Haptic Training Simulator for Pedicle Screw Insertion in Scoliosis Surgery

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Abstract. Pedicle screw insertion is a common treatment in fixing spinal deformities in idiopathic scoliosis. This paper discusses the development of a haptic device which will aid in training for pedicle screw insertion surgery. By translating the rotational and linear effects of the surgery into tunable haptic parameters, a realistic haptic simulator is created. Over 10 surgeons of varying experience levels used the simulator, and were able to tune the device to what they felt was most realistic. They were also asked to judge the system based on its feasibility and usefulness. The results indicate the simulator is feasible for surgical education.

Keywords: Haptics · Scoliosis · Pedicle screw insertion

1 Introduction

Idiopathic scoliosis is defined as a lateral curvature of the spine greater than 10 degrees with unknown cause. The reasons for treatment include improving physical appearance, reducing back pain, promoting physical comfort, and preventing excessive spine curvature. In pedicle screw instrumentation, screws are placed through the pedicle and inside the vertebral body, and are then connected by a short rod which straightens the spine [1]. This type of surgery is often conducted using the free-hand anatomic technique and relies on visual and haptic feedback. Steps in pedicle screw insertion consist of identifying the entry point, removing the cortical cortex of the pedicle, creating the channel, palpation and placing the screws.

In the critical phase of channel creation, the probe is pushed through the pedicle and towards the vertebral body [2]. The required depth is different for different regions of the spine. The advancement of the probe should be smooth and consistent. A sudden change in resistance means that the probe is touching

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the pedicle wall or the wall of the vertebrae body. In such situations, a sudden downward motion (breach) can occur if the surgeon continues to apply force to the probe [3]. Many visual aids are difficult due to deformity of the bone, in other words haptics is key.

Creating a pathway through the pedicle by the free-hand technique is composed of two main degrees of freedom: rotation and linear (translational). Rotating the probe removes the soft cancellous bone and applying force creates linear translational movement along the pedicle axis. Optimal screw insertion relies on the experience of the surgeon and his ability to differentiate the tactile sensations associated with different textures in the bone when performing channel creation [4]. This is because the surgeon has limited visibility of the internal organs or spinal cord. Complications can occur due to an incorrect entry point, an incorrect trajectory, or failure to recognize wall breaches. The complications cause neurological issues, visceral organ's damage and/or problems with mechanical motion [1,5,6].

Although the accuracy of surgeons is not solely due to the surgeons' experience or lack of experience, studies showed that experienced surgeons have a significantly lower chance of having a medical breach than novices [7]. With the steep learning curve in the procedure, this haptic simulator can allow surgeons and residents to learn and practice the differentiation between proper and improper haptic signals. Among the current approaches in the education of surgeons, haptics simulators provide the trainee with the safest and most repeatable environment. Traditional surgical training includes supervised practice on live patients or cadavers. The former risks the comfort and safety of the patient, and also extends the time and cost of the operation in order to allow for corrections to be made. The latter is expensive and also imposes unrealistic physiological responses due to the embalming chemicals and lack of blood pressure in cadavers. Moreover, it is very difficult to assess the practitioners' proficiency through these techniques. Virtual reality simulators provide the trainee with unlimited practice with no time constraints and, by integrating sensors to the simulator, makes it feasible to assess skills [8]. Current simulators usually employ visual and/or touch modalities to replicate the real environment. A problem with many of the available haptic platforms is that they fail to create realistic effects due to device limitations. For instance, the haptic feedback related to a spine biopsy simulator remained limited to interactions with soft tissues since the haptic device being used was unable to provide high realistic force peaks [9,10]. Most available works focus on the visual aspects of the surgical education [11–13]. As well, most virtual reality simulators for bone tissue procedures involve power drill simulations and not the free hand technique [14,15]. This paper presents a customized simulator that is the first to be able to emulate the high forces created in the free-hand pedicle screw insertion technique. This work adds a linear degree of freedom to the rotary stage of previous work [16] and can simulate the haptic effects associated with the coupled two degrees of freedom involved in the pedicle screw insertion: rotation and linear progression. Haptic model parameters for a spine surgery with normal bone density are clinically tuned within this user study.

Although the proposed haptic techniques in this study are capable of simulating various anatomical scenarios, for the initial prototype, virtual clinical testing is only performed for healthy vertebrae with normal size and normal bone density as opposed to an older, osteoporotic bone or a high-density bone in a young patient.

