surgeon has to manually cancel it. Several methods have been proposed recently to address this motion compensation problem [2], [4], [5]. However, it is observed that except the methods addressing motion compensation there's hardly any research reported in literature devoted to force control for beating heart surgery except the very recent work [6].

Haptic or force feedback as a very natural and important supplementary information to the surgeon in tradition surgery is unfortunately missing in most surgical robotic systems including Da Vinci^{TM} which represents the state-of-the-art commercial robotic system for MIS. To address this problem, considerable research efforts have been devoted to developing techniques which render surgeon haptic feeling of presence or exert desired force on tissue [7]–[9] etc. These works present detailed study of the force interaction model with soft organ tissue, and different modeling methods have been proposed. As shown through experimental study in [10] biological tissues are not elastic and the history of strain affects the stress. In [10], a quasi-linear viscoelastic function is proposed to represent the stress-strain relationship. Accurate models for soft tissue simulations could be obtained using Finite Element Method (FEM) [11] but it is numerically time-consuming, especially for dynamic simulation or real-time applications. In [8], a polynomial function of second order model is used to describe pre-puncture phase of needle insertion, and for the same application purpose a model based on nonlinear Kelvin model is developed in [12]. A compact dynamic force model is presented in [13] where force is modeled using a nonlinear dynamic model. In [14], a viscoelastic model based on fractional derivative is presented which is quite accurate especially for relaxation phenomenon.

However, it is noticed that the models as proposed in the aforementioned works [7]–[14] are mainly focused on force control and interaction with static or slow motion tissue and hence are not sure to be suitable for real-time control of beating heart surgery in terms of computation time and control design complexity. In the work [6] which is dedicated to force control of beating heart surgery, the interaction between the robotic instrument and the heart wall is described with a simple elastic model which is not proper according to the observations in [10].

In this work, we propose a force control method for beating heart surgery by employing a suitable viscoelastic interaction model chosen through experimental evaluations which should be both accurate enough to achieve the force control task and computationally efficient for real-time implementations. As the most important factor to consider in surgery, the stability of the robotic system is justified rigorously through theoretical analysis to guarantee safety. To our knowledge, this is the first attempt to perform force control in robotic-assisted beating heart surgery based on realistic viscoelastic model. The system performance and robustness against model parameter mismatch is evaluated through simulation studies based on a real robot platform and heart beating data recorded via Da Vinci^{TM} system.

II. Viscoelastic Interaction Model for Soft Tissue

To choose the viscoelastic interaction model for our force control design, three main criteria are to be met considering the specific requirements of beating heart surgery: accuracy, complexity (computation and design), transient performance

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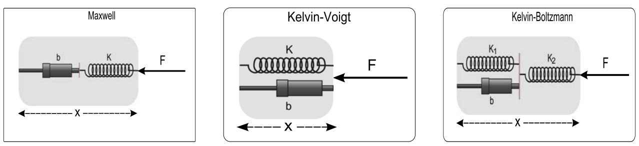
(fast heart motion). Then *in vitro* experiment studies have been conducted to evaluate the performance of different possible models and the one which fits best to our control task requirements is chosen in the force control design.

A. Possible Viscoelastic Model Candidates

Based on these criteria, we may first rule out FEM-based and high-order models due to their heavy computation and implementation burdens. We identify three classical linear viscoelastic models (Maxwell, Kevin-Voigt, Kelvin-Boltzmann) as illustrated in Fig 1 and one nonlinear viscoelastic model (fractional derivative [14]) from literature.



(a) (b) (c)

