Enhancement of Human Operator’s Perceptual Sensitivity for Telesurgical Systems via Polytopic System Approach

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Abstract—A suturing including knot tying is one of the more difficult operations to learn in telesurgical systems. Apprentice surgeons commonly suffer from suture breakage or knot failure. The difficulty, generally, comes from the absence of feedback of interaction force cues in a medical device (e.g., a needle and a thread). Even if there is haptic feedback to the operator, the operator may have a difficulty to detect a specific force such as suture breakage force. To deal with this problem, we propose a control method which can detect a suture breakage force more sensitively by considering human perception characteristics. A performance objective of the control method is designed according to the human perceptual factor, just noticeable difference. By convex optimization of the performance indices, a stabilizing $H_\infty$ controller is proposed for the telesurgical system. Finally, the proposed control scheme is validated via a simulation study.

I. INTRODUCTION

Numerous research efforts on haptic feedback in telesurgical systems have been conducted during past decades. It has been widely accepted to many surgeons that haptic feedback improves overall performance [1], [2]. The haptic feedback function, however, is not available in commercial telesurgical systems (e.g., da Vinci Surgical System) and it incurs difficulties especially in suturing and knot tying tasks [3].

In spite of the benefits of haptic feedback, it is still controversial to employ haptic feedback function in the telesurgical systems since there are still practical and technical issues/difficulties to implement the haptic feedback. Yuh et al. conducted a consecutive research about sensory substitutions of force information on knot tying task [3], [4]. They compared the direct haptic feedback case with other sensory substitution ones using applied forces on the subject during the task. Interestingly, the sensory substitutions showed a similar and even better performance than the direct force feedback. This is an interesting finding because it leads us to the importance of giving a warning of suture breakage to surgeon. They provided the adequate method to the surgeon to detect a specific force in the case of the sensory substitutions via color mapping or on-off aural mapping but there was no additional cue except interacting force in the haptic feedback case. This study shows, in other words, that it is very difficult to detect the breakage force only with the conventional haptic feedback due to human’s perceptual sensitivity of force. From this perspective, we have a strong motivation of developing haptic feedback control which can enhance the human operator’s sensitivity in detecting a specific force.

The control objective of the teleoperation system has gradually changed from without considering human factors to considering human factors. Conventional performance objectives such as transparency and position/force tracking abilities do not take account of human factors. Çavuşoğlu et al. proposed a new performance index, fidelity, considering human psychophysical characteristics [5]. There have been several studies [6], [7] about defining new performance measures with human characteristics and how to optimize a controller with the measures based on [5]. Although there are several studies of force scaling based on psychophysical thresholds [7], [8], control objectives are not designed for the specific task. For example, it is suitable to use the control scheme in [7] to increase an operator’s entire perception region but it is impossible to enable the surgeon to detect the suture breakage force.

In this paper, a novel haptic feedback control is proposed to detect a specific force. In contrast to transparency, in the proposed control, the desired impedance which should be transmitted to the operator is not the environment impedance. The desired impedance is designed according to human perceptual factor and environment impedance. To handle this performance objective systematically, the overall teleoperation system is modeled as a polytopic system. The well-known gain-scheduling control scheme [9], [10] is adopted to satisfy the performance of the system while guaranteeing the stability.

II. PRELIMINARIES

A. Surgical Suture

The da Vinci® surgical system of Intuitive Surgical® provides one camera and three robotic arms for slave [11]. These three robotic arms are controlled by the master device with both hands and a foot clutch to change the robotic arms. Robotic surgical procedures are generally composed of several operations such as suturing, electrocautery (cutting and coagulation), grasping, and so on. Because the system does not provide the haptic feedback, suturing is one of the more difficult skills to learn among these operations. Although the experienced surgeons may be able to recognize
the tension based on their experiences according to deformations of human organ, severe risks still remain for apprentice surgeons in terms of suture breakages or knot failures.

A surgical suture is a product containing a needle and a thread. It is mainly used to reconstruct anatomical structures. We conduct preliminary experiments to observe the material properties of the surgical suture. The thread of the surgical suture\(^1\) behaves as an elastic material with constant stiffness as shown in Fig. 1. The average stiffness and the breakage force of the material is approximately 300 N/m and 30 N, respectively. Therefore, we can employ a simple model to represent the surgical suture with Hooke’s law and constant stiffness \(k_{\text{env}}\).