2 Biomechanical Characteristics of the Surgical Procedure

The pedicle is composed of two types of tissue: cancellous bone, which is soft bone with a low density structure, and cortical bone, the harder, outer layer of the pedicle wall. The pedicle screw insertion procedure can be split into two degrees of freedom: rotational and linear progression. Probe rotation through cancellous bone causes vibrations, as well the sensation of going a series of bumps, similar to the scratching of a match across a surface. Probe rotation through the cortical bone, on the other hand, creates the sensation of viscous friction. Linear progression through the cancellous bone creates the sensation of moving over small bumps in a smooth and consistent manner. If linear progression is not performed correctly, perforation of the pedicle wall, or a breach, can occur. Breaches occur due to either the incorrect probe trajectory, or when the surgeon continually applies pressure when feeling high resistance on the probe. Images of correct and incorrect (breached) pedical screw insertion are shown in Fig. 1. Based on feedback from expert surgeons, the linear degree of freedom is composed of two system effects: linear motion and breach effects. These effects are dependant on the the anatomical characteristics of each patient, and as a result, the sensations are not consistent across all patients. The prototype for this study focuses solely on vertebrae with normal size and normal bone density, as would be found in healthy patients. Additionally, as will be explained in the further sections, breach effects will not be a focus of this study.

3 Experimental Platform

The rotary stage of the system was developed in an earlier iteration of the platform [16]. The mechanical component of this stage consists of a fabricated probe handle connected to the shaft of a non-geared DC motor. Connected to the motor is an encoder which measures the angular position of the probe. The hardware for the system interfaces with a PC using a data aquisition system (DAQ), and the haptic effects are programmed in MATLAB/Simulink and run at a sampling rate of 1000 Hz.

For the linear stage of the simulator, an electrical geared motor, with load capacity of 100 pounds, was chosen as the linear actuator for the system. This actuator is capable of supplying forces of over 100N. Delivering high forces is important for this application because surgeons typically exert extreme forces during the procedure. To measure the position along the axis of the linear actuator, a position sensor is used and is coupled to the rotary stage plate. Lastly,

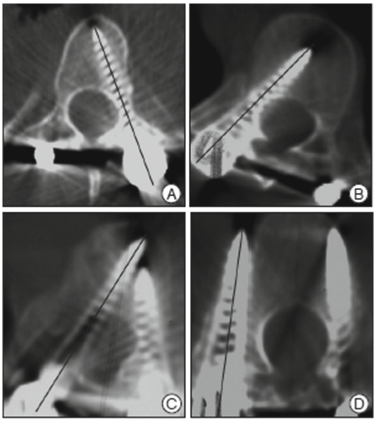

Fig. 1. Computed tomography (CT) scan of screws inside pedicle. (A) and (B) demonstrate a proper pedicle screw insertion. (C) shows the medial cortical breach and (D) shows a lateral cortical breach [2]

a tranducer is placed between the probe and the shaft of the motor to measure the user's force and torque. The entire experimental set up is shown in Fig. 2.

4 Control Structure and Design

The general control structure that is used in this work is shown in Fig. 3. It includes the rotary stage control scheme coupled with the linear stage control scheme. The rotary simulation model that was proposed in [16] consists of two main haptic effects: vibration effects and viscous friction effects. Vibration effects are modelled as a series of bumps through rotation and is simulated by using a derivative controller. The effects are felt as a resistive torque which increases in magnitude proportionally to the speed of rotation. In the rotary stage of the block diagram, shown in Fig. 3, the angle θ_d is the desired angular position and is generated using the trajectory planner block. This block uses Eq. 1 which is a function of the detent interval and the detent width (indicated by θ_i and θ_w, respectively).

$$\theta_d = \begin{cases} \theta', & \text{if } \theta' < \theta_L \\ \theta_L, & \text{if } \theta_L < \theta' < \dfrac{\theta_i}{2} \\ \theta_R, & \text{if } \dfrac{\theta_i}{2} < \theta' < \theta_R \\ \theta', & \text{if } \theta' > \theta_R \end{cases} \tag{1}$$

here, θ' is the remainder obtained if θ is divided by θ_i. The response torque is calculated using the PD controller in Eq. 2.

