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Hunt–Crossley model based force control for minimally invasive robotic surgery

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ABSTRACT

In minimally invasive surgery (MIS) the continuously increasing use of robotic devices allows surgical operations to be conducted more precisely and more efficiently. Safe and accurate interaction between robot instruments and living tissue is an important issue for both successful operation and patient safety. Human tissue, which is generally viscoelastic, nonlinear and anisotropic, is often described as purely elastic for its simplicity in contact force control design and online computation. However, the elastic model cannot reproduce the complex properties of a real tissue. Based on *in vitro* animal tissue relaxation tests, we identify the Hunt–Crossley viscoelastic model as the most realistic one to describe the soft tissue's mechanical behavior among several candidate models. A force control method based on Hunt–Crossley model is developed following the state feedback design technique with a Kalman filter based active observer (AOB). Both simulation and experimental studies were carried out to verify the performance of developed force controller, comparing with other linear viscoelastic and elastic model based force controllers. The studies and comparisons show that the Hunt–Crossley model based force controller ensures comparable rise time in transient response as the controller based on Kelvin–Boltzmann model which is reported as the most accurate description for robot-tissue interaction in recent literature, but it causes much less overshoot and remains stable for tasks with faster response time requirements.

1. Introduction

Minimally invasive surgery (MIS) is replacing traditional open surgical procedures which normally involve large incision to access the patient's body. MIS is performed using long instruments to enter the patient body through small incisions and leads to direct advantages including less pain, hemorrhaging and trauma, reduced risk of infections, shortened hospital stay, and hence less burden for both the patient and social health care system [1–3].

With the development of technology in the area of robotics, robotic devices have found their way to the operating room (OR) and lead to the new concept of minimally invasive robotic surgery (MIRS). Many robotic surgical devices have been developed for MIS operations in the literature [4–9] with the most widely used and best known example of da Vinci robot from Intuitive Surgical (Sunnyvale, CA, USA). The movements of the robotic devices, commanded by the surgeon, can mimic the motion of human hands inside patient body to accomplish safer and more precise operations. However, as pointed out in [10], the surgeon can not retain the haptic feeling of the interaction with tissue due to lack of direct contact with the working site and hence the amount of applied force on tissue surface cannot be accurately controlled.

Execution of proper contact force is necessary and even essential for many surgical operations [11–13]. For tasks like suturing and pre-tensioning, excessive force leads to tissue damage and too low force cannot make the tasks successful. For haptic teleoperation with force command, operating transparency can be obtained only when the desired interaction force is accurately generated between robotic tool and tissue. Moreover, using force control allows to perform the same operation with higher precision and dexterity by reducing human errors [14].

In the MIRS scenario, one has to control the contact force between robotic surgical tools and soft tissues (i.e. muscles, organs, veins, arteries, etc). To design the force controller, a proper tool-tissue contact model is required. In literature, however, force control methods are mainly based on pure elastic contact model which is easy to implement and applies to most hard contact cases.

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Even some existing works on force control for surgical applications assume an elastic contact model [15]. Unfortunately, this model is not suitable to describe the contact with soft human tissues which exhibits complex properties such as nonlinearity, viscoelasticity, anisotropy, etc [16].

Over the last decades, several compliant contact models have been proposed to describe the behavior between contact bodies [17]. Combination of linear springs and dampers is one way to describe the viscoelastic behavior of the contact force, although these approaches show physical inconsistencies in terms of power exchange during contact [18]. Such kind of linear contact models include Maxwell model, Kevin–Voigt model, Kelvin–Boltzmann model, etc. Recent works show that robot-tissue force control based on Kelvin–Boltzmann model outperforms control methods based on traditional elastic model in terms of rise time and stability [19]. On the other hand, nonlinear models, such as the Hunt–Crossley model, are expected to be more accurate for describing the real behavior during the contact with soft human tissues [18,20]. However, so far nonlinear models have only been used for parameter estimation [18,21,22], but not for force control design.

This paper presents the first attempt to adopt the Hunt–Crossley (HC) model for the control design of contact force between soft tissue and robotic tool in literature. The control design utilizes the active observer (AOB) technique which helps to compensate parameter and modeling mismatches during control. Stability of the HC model based force control system is analyzed, and the advantage of this approach with respect to the methods based on linear contact models is shown through both numerical simulation and *in vitro* experimental studies.

The rest of the paper is organized as follows: Section 2 introduces the brief background on soft tissue contact models reported in literature; Section 3 presents a comparison study between linear and nonlinear viscoelastic contact models through *in vitro* relaxation tests, Section 4 describes the development of a force control method based on the identified Hunt–Crossley model, Sections 5 and 6 show the performance comparisons between linear contact model based force controllers and Hunt–Crossley model based force controller through simulations and experiments, Section 7 summarizes the work reported in this paper.

