Augmenting Contact Stiffness in Passive Haptics – Preliminary Results with Twisted String Actuation

Quang Ha Van and Matthias Harders

Abstract—Haptic feedback is often employed in medical simulators, with the goal to improve user interaction and the training outcome. One option for providing touch sensations is using passive haptics, by including actual physical mock-ups of the anatomical objects in the simulated scene. While this approach has advantages, the usual one-to-one mapping between virtual and physical objects is a fundamental drawback, especially when invasive scene alterations are to be performed, such as cutting or drilling. In this work we propose to alleviate this situation by modifying the mock surgical instruments used for interaction. Twisted string actuation is employed to display variable stiffness while indenting an anatomical model. Quantitative experiments characterizing the performance of a testbed are reported and a prototype system for a surgical bone drill is introduced. Results show that the setup is capable of providing different stiffness augmentations, representing varying bone densities.

I. INTRODUCTION

Haptic feedback is often considered an essential component in medical training simulations [1]. Interfaces providing the feedback can generally be classified into two categories: active and passive. The former employs active mechanisms to deliver the sense of touch (see e.g. [2]), typically through surgical instruments attached to kinesthetic haptic devices. In contrast to this, passive haptics relies on physical props matching the visual virtual environment, without the use of any active mechanisms. This concept has been applied successfully in medical simulations, e.g. [3], [4], but also in more general virtual environments, e.g. [5]. The paradigm implies a one-to-one mapping between the virtual and the associated physical objects, thus resulting in several drawbacks. Scene modifications (e.g. due to tissue cutting) as well as variations (e.g. to represent different patients) cannot easily be incorporated.

In this work we propose to improve the applicability of passive haptics by augmenting surgical tools with novel mechanisms to alter a trainee's perception of interacting with the surgical scene. This is especially well-suited for training of minimally-invasive surgery, since in this case surgeons interact with tissues using instruments instead of their bare hands. The main target is to alter perceived displacements and tissue stiffness while contacting an anatomical mock-up model. Since mechanisms will be housed inside of surgical tools, they should be small and light-weight. Therefore, we examined twisted string actuators (TSAs) [6] as a possible solution. A TSA consists of two or more strings connected

to a motor on one end and to a load on the other end. Motor rotation causes the strings to twist, i.e. to wrap around each other, thus leading to a reduction in length of the string assembly. This results in a pulling force causing a linear motion of the load towards the motor. In this paper we will outline how this mechanisms could be used to vary perceived contact forces on surfaces.

II. RELATED WORK

A number of related lines of research exist targeting the enhancement of touch feedback in surgical training, going beyond standard haptic devices. One possible paradigm is to let a physical mock-up actively change its shape and properties following its virtual counterpart. Instead of a surgical tool, a mock-up anatomy could be adapted using mechanisms such as particle jamming [7], a SmartMesh [8] or a Formable Object [9]. In the first approach, objects are formed through a collection of adaptable cells, while in the latter two, objects consist of nodes connected by active joints. However, the currently available technology cannot easily display objects with high resolution. In addition, costly inverse calculations are required to determine which actuation is necessary to represent a desired shape. Those calculations are also difficult to realize in real-time.

Another reported strategy is to enhance passive haptics by relying on effects of visual dominance. That is, a discrepancy between the visual display of a virtual scene and a physical prop is introduced, with the target to modify perceived touch sensations. In [10], Hirano et al. studied the influence of visual information on stiffness perception during object indentation. Similar work in the context of surgical training has been reported in [11]. Distorting the mapping between the virtual and real environment has been further investigated in [12]; they introduced the idea of redirected touching, akin to redirected walking. This concept has been extended to surgical training by Spillmann et al. [13]. They employ space warping to map different virtual models to a single physical object to represent varying patient cases in a simulator, also addressing scene modifications such as tissue cutting. One drawback of these approaches is that the sensory mismatch is only useful up to a certain, usually small degree. Moreover, the effect of introducing distortions on the final training outcome has not been examined so far.