**B. Human’s Perceptual Sensitivity**

The kinesthetic perception of human not only relies on processing information regarding the forces generated by muscles and the associated movements of limbs, but also uses this information to derive cues about other mechanical impedance variables such as stiffness, viscosity, and inertia [12]. Especially, in telesurgery, the impedance is one of the most important stimuli for perceiving dynamic changes in the environment. This measure contains information about changes in the position, force, and mechanical properties. Accurate perception of mechanical properties is, therefore, important in an environment composed of internal organs and tissues [7].

It has been reported that there is a loss in perceptual sensitivity when force and displacement, velocity, and acceleration are combined to perceive the stiffness, viscosity, and inertia, respectively. The perceptual sensitivity for the impedance variables are considerably lower than the perceptual sensitivity measured for limb movement, position and force [12], [13], [14]. A measure of the perceptual sensitivity is commonly referred to as the just noticeable difference (JND) [15]. The JND represents the ratio of the smallest amount of stimulus change required to produce a change in sensation in a discrimination task over the stimulus intensity. The JNDS for perception of the impedance variables are known as \(\sim 22\%\), \(14 \sim 34\%\), and \(21 \sim 113\%\) for stiffness, viscosity, and inertia perceptions, respectively, while it is known as \(5 \sim 9\%\) and \(5 \sim 12\%\) for perception of position and force, respectively [12].

**III. METHODS**

**A. Increasing of Breakage Force Detection Ability by Transmitted Impedance Design**

As we discussed in the previous section, humans can discriminate two different environments only if the stimulus difference is larger than the stimulus’ JND. The material property of surgical suture can be drawn in an interaction force-stiffness diagram as shown in Fig. 2. It is worthy to note that the stiffness beyond the human’s perceptual sensitivity of the stiffness (represented as the shaded region in Fig. 2) is not possible to detect.

If the force is applied over a certain force threshold the surgical suture is eventually broken. The breakage force is denoted as \(f_{\text{brk}}\). It would be nice if the teleoperator allows interacting force \(f\) as high as possible, but less than \(f_{\text{brk}}\) for the surgeon’s high dexterity. This, however, may increase a potential risk of rupture because the material property varies with humidity, temperature, surface finish and so on. We, therefore, defined a maximally allowable interacting force as \(f_{\text{lim}} < f_{\text{brk}}\), which has to be detected by the surgeon. Our control objective is to let the surgeon detect \(f_{\text{lim}}\) accurately and sensitively to maintain \(f\) smaller than \(f_{\text{brk}}\).

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\(^1\)ETHICON PDS II, polydioxanon, violet monofilament, 2-0 3 Ph. Eur., ETHICON LLC.
The transparency objective is aimed to make the transmitted impedance to the operator, $Z_{to}$, as the same as the environmental impedance, $Z_{env}$. It would be impossible to detect $f > f_{lim}$ if $Z_{to}$ belongs to the shaded region in Fig. 2, which represents the set of undiscernible stiffness. The conventional transparency objective, from this perspective, could not be used to detect $f_{lim}$ accurately and in sensitive with $k_{env}$.

In this paper, therefore, we design a desired impedance to be transmitted to the operator, $Z_d$, as the following to achieve the control objective $Z_{to} = Z_d$.

$$Z_d = \begin{cases} Z_{env} & \text{if } f < f_{lim} \\ Z_{disc} & \text{if } f \geq f_{lim} \end{cases}$$

(1)

where $Z_{disc}$ is a discriminative impedance from $Z_{env}$ which is defined as $Z_{disc} > (1 + JND)Z_{env}$ based on Fig. 2. Eq. (1) represents that $Z_{to}$ is defined as the same with the transparency if the interacting force, $f$, is smaller than $f_{lim}$ while it becomes $Z_{disc}$ which is larger than $(1 + JND)Z_{env}$ if $f \geq f_{lim}$ to detect $f_{lim}$. Note that the surgeon perceives the artificial (or, perhaps, incorrect) impedance, $Z_{disc}$, while she/he can detect $f_{lim}$ accurately. It is also noticeable that it is possible to design $Z_{disc} < (1 - JND)Z_{env}$ for the better detection of $f_{lim}$ than the transparency as well. In this design, however, the surgical suture could be broken due to the human operator’s perception on the difference environmental impedance which is smaller than the original one.