Fig. 1. Linear Viscoelastic Models

- Maxwell Model
  As seen from Figure 1(a), Maxwell model is composed of a spring and a damper in series and the interaction force F is described as:
  $$F(t) = k\frac{dx(t)}{dt} - \alpha\frac{dF(t)}{dt} \quad (1)$$
  where $x(t)$ denotes position deformation, k represents the spring stiffness constant and b the damping factor, $\alpha = \frac{b}{k}$.
- Kelvin-Voigt Model
  Kelvin Voigt model is composed of a spring and a damper in parallel and the interaction force F is described as:
  $$F(t) = b\frac{dx(t)}{dt} + kx(t) \quad (2)$$
- Kelvin-Boltzmann Model
  Kelvin Boltzmann model combines the Kelvin Voigt model with a spring in series and the interaction force F is described as:
  $$F(t) = \gamma x(t) + \beta\dot{x}(t) - \alpha\dot{F}(t), \quad (3)$$
  where $\alpha = \frac{b}{x_1+x_2}$, $\beta = b\frac{k_2}{k_1+k_2}$, $\gamma = \frac{k_1k_2}{k_1+k_2}$.
- Fractional Derivative Model
  This viscoelastic model could be described as:
  $$F(t) = k\frac{d^r x(t)}{d^r t} \quad (4)$$
  where k is the stiffness parameter and the derivative order r is set to 0.125 according to the experiments reported in [14].

B. Experiment Evaluation

To identify the best model, relaxation tests were performed, which consist of performing a position step input on the soft tissue and measuring the exerted force. In this work *in vitro* tests were performed. A piece of beef was used in the experiments.

The experiment evaluation was done using the D2M2 robot which has five degrees of freedom with direct drive technology providing fast dynamics and low friction (Fig 2). A force sensor (ATI Mini40) is attached to the end effector. For the relaxation tests simple PI controller has been implemented. The sampling period was fixed to $0.7ms$. The position and the measured force were collected and used to estimate the parameters of each model. An off-line least square algorithm was used for the estimation. Using the identified parameters, off-line simulations were performed applying the input information collected on experiments in each model equation. The output for each model is then compared with the real output to analize its prediction capabilities.

Fig. 2. *In vitro* relaxation test using D2M2 robot

C. Result Analysis

Several relaxation tests were performed using position steps with different amplitudes as input. For the sake of presentation clarity, only one relaxation tests with position input defined as $6.5mm$ is presented. After the off-line estimation process, using the collected data and the models equation, each model was simulated and the force outputs can be seen in Fig. 3.

To validate the model estimation a cross validation was performed. The position information from a new database, collected in a experiment different from the one used to estimate the models, is applied on the four models to assess the prediction capability of each model under unknown data. The real measured force and the forces estimated by the models can be seen in Fig. 4, as well as the error between the measured real force and the estimated one.

From the experiment result, it is seen that the Maxwell model converge to 0 very fast which is obvious against practical experience, the fractional derivative model presents good relaxing performance but its transient performance is not satisfactory. It should be noted here that for fast beating heart surgery, the transient performance of the interaction model is of great importance for force control since the contact surface keeps moving with high velocity. Kelvin-Boltzmann model and the Kelvin-Voigt model had the most realistic response, reaching approximately the final value and following the tissue dynamics, but the Kelvin-Boltzmann model provides the lowest error. Based on the experiment evaluation, it is shown that Kelvin-Boltzmann model outperforms other possible viscoelastic models in terms of both transient performance and accuracy. Therefore, Kelvin-Boltzmann model is chosen as the interaction model to develop our force control method.

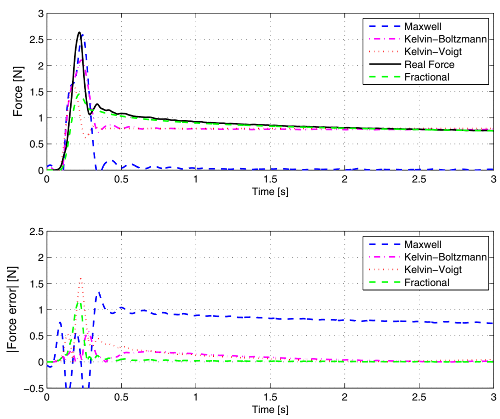


Fig. 3. Relaxation Test

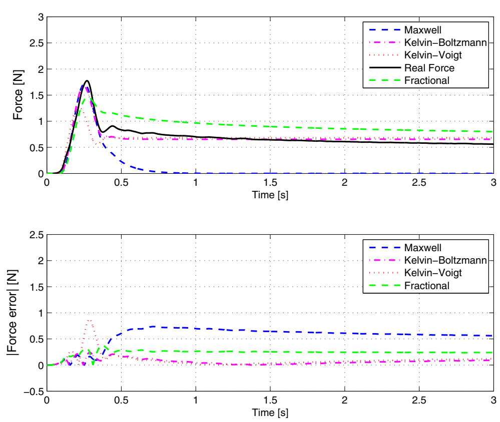


Fig. 4. Relaxation Test: Cross validation

III. Force Control Design Based on Viscoelastic Model

A. System Dynamics

The dynamic equation of the robot in the Cartesian space in contact with an environment is given by [15]:

$$M_{x_r}(x_r)\ddot{x_r} + C_{x_r}(\dot{x_r}, x_r)\dot{x}_r + g_{x_r}(x_r) = F_a - F_e \quad (5)$$

where x_r is the operational space coordinates of the robot end-effector, M_{x_r} is the operational space mass matrix, $C_{x_r}(\dot{x_r}, x_r)\dot{x}_r$ represents the vector of Coriolis and centripetal forces, $g_{x_r}(x_r)$ is the gravity term. F_a denotes the end-effector force due to joint actuation and F_e is the interaction forces due to contact with the environment.

B. Force Control Design and Stability Analysis

In the well-defined operating room (OR) environment, it is possible to obtain precise information of the system dynamics through careful pre-calibration. And with the available force feedback signal F_e obtained through the force sensor, the nonlinear dynamic system (5) could be decoupled by designing the robot force F_a as

$$F_a = F_e + \hat{M}_{x_r}(x_r)u + \hat{C}_{x_r}(\dot{x_r}, x_r)\dot{x}_r + \hat{g}_{x_r}(x_r) \quad (6)$$

where $\hat{M}_{x_r}(x_r)$, $\hat{C}_{x_r}(\dot{x_r}, x_r)$ and $\hat{g}_{x_r}(x_r)$ are respectively the estimation of F_e, $M_{x_r}(x_r)$, $C_{x_r}(\dot{x_r}, x_r)$ and $g_{x_r}(x_r)$ and u represents the auxiliary control signal. Hence the desired decoupled system could be expressed as

$$\ddot{x}_r = u. \quad (7)$$

which presents a unity mass system along each Cartesian dimension.

For simplicity of technical development and without loss of generality, we assume that the initial position of robot end-effector is on the heart surface ($F_e = 0$). Then the interaction model (3) could be rewritten under this framework as

$$F_e = \gamma x + \beta\dot{x} - \alpha\dot{F}_e, \quad (8)$$

where $x = x_r + x_h$ is the interaction motion of robot end-effector x_r and the heart beating x_h.

Differentiate both sides of above equation, we have

$$\alpha\ddot{F}_e + \dot{F}_e = \gamma\dot{x} + \beta\ddot{x}, \quad (9)$$

and substituting the linearized system (7) it has

$$\alpha\ddot{F}_e + \dot{F}_e = \gamma\dot{x} + \beta(u + \ddot{x}_h), \quad (10)$$

Denoting the desired constant force to be exerted on the heart surface as F_d, the auxiliary control input u is designed as

$$u = -\ddot{x}_h - \frac{\gamma}{\beta}(\dot{x}_r + \dot{x}_h) - \frac{a_1}{\beta}\Delta F - \frac{a_2}{\beta}\dot{F}_e \quad (11)$$

where $\Delta F = F_e - F_d$ and a_1, a_2 are positive constant control parameters.

Since desired force F_d is constant it has $\dot{\Delta F} = \dot{F}_e$, then by using the auxiliary control input u as defined in (11) we have (10) as

$$\alpha\Delta\ddot{F} + (1 + a_2)\Delta\dot{F} + a_1\Delta F = 0, \quad (12)$$

By defining a function y as

$$y = \alpha\Delta\ddot{F} + (1 + a_2)\Delta\dot{F} + a_1\Delta F, \quad (13)$$

the transfer function of the force tracking system is obtained as

$$\frac{\Delta F(s)}{Y(s)} = \frac{1}{\alpha s^2 + (1 + a_2)s + a_1}. \quad (14)$$

Since α, $1 + a_2$ and a_1 are all positive the system is guaranteed to be asymptotic stable, which means the force tracking error ΔF converges to 0 asymptotically and hence the force control task is achieved.

