31st Annual International Conference of the IEEE EMBS
Minneapolis, Minnesota, USA, September 2-6, 2009

Design and Implementation of Series Elastic Actuators for a Haptic Laparoscopic Device

E. Basafa, M. Sheikholeslami, A. Mirbagheri, F. Farahmand, G. R. Vossoughi

Abstract— **The design of a laparoscopic haptic device based on a 4-DOFs mechanism and Series Elastic Actuators (SEA) is described and the results of the theoretical and experimental examinations are presented. With a sufficient bandwidth and low impedance, the system provided a stable interaction with soft tissues, e.g., human liver, in virtual environments.**

I. Introduction

VIRTUAL environments have provided new means for the surgery trainees to practice modern complicated surgical procedures, especially minimally invasive surgeries (MIS) [1]. In some of surgical simulators the commercially available haptic devices, e.g. PHANToM® (Sensable Technologies, Woburn, MA) have been utilized [2]. These devices, however often suffer from redundant or insufficient degrees of freedom (DOFs), incompatible workspace geometries or inappropriate force producing capabilities. Hence, design of a specialized force feedback device for laparoscopic surgery training has been of much interest among researchers [3].

The most fundamental consideration in the design of a haptic device is the number and arrangement of the DOFs. A laparoscopic instrument has four DOFs once it enters the patient body through an incision point, one translational and three rotational. Several mechanisms have been proposed in the literature to provide these DOFs, e.g., spherical and concentric multilink spherical [4,5], parallel [6], and cable based [7]. These mechanisms, however, should also satisfy the other mechanical requirements of a desirable haptic device, such as high structural stiffness and low friction, inertia and backlash. The other basic consideration in the design of haptic devices is how to actuate the DOFs of the mechanism, i.e., the actuators selection and implementation method. An appropriate actuator shall meet the specific requirements for the application of interest, particularly the force/torque capacity, the closed loop force/torque bandwidth and the output impedance. Moreover, it is to be implemented properly to the mechanism, e.g., by mounting to the frame, so that the structural stiffness remained high.

In many conventional haptic devices the actuator, i.e., motor, is connected to the joint directly or via a stiff force/torque sensor. This study describes the application of series elastic actuators (SEAs) to a haptic device for laparoscopic surgery simulation. It was expected to provide a better closed loop behavior and reduced output impedance, considering the compliant elastic properties of SEAs [8]. The latter is important in simulation of haptic interactions, since the user should feel small or no forces when moving the device freely in the virtual space without touching an object.

E. Basafa received his master in Biomechanical Engineering from Sharif University of Technology, Tehran, Iran and is now a Ph.D. student of Mechanical Engineering at the Johns Hopkins University, Baltimore, MD 21218, USA (e-mail: basafa@jhu.edu).

M. Sheikholeslami received his B.Sc in Mechanical Engineering from Sharif University of Technology, Tehran, Iran and now is a M.Sc student of Mechanical Engineering at McGill University, Montreal, QC, Canada (email: majid@cim.mcgill.ca).

A. Mirbagheri is a Ph.D. Candidate of Mechanical Engineering at Sharif University of Technology and a research assistant at Research Center for Science and Technology In Medicine (RCSTIM), Tehran, Iran (e-mail: mirbaghery@mech.sharif.edu).

F. Farahmand is a professor and chairman of School of Mechanical Engineering at Sharif University of Technology, Tehran, Iran (e-mail: farahmand@sharif.edu).

G. R. Vossoughi is an associate professor of School of Mechanical Engineering at Sharif University of Technology, Tehran, Iran (e-mail: vossough@sharif.edu).

II. Method

A parallel mechanism proposed by Rosenberg [9] was selected for the mechanical compartment of the haptic device. This mechanism provides four DOFs: three rotational degrees (pitch, yaw, and roll) around a fixed (incision) point and one axial translation. A major advantage of the mechanism over others in the literature [4-7] is the reduced number of links causing a relatively higher stiffness and larger workspace. Moreover, the mechanical configuration of the mechanism allows the actuators for two rotational DOFs to be mounted on the frame. This helps to keep the inertia low providing a better back drivability which is important feature for haptic interactions [10]. The linear DOF of the device was implemented using the Watt mechanism [11]. This four-bar linkage provides a straight path for the midpoint of its coupler within a range of its motion. Hence a linear movement could be obtained with no force acting perpendicular to the path of the motion, maintaining the friction low.