Note that, in this paper, we only consider stiffness for the environmental impedance $Z_{env}$, which is represented as $k_{env}$ so that $k_{disc}$ and $k_d$ were defined to represent the stiffness term of $Z_{disc}$ and $Z_d$, respectively.

The desired stiffness $k_d$ is designed based on (1) as shown in Fig. 3. Then, the relationship between the interacting force $f$ and position $x$ acting on $k_d$ is shown in (2).

$$f = \begin{cases} k_{env}x & \text{if } x < x_{lim} \\ k_{env}x_{lim} + k_{disc}(x - x_{lim}) & \text{if } x \geq x_{lim} \end{cases}$$

(2)

where $x_{lim}$ is easily defined as $f_{lim}/k_{env}$. Eq. (2) can also be formulated $f = k_d(x)x$ where

$$k_d(x) = \begin{cases} k_{env} & \text{if } x < x_{lim} \\ k_{disc} - (k_{disc} - k_{env}) \frac{x_{lim}}{x} & \text{if } x \geq x_{lim} \end{cases}$$

(3)

The lower and upper bounds of $k_d(x)$ can be set with (3).

$$k_d(x) \in [k_{env}, k_{disc}]$$

(4)

Because $k_d(x)$ is continuously varying between $k_{env}$ and $k_{disc}$, it is not easy to obtain the optimized control for $f - k_d(x)x$. In the next section, we present how to design a control to satisfy $Z_{to} = Z_d$ with (1) while maintaining the stability of the system.

B. Polytopic Representation of Telesurgical System

In this paper, we consider simplified problem for one hand and one robotic arm. The $n$-DOF master device is generally modeled with second-order nonlinear dynamics as follows.

$$M_m(X_m) \ddot{x}_m + C_m(X_m, \dot{X}_m) \dot{x}_m + G(X_m) = F_m + U_m$$

(5)

where $X_m \in \mathbb{R}^n$ is a position vector of the master device, $F_m \in \mathbb{R}^n$ is a human operator force vector, $U_m \in \mathbb{R}^n$ is a control input vector for the master device, $M_m$ is the inertia matrix, $C_m(X_m, \dot{X}_m)$ represents Coriolis and centrifugal force, and $G(X_m)$ is the gravity vector. Previous studies of [16], [17], [18] contain the adaptive feedback linearizing control laws. As a result of adaptive control and linearization, (5) can be written as

$$m_m \ddot{x}_m + b_m \dot{x}_m + k_m x_m = f_m + u_m + \eta_m,$$

(6)

where $m_m, b_m, \text{and } k_m$ are mass, damping, and stiffness coefficients of the master device with adaptive controls, and $x_m$ is the position; $u_m$ is the control input and $\eta_m$ is the error term. The details of notation and theoretical descriptions can be found in [17]. Because the dynamics in (6) become easily decoupled in different axes of motion, we are dealing with the one degree of freedom problem for simplicity.

The dynamics of a slave robot are also defined as

$$m_s \ddot{x}_s + b_s \dot{x}_s + k_s x_s = f_s + u_s + \eta_s,$$

(7)

where $m_s, b_s, \text{and } k_s$ are mass, damping, and stiffness of the slave robot with adaptive control, and $x_s$ is the position; $u_s$ is the control input and $\eta_s$ is the error term. Transmission time delay is beyond the scope of this paper because it is negligible if the master device and the slave robot are located in a same operating room.