In order to achieve critically damping, i.e. with 0 overshoot in force tracking to guarantee safety, the control parameters a_1, a_2 should be chosen as $a_1 = \frac{(1+a_2)^2}{4\alpha}$. In the scenario of heart beating surgery, higher control parameters a_1, a_2 could be used to get higher undamped natural frequency in order to fasten the transient response of the control system without affecting critical damping ratio.

Remark: It is noted that in the design of auxiliary control u, derivative of measured force signal $\dot{F}_e$ is employed. In practical implementation, the force measurement signal F_e is often noisy and hence $\dot{F}_e$ is not reliable. By using the Kelvin-Boltzmann model as in (8), we may replace $\dot{F}_e$ with

$$\begin{aligned} \dot{F}_e &= -\frac{1}{\alpha}F_e + \frac{\gamma}{\alpha}x + \frac{\beta}{\alpha}\dot{x} \\ &= -\frac{1}{\alpha}F_e + \frac{\gamma}{\alpha}(x_r + x_h) + \frac{\beta}{\alpha}(\dot{x}_r + \dot{x}_h). \quad (15) \end{aligned}$$

Then we get the auxiliary control input u in another form as

$$u = \frac{a_2}{\alpha\beta}F_e - \ddot{x}_h - \frac{\gamma a_2}{\alpha\beta}(x_r + x_h) - (\frac{\gamma}{\beta} + \frac{a_2}{\alpha})(\dot{x}_r + \dot{x}_h) - \frac{a_1}{\beta}\Delta F. \quad (16)$$

In this form, only directly measured force signal F_e and position information are used, and the force derivative signal $\dot{F}_e$ is avoided. x_r could be accurately obtained from robot joint encoders and heart motion information x_h is pre-known or could be online estimated or predicted using motion compensation techniques as in [2], [4], [5] etc. The auxiliary control input u in (16) is exactly identical with the one in (11) in theory, but it is easier and more feasible to implement in practice.

Thus, the robot force F_a is designed with the auxiliary control input u as

$$\begin{aligned} F_a = F_e + \hat{M}_{x_r}(x_r)[&\tfrac{a_2}{\alpha\beta}F_e - \ddot{x}_h - \tfrac{\gamma a_2}{\alpha\beta}(x_r + x_h) \\ &-(\tfrac{\gamma}{\beta} + \tfrac{a_2}{\alpha})(\dot{x}_r + \dot{x}_h) - \tfrac{a_1}{\beta}\Delta F] \\ &+\hat{C}_{x_r}(\dot{x_r}, x_r)\dot{x}_r + \hat{g}_{x_r}(x_r) \quad (17) \end{aligned}$$

which guarantees the desired force F_d be exerted on the beating heart with the help of force feedback.

IV. Performance Evaluation through Simulation Studies

To evaluate the control performance of the proposed force control method and the system's capability to handle disturbance and model parameter mismatch, simulation studies have been carried out based on the D2M2 robot model with all dynamic and kinematic parameters calibrated from the real robot and using *in vivo* heart beating motion data recorded through Da Vinci^{TM} system.

The three-dimensional heart beating motion data are illustrated as in Fig 5. Since the robot end-effector just touches a specified point on the beating heart, no torsion is involved in the force measurement hence it's reasonable to assume that the force measurements for three axes are non-coupled. According to the experiments introduced in Section II, the Kelvin-Boltzmann model identified and used in the simulation study is

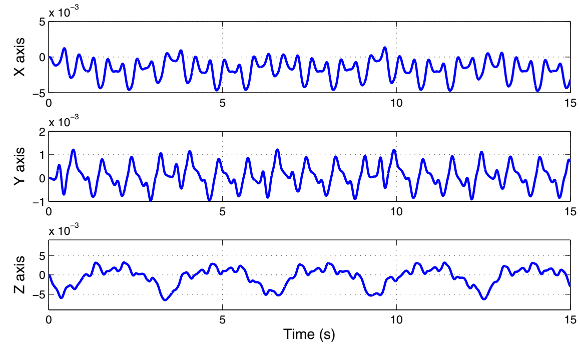


Fig. 5. 3D heart motion

$$F_e = 100.65x_r + 30.705\dot{x}_r - 0.0567\dot{F}_e. \quad (18)$$

The control parameters in the robot force control input are set to $a_1 = 30$, $a_2 = 1$. To fully explore the capability of the proposed control method, the simulation is composed of 3 phases: 0-3 second, a step force command with amplitude of 4 N is sent to the robot at time 1 second without presence of the beating heart motion disturbance yet; 3-6 second, the heart beating disturbance as in Fig 5 is introduced into the interaction force between the robot and tissue; 6-15 second, pulse series of amplitude of 4 N are added in the force command with the presence of heart beating disturbance. Simulation results are illustrated in Fig 6. It is seen that the contact forces along all 3 axes converge fast to desired values in different phases even with the fast heart beating disturbance.