For the current design, three active DOFs were considered for the device; the axial rotation DOF of the handle was assumed to be passive. The design characteristics of the actuators were considered in detail. The maximum forces applied to the surgical instrument during a general laparoscopic surgery have been reported to be of the order of 1-5 N [12]. So, assuming a 30 cm handle length for the instrument, a 1.5 N-m maximum torque is sufficient for the

978-1-4244-3296-7/09/$25.00 ©2009 IEEE

actuators. For the force bandwidth, we chose a target of 15 Hz. This approximates the small bandwidth of the motion of the human proximal limbs (arms) [13], and is about ten times the bandwidth of a surgeon's hand movements during laparoscopic surgery.

The schematic of a series elastic actuator designed based on the above considerations is shown in Fig. 1 [14]. Let's suppose a pure velocity source is used as the motor and controller of the system. The transfer function of the system in the absence of the output motion can be written as:

$$G_{SEA}(s) = \frac{f_{load}}{f_{desired}} = \frac{K_s}{\tau s + K_s} \tag{1}$$

where K_s is the spring (elastic element) stiffness and τ is the plant time constant. We can also write the output impedance transfer function in the form of the following equation:

$$imp(s) = \frac{f_{load}}{X_{load}} = \frac{-K_s \tau s}{K_s + \tau s} \tag{2}$$

It is obvious from Equ. (1) that increasing the spring stiffness increases the effective force bandwidth of SEAs. On the other hand, Equ. (2) suggests that the spring stiffness is to be minimized in order to provide a lower impedance. Therefore a trade-off should be made between these two factors. We selected a fairly compliant spring for SEAs, considering the fact that the required bandwidth for a laparoscopic haptic device is relatively low. A small time constant was also considered for the SEAs due to the fact that it yields better results for both the force bandwidth and the output impedance [15].

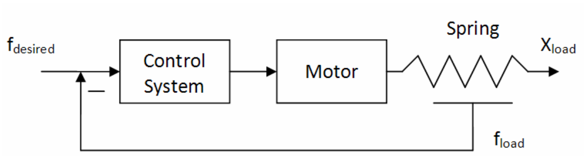

Fig. 1. The schematic of a series elastic actuator

Fig. 2 shows one of the SEA units developed for actuation of the three DOFs of the laparoscopic haptic device. The HS-5955TG servo motor (Hitec RCD, Poway, CA) was used as the torque source of the SEAs. Before implementing into the actuator, the step response of the servo was recorded in order to determine the system's time constant. The original internal potentiometer of the servo was replaced with a potentiometer mounted between the servo shaft and the actuator output shaft and wired to the inside control board of the servo to provide the position *difference* error. The modified servo was mounted on a frame and a second potentiometer was placed parallel to its output shaft to provide the absolute actuator angle for the controller. The actuator motion was conveyed to the potentiometer via a small pulley and belt arrangement.

The elastic element of the SEAs was implemented using

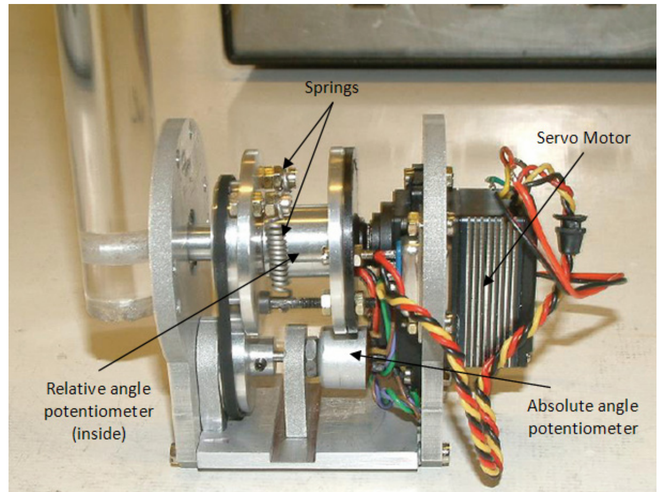

Fig. 2. A series elastic actuator unit developed for the laparoscopic haptic device of the present study.

two tension springs mounted between the output shaft of the servo and the output shaft of the actuator and connected via a bearing. This design resulted in a linear torque-deflection angle relationship for the springs with a constant stiffness of 8.75 Nm/rad, as long as the relative angular motion of the two shafts was limited to ±12 degrees.