The human operator is simply modeled with damping $b_{op}$ and stiffness coefficient $k_{op}$. The human operator generates $f_{op}$ so that $f_m$ is the transmitted force to the master device by the human operator.

$$f_m = f_{op} - b_{op} \dot{x}_m - k_{op} x_m$$

(8)

As already discussed the surgical suture in previous section, a surgical thread is modeled with stiffness coefficient $k_{env}$ as $f_s = k_{env} x_s$. Eq. (7) is reformulated as follows.

$$m_s \ddot{x}_s + b_s \dot{x}_s + k_s x_s = k_{env} x_s + u_s + \eta_s,$$

(9)

The following equations are obtained from the model of each component of the teleoperation system.

$$\begin{align*}
\dot{x}_m &= \frac{1}{m_m} \left( f_{op} + \eta_m - (b_{op} + b_m) \dot{x}_m - (k_{op} + k_m) x_m + u_m \right) \\
\dot{x}_s &= \frac{1}{m_s} \left( \eta_s - b_s \dot{x}_s - (k_s - k_{env}) x_s + u_s \right)
\end{align*}$$

(10)

The position and shaped impedance tracking errors are defined as control objectives. The difference between transmitted impedance $Z_{to}$ and shaped impedance $Z_d$ is impossible to represent linearly so that the impedance tracking error is represented with $(Z_{to} - Z_d) \cdot \dot{x}_m$. The position tracking error is also considered as follows.

$$\begin{align*}
e_z &= f_m - k_d(x_m)x_m \\
e_p &= x_s - x_m
\end{align*}$$

(11)
It is not easy to set proper values for the weighting functions to minimize the tracking errors. If the weighting functions are not designed properly, the control objectives could be biased for either shaped impedance tracking error or position tracking error. The weighting functions, $W_z$ and $W_p$, are used for $e_z$ and $e_p$, respectively. Penalty functions $W_{pnl}$ are also included in the control objectives to prevent infinitely large control actions of $u_m$ and $u_s$.

The stiffness coefficient $k_d$ is dependent on the interaction position $x_s$. Because $x_s$ is a function of time, $k_d$ is possible to be considered as a time-varying parameter. We choose to model the time-varying parameters of the environment coefficients in the teleoperation system with $\theta$ as given in (12). The parameter $\theta(t)$ is automatically computed upon the transmitted impedance design and according to the position of slave.

$$\theta(t) = [k_d(x_s)]$$

(12)

To compute the controller of the system, $k_d(x_s)$ is assumed to have its value bounded with upper limit $k_{disc}$ and lower limit $k_{env}$.

$$k_d(x_s) \in [k_{env}, k_{disc}]$$

(13)

The state space equation of the teleoperation system with time-varying parameters are obtained as given below.

$$\dot{X} = AX + B_1w + B_2u$$

(14)

The exogenous input to the system, $w$, consists of the active human force and the environment force estimation errors while the control input, $u$, includes the actuator torques of the master and the slave. The measurement, $y$, for position-position based control architecture as shown in Fig. 4 is represented as follows.

$$\begin{align*}
    w^T &= \begin{bmatrix} f_{op} \eta_m \eta_s \end{bmatrix} \\
    u^T &= \begin{bmatrix} u_m u_s \end{bmatrix} \\
    y^T &= \begin{bmatrix} x_m x_s \end{bmatrix}
\end{align*}$$

(15)

### TABLE I

**Designed Weighting Functions for Controller Design**

<table>
<thead>
<tr>
<th>Weighting function</th>
<th>Design</th>
</tr>
</thead>
<tbody>
<tr>
<td>$W_p$</td>
<td>$x^2 + 1.0(\gamma + 40)\theta$ and $x^2 + 2.0(\gamma + 400)\theta^2$ for $\theta = 0.01^3$ and $\theta = 0.1^3$</td>
</tr>
<tr>
<td>$W_{pnl}$</td>
<td>$x^2 + 2.0 + 400\theta$ and $x^2 + 4.0 + 400\theta$ for $\theta = 0.01^3$ and $\theta = 0.1^3$</td>
</tr>
</tbody>
</table>

The state vector $X$ has four states as shown below in (16).

$$X^T = \begin{bmatrix} \dot{x}_m \ x_m \ \dot{x}_s \ x_s \end{bmatrix}$$

(16)

The controlled output $q$ is formulated with $X$, $w$, and $u$ as shown in (17).