$$\tau_c(t) = k_p(\theta_d - \theta') - k_d\dot{\theta} \tag{2}$$

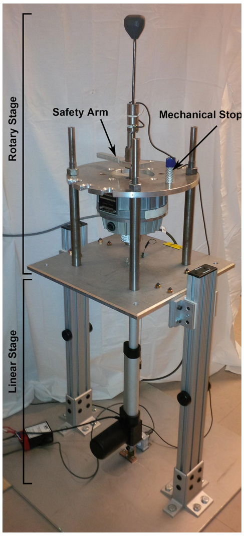

Fig. 2. Haptic simulator experimental setup

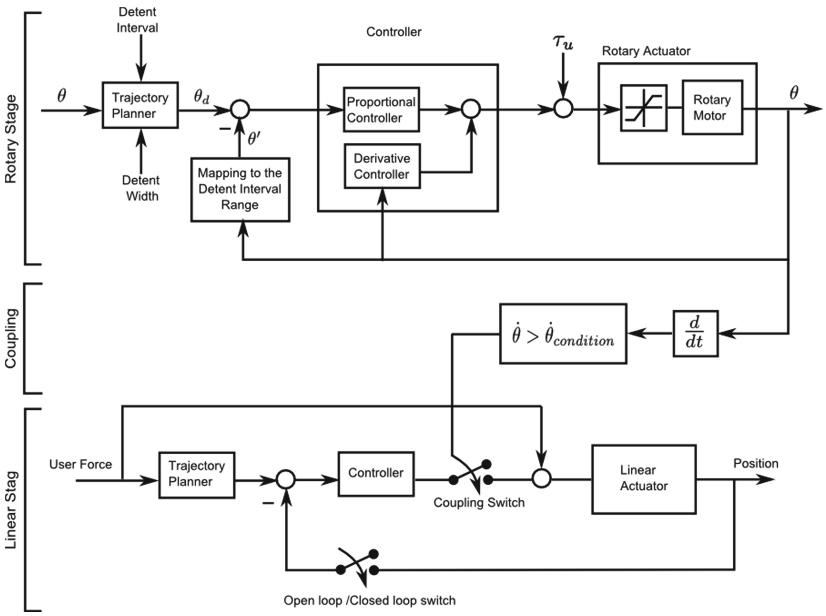

Fig. 3. General control diagram of the haptic training simulator

The proportional component of the controller creates a series of detents in the probe position. When the position of the probe lies in the first and last case in Eq. 1 , the response torque is zero. As the probe position moves away from θ_L, the controller creates an increasing torque similar to the feeling of a spring compressing. As the the probe passes the midpoint and gets closer to θ_R, the sensation is similar to that of a spring returning to its starting position.

All four haptic parameters for the rotational stage (detent interval, detent width, detent magnitude, viscous friction coefficient) are tuned to what surgeons feel is most realistic to the actual surgical scenario during clinical testing.

The second stage of the control scheme, shown as the linear stage in Fig. 3, simulates the linear dynamics of the system. The device used for simulating these effects should be able to replicate the vibration and resistance sensation felt as the probe proceeds through the pedicle. For simulating breach effects, the actuator must be able to deliver very large forces. Additionally, the device must use impedance control. For simulating the haptic effects of the linear stage, two control strategies are presented. One strategy employs a closed loop PID control technique, while the other strategy uses an open loop control scheme. For simulating linear progression, both control techniques simulate vibration by making step-wise motions whose progression speed is the control variable. Breach simulations involve simulating the dropping of the probe with a certain displacement in a short amount of time.

Through observations and discussions with expert surgeons it was found that rotation was a key factor in linear progression, i.e. the rotational and linear effects are coupled, shown in Fig. 3 as the coupling stage. To simulate this, the angular velocity of the probe rotation is used for determining how subtle rotation should be for linear progression. This is verified through a comparison block which controls the coupling switch between the linear and rotational stages.

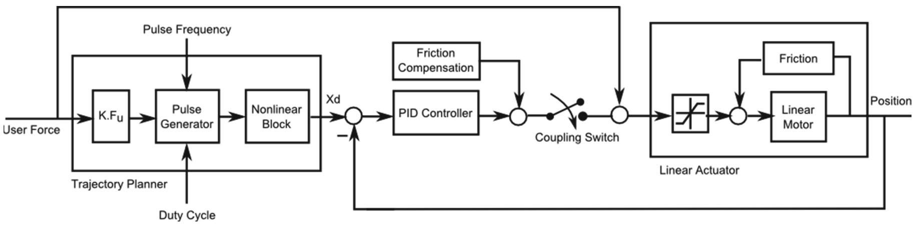

Fig. 4. Block diagram of the PID control scheme

4.1 Closed Loop PID Control

Although the closed loop control technique was not implemented for the final prototype, the following is a brief description of its design. For the control of the linear stage of the haptic device, a trajectory planner is used to generate the desired reference trajectory and a PID controller is used for driving the motor to follow the trajectory. A block diagram for this approach is shown in Fig. 4.