An electronic PCB including two blocks was developed to connect the actuator to a PC: the first block was a 10-bit, 8-channel A/D converter used to convey the position of the absolute (output) potentiometer to the PC. The second block was a 6-channel PWM signal generator which commanded the desired position to the servo controller, according to the command sent from the PC. Both blocks were designed and programmed on an ATMEGA128L (ATMEL Corp., San Jose, CA) microcontroller. The RS232 port of the PC with a baud-rate of 115200 bits/s was used to communicate with the PCB. A specific thread in a C++ program was used to conduct this communication with the highest possible priority and speed. The servo position command and the potentiometer angle were updated with frequencies of 500Hz and 800Hz, respectively. Fig. 3 shows the overall system architecture for each servo control.

The performance of the SEA units developed was evaluated in a number of experimental tests. The torque capacity of the actuators was determined by attaching known weights to a handle rigidly connected to the actuator and commanding the actuator to hold them horizontally. To determine the force bandwidth of the actuator, its output shaft was first fixed to the frame using a screw. Then the actuator was commanded to apply sinusoidal varying torques with magnitudes of %25 and %40 of the maximum applicable torque and frequencies between 2 and 120 rad/s (0.32 to 20 Hz). For each torque and frequency, the spring deflection (output torque) was recorded three times, for a 5 sec interval each, and averaged.

The output impedance frequency response of the actuators was determined by commanding a zero force to the actuator

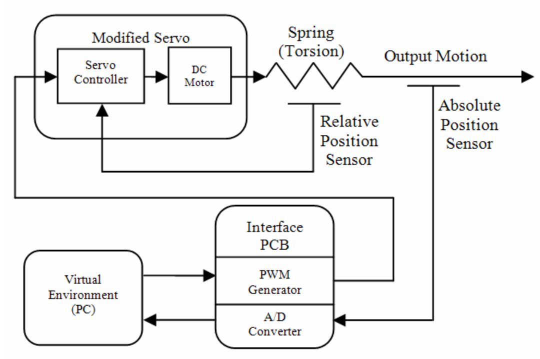

Fig. 3 Schematic of the actuator design and control

and moving the handle attached to the output shaft with different speeds. The output motion and forces were then recorded and a Fast Fourier analysis was used to determine their dominating frequencies and magnitudes at different speeds. Finally the actuator was tested while interacting with a spring and a damper, simulating a simple virtual environment, to obtain the stiffness range with which it could interact. In this experiment the motion of the handle attached to the actuator was assumed to be linear in the short arc of its movement.

After testing the actuators, they were implemented into the mechanical compartment of the laparoscopic haptic device fabricated in our lab. All the mechanical parts for the gimbal and linear mechanism were made of light weight aluminum alloy and the kinematics and dynamics equations of the device were derived based on its geometrical configuration and the weights of its components. A handle-shaped part, equipped with a potentiometer, was added to the device to complete the design and provide the angular feedback from the twist angle. In order to eliminate the weight of the linear mechanism which moved along with the handle, simple geometrical relationships were employed to find the compensating torques at the two DOFs of the gimbal mechanism. Finally, the device was connected to the PC using the PCB described above and its performance was examined while probing a virtual liver model developed at our lab [16].

III. Results

The best mathematical function fitted to the step response of the servo motor of the actuators had a time constant of 0.12 sec. This data along with the spring constant was used to obtain the theoretical frequency response of the actuator using Equ. (1). The theoretical and experimental results for the force bandwidth frequency response of the actuators are shown in Fig. 4 indicating a force bandwidth of about 12 Hz. The minimum and maximum controllable torque magnitudes were obtained to be 0.06 and 1.38 Nm, respectively, indicating a dynamic range of 23:1. The theoretical and experimental results for the output impedance frequency response of the actuator are illustrated in Fig. 5. The