More related to our work is the concept of haptic augmentation, which has been outlined in [14]. The fundamental principle is to determine the difference between measured actual contact forces and desired force feedback, and to employ a haptic device to superimpose augmentation forces

Quang Ha Van and Matthias Harders are with the Department of Computer Science, University of Innsbruck, Innsbruck 6020, Austria {Ha-Van.Quang, Matthias.Harders}@uibk.ac.at

according to that difference. This gives a user the impression of a different object stiffness (see also [15]). This concept has later been extended to medical simulation, specifically breast tumor palpation in [16]. In these systems kinesthetic haptic devices were used. We intend to apply similar augmentation concepts, however, embedded into a surgical instrument.

Some related work exists that deals with employing vibrotactile feedback in hand-held devices to enhance user perception, e.g. [17], [18], but with a focus on textures. While vibrotactile cues do play a role in user judgements of object stiffness, our work aims at including actual displacements and stiffness changes during contact.

Twisted string actuation exhibits several advantages, such as high reduction ratio, small weight, and low cost. It has been used in different application scenarios, especially when size and weight are of crucial importance, for instance for robotic fingers [19] or in tensegrity robots [20]. We will employ TSAs as a mechanism that can be easily integrated into surgical instrument mock-ups and allow haptic augmentation to alter perceived contact stiffness.

III. STIFFNESS AUGMENTATION USING TWISTED STRING ACTUATION

A. Mechanism Outline

The key idea of our augmentation mechanism is to integrate a TSA in parallel with an elastic compression spring, as illustrated in Fig. 1. The TSA is actuated in such a way that a user pushing on the mechanism (on the right side of the figure) experiences a desired apparent stiffness. Note that this is equivalent to a user holding the mechanism in her hand and pushing with its tip onto a rigid surface instead. Assuming a compression distance ΔX , a spring force with magnitude F_{SPR} and an opposite TSA force magnitude of F_{TSA} , for that specific compression, a user perceives an apparent stiffness, according to Hooke's Law, of

$$k_{APP} = \frac{F_{SPR} - F_{TSA}}{\Delta X}. \quad (1)$$

The difference between F_{SPR} and F_{TSA} corresponds to the interaction force between tool-tip and user, which can be measured with a force sensor. If a certain apparent stiffness k_{APP} is to be displayed with the mechanism, then the TSA has to exert, for a specific compression, a force of

$$F_{TSA} = (k_{SPR} - k_{APP}) \cdot \Delta X. \quad (2)$$

Note that a TSA can only generate pulling forces, therefore, the maximum stiffness displayable with this mechanism is given by the stiffness of the compression spring. In addition to this, the mechanism also exhibits a minimum apparent stiffness limit. For low stiffness, a high TSA augmentation force is required, denoting a large string twisting angle. Beyond a certain rotation maximum, the string assembly will become over-twisted.

For the calculations, knowledge of the compression distance and/or the twisted string contraction length is required. An initial mathematical model of TSAs was derived in [6], and then later improved in [21]. The authors proposed to

Fig. 1: Idea of stiffness augmentation using TSA.

approximate the contraction length X_t of the TSA strings, at time instant t , by:

$$X_t = \sqrt{X_0^2 \left(1 + \frac{F_i}{k_{ST}}\right)^2 - \theta_t^2 r_t^2}, \quad (3)$$

where k_{ST} is the stiffness of each string, X_0 , X_t the initial rest and contracted length of the string assembly, θ_t the accumulated twisting angle, r_t the cross-sectional radius of the cylindrical string assembly in twisted state. For each string, F_i is the axial tension force magnitude, calculated by:

$$F_i = \sqrt{F_\tau^2 + F_{TSA}^2}, \quad F_\tau = \frac{\tau_L}{r_t}, \quad (4)$$

with τ_L denoting the twisting moment and F_τ the force magnitude opposing the twisting moment. The varying radius of the string assembly r_t is given as:

$$r_t = r_0 \frac{X_0}{X_t}, \quad (5)$$

where r_0 is the cross-section radius of the string in untwisted state.