$$q = \begin{bmatrix} e_z \\ e_p \\ u_m \\ u_s \end{bmatrix} = C_1(\theta)X + D_{11}w + D_{12}u$$

(17)

Weighting functions are then multiplied to the controlled outputs before designing the controller as given in (18).

$$W_q \cdot q = \text{diag}(W_z, W_p, W_{pnl}, W_{pnl}) \cdot \begin{bmatrix} e_z \\ e_p \\ u_m \\ u_s \end{bmatrix}$$

(18)

The measured signals, $y$, are therefore, written as follows.

$$y = C_2X = \begin{bmatrix} 0 & 1 & 0 & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix} X$$

(19)

The controlled teleoperation system is, finally, rewritten as follows.

$$\begin{align*}
    \dot{X} &= AX + B_1w + B_2u \\
    q &= C_1(\theta)X + D_{11}w + D_{12}u \\
    y &= C_2X
\end{align*}$$

(20)

The teleoperation system is expressed as a convex combination of polytope as shown in (21) since the parameter dependence of the teleoperation system is affine.

$$\begin{align*}
    \begin{bmatrix} A & B_1 & B_2 \\ C_1(\theta) & D_{11} & D_{12} \end{bmatrix} \in P :=
    \begin{bmatrix} A & B_1 & B_2 \\ C_1(\theta) & D_{11} & D_{12} \end{bmatrix} & \in P := \\
    Co & \{ \begin{bmatrix} A & B_1 & B_2 \\ C_1(\theta) & D_{11} & D_{12} \end{bmatrix} , i = 1, 2 \}
\end{align*}$$

(21)

**C. Control Design**

The convex gain-scheduling controller is computed at each vertex of the parameter polytopes, once $R$, $S$ and $\gamma$ are obtained by solving LMIs [19]. The gain-scheduling controller $K(\theta)$ shown in (22) is also a convex combination of vertex controllers $K_i$. In (23), $N_R$ and $N_S$ denote the bases of the null space of $(B_1^2, D_{11}^2)$ and $(C_2, 0)$, respectively. Weighting functions in Table I are used to design the controller.

$$\begin{align*}
    K(\theta) : \begin{bmatrix} A_K(\theta) & B_K(\theta) \\ C_K(\theta) & D_K(\theta) \end{bmatrix} &= \begin{bmatrix} A_{Ki} & B_{Ki} \\ C_{Ki} & D_{Ki} \end{bmatrix} \\
    Co & \{ K_i := \begin{bmatrix} A_{Ki} & B_{Ki} \\ C_{Ki} & D_{Ki} \end{bmatrix} , i = 1, 2 \}
\end{align*}$$

(22)
IV. RESULTS

The device parameters for a simulation study are adopted from the authors’ earlier work [10]. The parameters of the master device are $m_m = 0.90$ kg and $b_m = 2.99$ Ns/m, while those are $m_s = 0.30$ kg and $b_s = 1.37$ Ns/m for the slave manipulator. Parameters of human operator are defined as $m_{op} = 10$ Ns/m and $k_{op} = 200$ N/m. The stiffness of the surgical suture, $k_{env}$ is assumed to be 300 N/m. $f_{lim}$ is set to 3 N so that $x_{lim}$ is 0.01 m. Then, $f_{disc}$ is designed to be greater than $(1 + JND)k_{env}$. In the simulation, $f_{disc}$ is set to 366 N/m with the stiffness JND of 22%. To observe a big difference between $k_{env}$ and $k_{disc}$, $k_{disc}$ is set to 2000 N/m. Chirp signal is used for the input force $f_{op}$. Because surgeons carry out surgical procedures with very slow and careful movements, the frequency of $f_{op}$ is varying from 0.01 Hz to 0.5 Hz [20]. The amplitude of the signal is 50 N, which is greater than the average force of surgical procedures, to observe the change of the perceived impedance.

The gradient of position-force relation usually denotes the stiffness at various positions. As shown in Fig. 5, root mean square of the position tracking error is lower than 0.4 mm. The force tracking error, however, is not negligible as shown in Fig. 6 because the controller is to minimize the error between transmitted impedance to the operator and the specially designed impedance like Fig. 2.