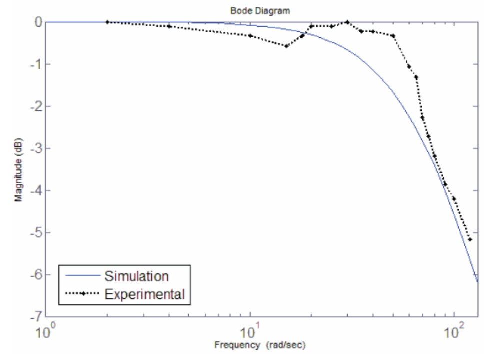

Fig. 4. Force bandwidth frequency response

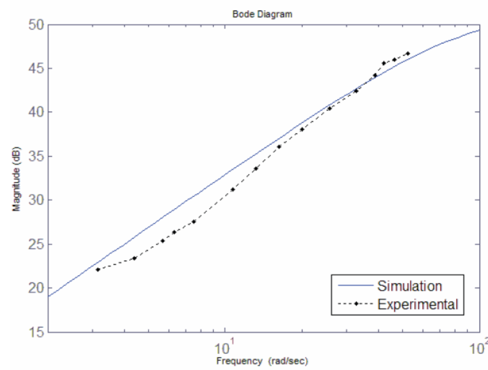

Fig. 5. Output impedance frequency response

maximum undamped spring stiffness with which a stable interaction could be performed was obtained to be 350 N/m. However, higher stiffnesses up to 400 N/m could be touched if some damping (up to 1 Ns/m) was considered for the springs.

Fig. 6 shows the actuators implemented into the mechanical compartment of the laparoscopic haptic device and being used in a virtual environment. The two DOFs of the gimbal had a range of motion of ±75º and the linear DOF could travel a distance of >10 cm being 30 cm away from the pivot point when fully retracted. A box of $50x50x10\ cm^3$ can roughly approximate this workspace, which is larger than the workspace dimensions reported for most commercially available and widely used haptic devices, e.g. Phantom® Omni and Premium 1.0 and CyberForce® (Immersion Corp., San Jose, CA). The passive twist angle had also a range of rotation of >270º. The linear DOF could produce forces of up to 10 N. The two gimbal actuators had output torques to generate forces of up to 5 N in the other two directions at the handle when the handle was fully retracted. This magnitude increased when the handle moved inside the gimbal along the linear DOF, however, it was independent of the two gimbal angles, thanks to the parallel mechanism. The forces were also equal or larger than those reported for previously mentioned devices.

Preliminary tests on the haptic device while probing a virtual liver model (Fig. 6) showed that the haptic device could effectively convey forces to the users during a simulated surgical procedure.

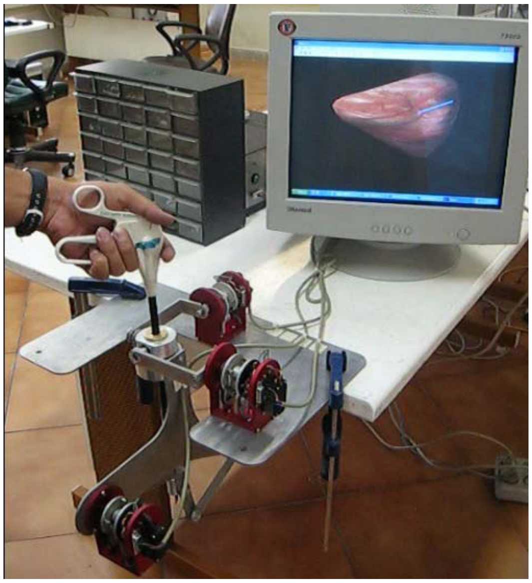

Fig. 6. The haptic device being used in a virtual environment to probe a liver model

IV. CONCLUSIONS & FUTURE WORK

A new laparoscopic haptic device was introduced based on a parallel mechanism and series elastic actuators. The device provides the required DOFs and has a large workspace sufficient for using in various laparoscopic surgery simulations. The SEAs provide lower output impedances in comparison with the conventional force/torque actuators. This feature, along with the gravity compensation makes the device virtually "transparent" to the users. Although the current design has a relatively low force bandwidth, it is adequate for probing soft tissues in a virtual environment of laparoscopic surgery, which does not often need a high bandwidth and update rate. Furthermore the low cost of manufacturing the device makes it a good candidate for wide use in medical education applications. The results of the experimental tests showed that the designed actuators provide a high dynamic range and sufficient torque capacity for interacting with soft materials. The stable interaction obtained for materials with up to 400 N/m stiffness, suggest that the haptic device can effectively be used for simulating usual laparoscopic surgeries on the abdominal organs and tissues. In particular, the preliminary tests conducted on a virtual liver model developed at our lab [16] indicated an acceptable performance for the system. However, for interactions with highly stiff materials, e.g. cartilages and bones as happens during arthroscopy, implementation of additional damping characteristics into the elastic element in needed which is often difficult due to the increasing noise of the system.