For linear progression, the trajectory planner uses three inputs to generate the desired trajectory. First, the user force is scaled by a factor of k, which determines how much force the user should put on the probe to get motion, and is then converted into a series of pulses. The pulse frequency and duty cycle are tuning parameters that can change the vibration sensation. The pulses are then passed through a nonlinear block where the current signal is added with its one-sample-delayed signal. This block updates the reference trajectory.

To simulate breach effects, the desired displacement, the user's force and a force threshold are used as inputs. When the user's force is greater than the force threshold, a step signal, with the size of the desired displacement, is injected into the nonlinear block mentioned earlier. This check remains idle for a short period of time to avoid multiple true conditions in the presence of noisy signals.

4.2 Open Loop Control

For open loop control, the displacement is no longer the main focus, instead, the speed of movement is controlled. The block diagram for this control scheme is shown in Fig. 5.

For simulating linear progression, vibrations can be achieved by simply feeding the actuator with a series of pulses. The user force is first measured and scaled by a factor of k to specify how resistive the motion will be. The frequency and duty cycle of the pulses determine the sensation the user feels. If there is friction inside the motor, part of the control signal is used to overcome the friction and move the actuator as intended.

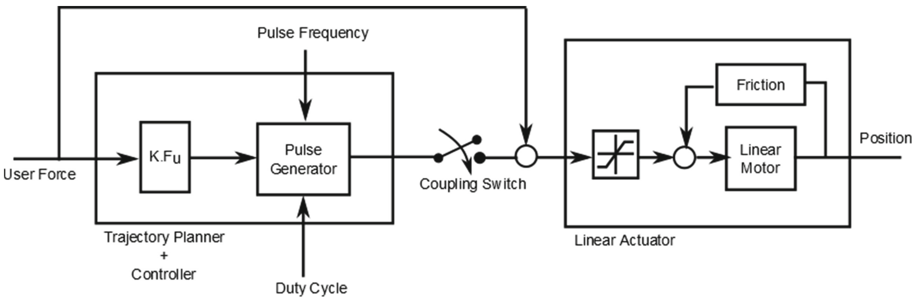

Fig. 5. Block diagram of the open loop control scheme

For breach simulation, the force condition is checked, just as in the closed loop control scheme. However, instead of feeding the signal into the Zero-Order-Hold block, a pulse of maximum input and specified duration is fed directly into the actuator. When the actuator is fed with the maximum input, the displacement of the actuator is controlled by the duration of the pulse. This relationship between the pulse width and actuator displacement can be approximated by a linear function.

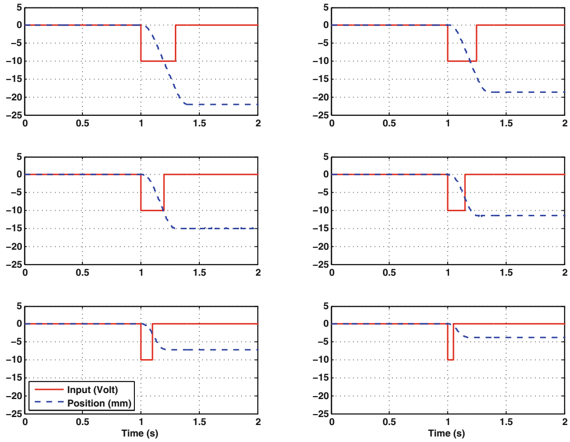

Fig. 6. Displacement of the actuator with maximum input voltage and six different pulse duration including 0.3, 0.25, 0.2, 0.15, 0.1 and 0.05 s.

Open Loop Control Design. In open loop control, as mentioned previously, it was found that there is a linear relationship between the input pulse width and the displacement of the actuator. Using a series of tests where displacements were measured from differing pulse widths, it was found that the ratio between pulse duration and displacement is 1 : 7.44 with a standard deviation of 0.17. The results of each test signal is shown in Fig. 6.

5 Clinical Tuning

To further refine the simulation model to more accurately simulate the biomechanical properties of the scoliosis procedure, model parameters were tuned with the aid of participating surgeons. For this study, 11 surgeons were recruited to define the model parameters for normal density bone. 8 participants were orthopedic surgeons and 3 were neurosurgeons. Among the orthopedic surgeons, one was senior, three were fellows and four were residents. The set of neurosurgeons consisted of one surgeon from each experience level. Senior surgeons had over 15 years of experience, fellow surgeons had 6–10 years of experience and residents had between 4–9 years of experience.