If the results are drawn in position-force graph, impedance changes are easily shown as Fig. 7. However, it is observed that there is a sector of the stiffness change. The reason of this sector is due to high stiffness of $k_{disc}$. If $k_{disc}$ is decreased than 2000 N/m, the stiffness change is smoother than the one in Fig. 7. In Fig. 8, the frequency of $f_{op}$ is restricted to below 0.1 Hz. The stiffness is changed from 220 N/m to 1730 N/m.

\[
\begin{pmatrix}
(N_R^0)^T \\
0 \\
I
\end{pmatrix}^T
\begin{pmatrix}
AR + RA^T & RC_i^T \\
B_1 & C_{1i}
\end{pmatrix}
\begin{pmatrix}
N_R^0 \\
0 \\
I
\end{pmatrix} < 0, i = 1, 2,
\]

\[
\begin{pmatrix}
(N_R^0)^T \\
0 \\
I
\end{pmatrix}^T
\begin{pmatrix}
A^T + AS \\
B_1^T
\end{pmatrix}
\begin{pmatrix}
C_{1i} & D_{1i}^T \\
D_{1i} & -I
\end{pmatrix}
\begin{pmatrix}
N_R^0 \\
0 \\
I
\end{pmatrix} < 0, i = 1, 2,
\]

\[
\begin{pmatrix}
N \\
I \\
I S
\end{pmatrix} \geq 0
\]

(23)
V. DISCUSSIONS

Previous research efforts regarding to haptic feedback in teleoperation systems are mostly focused on to achieve transparency or fidelity. This paper, on the other hand, is developing a haptic feedback method to solve a specific as well as a practical problem in telesurgery. The developed control method shows a possibility to enable the surgeon to detect a specific force more accurately and sensitively.

The major difference between the proposed method and the sensory substitutions of [3], [4] is the physical interactions with the operator. The developed haptic feedback method is not only giving a warning sign to the operator but also physically preventing the excessive force of the slave side. Along with this perspective, the study about the operator awareness and the safety in telesurgical systems will be studied in near future.

The developed control has to be validated more rigorously and systematically with an experiment with real environments (e.g., in-vitro experiment). In addition, $k_{\text{env}}$ and $k_{\text{disc}}$ have to be designed with a priori experimental data of several surgical sutures. In this paper, $k_{\text{disc}}$ is set to seven times greater than $k_{\text{env}}$. The change is surely detected by the operator, but the high stiffness may disturb the operation as well. Therefore, guidelines of $k_{\text{disc}}$ have to be developed based on psychophysical evaluation. Psychophysical evaluation study with subjects including surgeons is also one of future research directions. Comparison study with previous methods also could be conducted via the evaluation study.

Although the suture is modeled as a spring in the formulation, the unilateral constraint is possible to consider. The transition between free and constrained motion can be solved with same framework. The environment impedance $Z_{\text{env}}$ is regarded as a parameter-varying model from zero to $k_{\text{env}}$. The designed transmitted impedance $Z_d$ has to be covered wide range of zero to $k_{\text{disc}}$. In this case, it contain two parameters in the state-space model. As we discussed in [10], backdrivability is difficult to obtain in gain-scheduling control scheme. This problem could be addressed with a design of the parameter polytope or a design of new control objectives. The remaining open problems and experimental evaluations are subjects of future work.

VI. CONCLUSIONS

In this paper, we proposed a stabilizing haptic feedback control to detect a specific force. The lack of haptic feedback during surgical procedures may incur suture breakages or knot failures in case of novice surgeons. Because direct haptic feedback is not an adequate way to detect the suture breakage force as pointed out by previous researchers, a novel haptic feedback control method is developed with the perspective of human perception.

To detect the specific force, transmitted impedance to the human operator has to be designed with respect to human perceptual sensitivity. Because the designed impedance is varied according to its interaction position, the overall system is handled via the polytopic system approach. Then, the gain-scheduling control scheme is adapted to develop an optimized controller while guaranteeing robust stability. By employing frequency-dependent weighting functions and control objectives, a sub-optimal controller is obtained for telesurgical systems to enable the human operator detect the suture breakage force. The proposed methodology is also validated with numerical simulations.

REFERENCES