Based on this, the contraction length ΔX could be estimated as:

$$\Delta X = X_0 - X_t = X_0 - \sqrt{X_0^2 \left(1 + \frac{F_i}{k_{ST}}\right)^2 - \theta_t^2 r_t^2}. \quad (6)$$

Unfortunately, two difficulties are encountered with the above modelling. First of all, computing F_{TSA} according to Eq. 2 requires ΔX ; but to obtain the latter F_i is needed, which in turn depends again on F_{TSA} . Moreover, F_i depends on user load and desired stiffness, and cannot be directly measured. Therefore, instead of using Eq. 6 we opted for integrating a linear sensor to measure compression distance ΔX .

B. Control Architecture

Various options to control our outlined mechanism exist. Palli et al. [6] proposed a sliding control strategy for a robotic hand in which $[\theta \ \dot{\theta}]$ is the system state and the pulling force F_{TSA} the system output. This strategy was claimed to be more robust in the presence of measurement noise during continuous set-point variation. However, in [22], Inoue et al. stated that a successful tracking could be realized even with a simple proportional controller. Therefore, for our initial prototype we employ a conventional PID controller.

According to [6] and [21], the pulling force F_{TSA} can be estimated as:

$$F_{TSA} = n \frac{k_{ST}}{X_0} \left(X_t - \frac{X_0 X_t}{\sqrt{X_t^2 + r_t^2 \theta_t^2}} \right), \quad (7)$$

with n being the number of strings. However, due to the previously mentioned problems we follow the simplified model of [6], which neglects string compliance and load dynamics. The pulling force is determined by:

$$F_{TSA} = k_L \left(X_0 - \sqrt{X_0^2 - \theta_t^2 r_t^2} \right), \quad (8)$$

with k_L being the stiffness of an assumed purely elastic load at the end of the string. Thus, we employ the twisting angle θ as a control input to archive desired pulling forces of the TSA.

Note that one possible approach would be to control the stiffness directly. Since the actual perceived apparent stiffness cannot be measured directly, it has to be estimated based on the interaction force and string contraction (Eq. 1). Simple PID control would yield a twisting angle feeding into a position controller. However, such a control architecture for stiffness would suffer from the division-by-zero problem for the rest position where $\Delta X = 0$. In addition, further force model terms such as for damping cannot easily be integrated. Therefore, we propose a control architecture focusing on the force, as illustrated in Fig. 2. A force model is introduced to determine the desired force from desired apparent stiffness and displacement (note: desired quantities are shown with subscript d , measured ones with subscript m). A PID controller is employed to eliminate the difference between desired and measured apparent force, while a second PID controller is part of the inner loop to manipulate the twisting angle θ . Note that the apparent stiffness k_{APP} and the damping factor b are desired properties, while displacement ΔX and compression speed V are measured components.

Fig. 2: Control architecture.

IV. PERFORMANCE EVALUATION

A. Prototype Test Setup

We have built an initial prototype testbed, in order to evaluate the performance of the proposed basic mechanism, shown in Fig. 3. As maximum stiffness in the initial configuration is constrained by the stiffness of the compression spring, a second (extension) spring has been added in opposite direction. The compression and extension springs have a rest length of 36 mm and 32 mm, respectively. A Maxon 118752 DC motor is employed with an industrial motor controller (Escon 36/2 DC), configured for current control. Six strings (0.25 mm diameter cotton yarn fiber) are used for

the twisted string assembly. In order to measure interaction forces, an ATI Nano 17 force sensor is attached behind the device tip. Sensors and controller are connected to a National Instruments NI-6229 acquisition board, housed in a PC running RTAI Linux (10 kHz frequency). The motor encoders (HEDM 5500) are connected directly to the PC, on which the twisting angle is calculated. Also, an external sensing mechanism is employed to obtain compression distance (not depicted).

Fig. 3: Prototype testbed.