2. Soft tissue contact models

Several models have been developed in literature to describe the viscoelastic behavior of soft tissues [23]. The most complete study on viscoelastic tissue model is addressed in [16], where a quasi-linear viscoelastic (QLV) model is proposed to represent the stress–strain relationship as follows

$$F(t) = \underbrace{G(t)\sigma^e[\varepsilon(0)]}_{I} + \underbrace{\int_0^t G(t-\tau)\frac{\partial \sigma^e(\tau)}{\partial \tau}d\tau}_{II} \tag{1}$$

where $F(t)$ denotes the contact force; σ^e, ε denotes the instantaneous elastic stress and strain respectively; $G(t)$ is the reduced relaxation modulus. This modeling function is composed of two parts: the first part (I) is the instantaneous stress response and the second (II) gives the stress related to the past history [16]. Although accurate for off-line analysis, this model is complex and difficult to be used for contact force control design.

One simple and intuitive way to describe the interaction between robotic tools and soft tissues is to analytically build the force–displacement relationship. Analytical models are usually presented as a combination of springs and dampers [16], and are defined by the following components: the exerted force by the tissue, $F_e(t)$, when a strain is applied; the indentation (or penetration), $x(t)$, computed as the amount of displacement of the tissue from the rest position; the velocity of the deformation $\dot{x}(t)$; the elastic and damping coefficients K and b respectively.

Following this modeling method, several linear models have been developed. The first model, often used in traditional force control [24], is the elastic model (Fig. 1(a)) described by

$$F_e(t) = Kx(t). \tag{2}$$

The Maxwell (MW) model is represented by the series of a spring and a damper (Fig. 1(b)) and is expressed as

$$F_e(t) = b\dot{x}(t) - \alpha \dot{F}_e(t) \tag{3}$$

where $\dot{F}_e(t)$ is the derivative of the exerted force and $\alpha = b/K$.

The Kelvin–Voigt (KV) model consists of a spring in parallel with a damper (Fig. 1(c)) and is described by

$$F_e(t) = Kx(t) + b\dot{x}(t). \tag{4}$$

Another viscoelastic model is the Kelvin–Boltzmann (KB) model which is obtained by adding a spring in series to Kelvin–Voigt model (Fig. 1(d)) and its characteristic equation is given by

$$F_e(t) = Kx(t) + \eta\dot{x}(t) - \gamma\dot{F}_e(t) \tag{5}$$

where $K = k_1k_2/(k_1+k_2)$, $\eta = bk_2/(k_1+k_2)$, $\gamma = b/(k_1+k_2)$ with k_1, k_2 and b denoting the elastic and damping coefficients respectively.

The above linear models may apply to contacts with objects of linear and homogeneous properties, but physical limitations can be observed when contact with soft tissue is considered. As illustrated in Fig. 2, during the contact between a rigid tool and the soft tissue, two phases can be identified: the first one, corresponding to loading, takes place at starting contact time t_0 and ends at t_{Max} when the maximum displacement in the soft tissue x_{max} is reached; the second one, corresponding to unloading, takes place from t_{max} to the instant t_{Final} when the tool and the soft tissue separate. Combining the loading and unloading behavior, an hysteresis loop can be defined as the force–displacement relationship for soft tissue during the contact.

The power flow during the contact is calculated by $P(t) = F_e(t)\dot{x}(t)$ as plotted on the right side of each subfigure in Fig. 3. For linear viscoelastic models, the dissipated energy ΔH, represented by the area enclosed by the hysteresis loop, can be computed as the algebraic sum of the energies H_1, H_2 and H_3, which are calculated as integrations of power flow for different periods as shown in Fig. 3(a) [18]. Linear viscoelastic models show the same behaviors

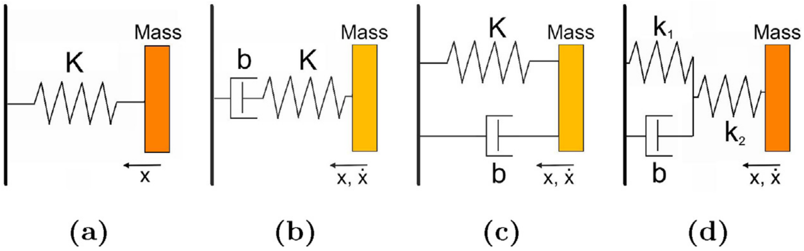

Fig. 1. Linear contact models: (a) elastic, (b) Maxwell, (c) Kelvin–Voigt, (d) Kelvin–Boltzmann.