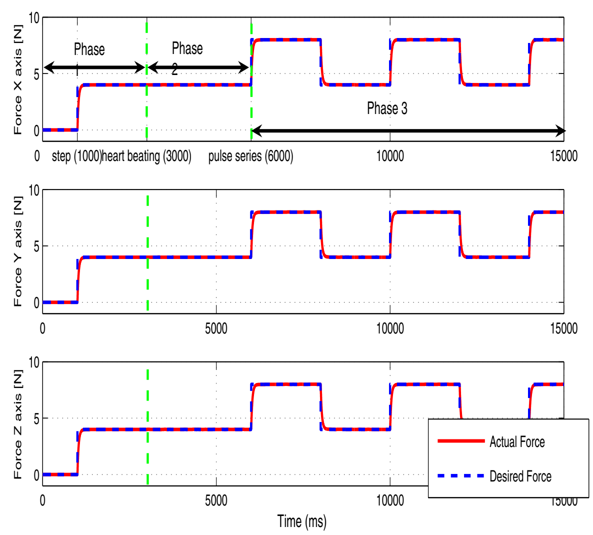


Fig. 6. Force Tracking Performance for X, Y, Z axis

To test the robustness of proposed method against force measurement noise and interaction model mismatch, the same control task is performed as in previous simulation study. White noises of maximum amplitude $\pm 1.2N$ are introduced in the force measurement loop, also the interaction model is estimated with errors as

$$F_e = 150x_r + 40\dot{x}_r - 0.1\dot{F}_e. \quad (19)$$

Force tracking performances are illustrated in Fig 7 and it is shown that even with the presence of measurement noise and model parameter mismatch the proposed force control method still work well to achieve the control task and thus exhibits its robustness.

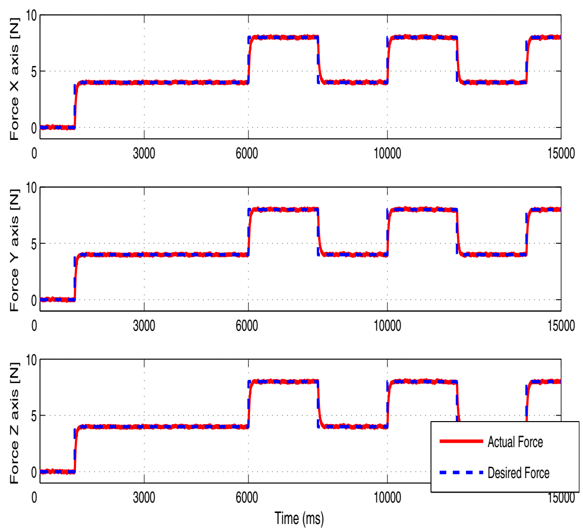


Fig. 7. Robust Force Tracking Performance

V. Conclusions and Future Works

This paper presents the work of force control design for robotic-assisted beating heart surgery based on viscoelastic interaction model. To the best of our knowledge, this is the first attempt to perform robot force control on the beating heart using a realistic biological tissue interaction model. The viscoelastic model is chosen and evaluated through *in vitro* experiments. The stability of developed control method is analyzed rigorously by theory. Simulation studies based on real robotic system setup are conducted to confirm the effectiveness the developed control method and show its robustness against measurement disturbance and model uncertainties. For future works, first the developed control method is to be implemented for *in vitro* experiments in lab tests which are currently undergoing. Then the method will be further improved by including real-time heart motion modeling techniques. In the end, it is planned to carry out *in vivo* experiments to test its feasibility and find out potential constraints for practical clinical applications.

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