For the study, each participant stands beside the device and holds the probe with one hand, while holding the top plate of the device with the other. Prior to testing, each participant goes through a training session where the haptic effects are introduced. Each model parameter is changed over a wide range of values, allowing the surgeons to experience each sensation independently. After training, each surgeon is asked to tune the parameters so that the simulation is equivalent to feeling normal, healthy bone. Once all parameters are adjusted, the surgeons are given another chance to feel their tuning and perform any final tuning. After tuning, the surgeons are told to perform the procedure of probe channelling on the simulator while the force, torque, linear position and angular position are recorded. At the end of each trial, the surgeons complete a four question, five-point Likert scale survey probing how well the effects of haptic simulator compared to the real surgery. The survey consisted of the following statements: (1) The haptic sensations associated with the rotation of the probe was simulated realistically; (2) The haptic sensations associated with the linear progression of the probe was simulated realistically; (3) Overall, the simulator produced realistic haptic sensations felt during probe channelling; (4) The simulator could potentially be a useful tool for teaching pedicle screw insertion surgery.

5.1 Results

Although it is not feasible to find exact parameter values that generate the most realistic haptic sensations on normal bone, there is sufficient evidence to conclude that at least 50 % of surgeons can perform within 25 % tolerance of the average of the two senior surgeons' tuned values for five parameters. These parameters are detent interval, viscous friction coefficient, duty cycle, frequency and scaling gain [17]. The exact statistics for each parameter, for the senior surgeons, are shown in Table 1.

The survey results show that the simulator provided a realistic haptic simulation of rotation and linear progression. They also agreed considerably that the simulator can potentially be a useful tool for teaching pedicle screw insertion surgery. For all four questions, the fellow and senior surgeons either agreed or strongly agreed.

Table 1. Percentage of surgeons able to adjust the simulator parameters to within 25 % of the average of the two expert surgeons values (N = 9) [17]

	Parameter	Seniors' Avg.	Completion rate	Exact prob
Rotary stage	Detent interval	1.85	55.6 %	62.30 %
	Detent width	0.4	33.3 %	17.19 %
	Detent magnitude	0.85	44.4 %	37.70 %
	Viscous friction coeff	15	66.7 %	82.81 %
Linear stage	Duty cycle	0.7	88.9 %	98.93 %
	Frequency	8.25	77.8 %	94.53 %
	Scaling Gain	0.16	66.7 %	82.81 %

6 Conclusion

There are relatively few surgeons in Canada performing pedicle screw insertion. Therefore, participant recruitment was a major challenge for the clinical tuning of parameters. Moreover, not all surgeons perform the surgery regularly and most residents develop their surgical skill set on cadaver bones rather than the healthy bones of young patients, who comprise the highest volume of scoliosis surgical cases. Despite the small sample size, among the participants is a senior surgeon who performs a major proportion of all this type of surgery in Canada.

The participants' expertise were quite varied by multiple factors including their level of training, number of performed operations in operating room, number of performed operations on bony tissue, and the number of operations performed specifically for pedicle screw insertion. Some had more surgical experience with robotic tools. The neurosurgeons less often perform this procedure, performing them sometimes only once or twice a year.

According to the questionnaire, all of the senior and fellow surgeon participants found the haptic training simulator to be a useful tool in teaching probe channelling in pedicle screw insertion. The current device is capable of simulating the various force and torque effects a surgeon feels in this surgery. The current simulator is a first of its kind in the field of spine surgery, with the ability of replicating the haptic sensations in free-hand probe channelling through the bone with high-fidelity haptic feedback.

6.1 Future Works

As previously discussed, breach simulation is an important adverse event that surgeons should be aware of in pedicle screw insertion surgery. Breach simulations, however, were not very effective in this iteration of the simulator due to hardware limitations. Replacing the linear motor with a faster one, and incorporating lighter hardware, may make breach simulations feasible. Also, to increase realism for a more immersive simulation, a graphical interface is necessary. The apparatus can be overlaid by a visual interface that looks like a patient's body, thus, creating a more authentic feel to the simulation. Lastly, since the haptic simulator is planned to serve as a surgical training tool, future work includes determining appropriate training techniques for the surgery and investigating ways of performing surgical skill assessment.

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