Finally, it should be noticed that the current design utilizes the control board inside the motors for the purpose of torque control. Although this solution results in a simpler control scheme, the relatively slow update rate of the position makes simulating stiff objects more difficult. Another way of having control over the output torque is to use the DC motor current as the input command to the actuator, which might be performed with higher update rates. Also replacing the potentiometers with encoders can reduce the noise problem in calculating the angular velocity and help to model stiffer objects. Work is under way for developing new virtual scenarios for laparoscopic surgery, including simulation of cuts and sutures. These environments can be used along with the haptic device to provide a complete simulator for more efficient laparoscopic training.

REFERENCES

[1] U. Kuhnapfel, H. K. Cakmak and H. Maab, "Endoscopic surgery training using virtual reality and deformable tissue simulation", *Computers and Graphics*, vol. 24, pp. 671-682, 2000

[2] M. Tavakoli, R. Patel and M. Moallem, "A haptic interface for computer-integrated endoscopic surgery and training", *Virtual Reality*, vol. 9, no. 2, pp. 160-176, 2006

[3] R. Baumann and R. Clavel, "Haptic interface for virtual reality based minimally invasive surgery simulation", in *Proc. IEEE Intl. Conf. Robo*tics and Automation, Leuven, Belgium, 1998

[4] A. Faraz and S. Payandeh, "A robotic case study: optimal design for laparoscopic positioning stands", in IEEE International Conference on Robotics and Automation, 1994

[5] G. J. Hamlin and A. C. Sanderson, "A novel concentric multilink spherical joint with parallel robotics applications", in IEEE International Conference on Robotics and Automation, 1994

[6] R. Baumann, W. Maeder, D. Glauser and R. Clavel, "The PantoScope: a spherical remote-center-of-motion parallel manipulator for force reflection", in IEEE International Conference on Robotics and Automation, 1997

[7] L. B. Rosenberg, "Interface apparatus with cable-driven force feedback and grounded actuators", June 6, US Patent No. 7056123, 2006

[8] J. E. Pratt, B. Krupp and C. Morse, "Series elastic actuators for high fidelity force control", Industrial Robot: An International Journal, vol. 29, pp 234-241, 2002

[9] L. B. Rosenberg, "Laparoscopic simulation interface", U.S. Patent 7023423, April 4, 2006

[10] P. A. Millman, M. Stanley and J. E. Colgate, "Design of a high performance haptic interface to virtual environments", in *Proc. IEEE Virtual Reality Annual International Symposium*, pp. 216-222, 1993

[11] G. H. Martin, *Kinematics and dynamics of machines*, New York, NY: McGraw-Hill, 1982

[12] G. Tholey, A. Pillarisetti and J. P. Desai, "On-site three dimensional force sensing capability in a laparoscopic grasper", Industrial Robot: An International Journal, vol. 31, no. 6, pp. 509-518, 2004

[13] R. E. Ellis, O. M. Ismaeil and M. G. Lipsett, "Design and evaluation of a high performance haptic interface", Robotica, vol. 14, pp. 321-327, 1996

[14] E. Basafa, H. Salarieh and A. Alasty, "Modeling and control of nonlinear Series Elastic Actuator", in Proc. ASME/IEEE Intl. Design Engineering Technical Conf. & Computers & Information in Engineering Conf., September 4-7, Las Vegas, Nevada, USA, 2007

[15] D. W. Robinson, "Design and analysis of series elasticity in closed-loop actuator force control", PhD dissertation, Dept. Mech. Eng., Massachusetts Institute of Technology, Boston, MA, 2000

[16] E. Basafa, F. Farahmand and G. Vossoughi "A non-linear mass-spring model for more realistic and efficient simulation of soft tissues surgery", in *Proc. 16Th Annual Medicine Meets Virtual Reality Conference (MMVR16)*, Jan. 29 – Feb. 1, Long Beach, CA, 2008