- (1) Motor; (2) strings; (3) fixed separator; (4) compression spring; (5) force sensor at tip; (6) extension spring

B. Force Control Evaluation

As described in the previous section, force control will be used instead of directly manipulating stiffness. In this section, we will first examine the capability of the system to generate TSA pulling forces. In order to measure the latter, the device tip has been temporarily fixed and the force sensor has been flipped to the opposite direction.

In the first experiment we examined the relationship between accumulated twisting angle θ and pulling force F_{TSA} . While varying the angle in the range of $[2\pi, 12\pi]$, the resulting pulling forces were measured. Both the collected raw data and a fitted polynomial curve are illustrated in Fig. 4. The non-linear relationship between the quantities can be observed. It should be noted that the amount of twisting angle required for a desired force change depends on the starting angle. Especially for the rest state, a larger force step will require a large change in twisting angle θ , wherefore some delay can be expected.

Fig. 4: Relationship between twisting angle and pulling force.

In a second experiment, the force control was evaluated with two desired time signals: step function and sine-wave. Twisting angles and pulling forces have been recorded over time. Both the measured and the desired quantities are plotted

in Fig. 5. In general the proposed mechanism was capable of controlling the pulling force to follow reasonably well the desired trajectories in both cases. However, two observations have to be made: first, as explained above, there is a time delay for the TSA to start generating a larger pulling force; therefore, a larger response time results for the step function; second, for the sine-wave the strings did not fully untwist again for the zero force case. This can be explained by the different transmission ratios of TSAs in the contraction and relaxation phase, as mentioned in [23].

Fig. 5: Force control evaluation.

C. Stiffness Augmentation

Next, we examined the capability of the mechanism to display different desired augmented stiffnesses. The augmentation should mimic the behavior of a linear elastic spring, with desired forces given by Hooke's law. Desired apparent stiffnesses ranged from soft ($k = 0.05$ N/mm) to hard ($k = 0.8$ N/mm), as governed by the properties of the real springs.

For this experiment the force sensor was returned to the appropriate orientation and the device tip again unconstrained. In order to ensure controlled input, tip motion was generated with the help of a 1D linear robot. The latter was run in speed control, and at the same time used to provide estimates of compression distance.

The robotic mechanism pushed the device tip at a constant speed of $V = 5$ mm/s, to a compression depth of $\Delta X \approx 12$ mm, thus avoiding over-compression. During the indentation, tool-tip displacement and interaction force were measured via the robot and the force sensor. The resulting displacement-force curves are illustrated in Fig. 6. As can be seen, the stiffness of the real system (consisting of the two springs) exhibits initially some non-linearity. For the linear portion the overall stiffness of the system is about 0.9 N/mm. Nevertheless, it was possible to generate different apparent stiffnesses mimicking linear elastic springs, within the discussed maximum and minimum constraints. The drop at the end of displacement-force curves is an artifact caused by the abrupt stop of the external robot motion input mechanism.

Finally, for two selected stiffness cases – soft ($k = 0.05$ N/mm) and hard ($k = 0.8$ N/mm) – both the desired and the measured forces as well as apparent stiffness over time are depicted in Fig. 7. Note that the estimated stiffness is

Fig. 6: Measured displacement-force curves for augmented stiffness, varied in the range of $[0.05, 0.8]$ N/mm

Fig. 7: Force and stiffness tracking.

high at the beginning, due to high interacting forces at small displacements. This is again due to the previously mentioned time delay before the TSA can generate a sufficient pulling force. At the beginning the displayed stiffness is the actual stiffness of the two springs. It may be possible to alleviate this behavior in the future by already introducing an initial twist in the strings in the starting state. Except for this, the proposed mechanism was capable of displaying desired apparent stiffness without significant deviation.

D. Step Response

Next, we examined the capability of the mechanism to change the apparent stiffness during interaction. Two different scenarios were examined – a step from low to high stiffness, and vice versa. For the former, desired apparent stiffness was increased from $k = 0.05$ N/mm to $k = 0.8$ N/mm, while for the latter the step was in the opposite direction. The step was located at a compression depth of 3 mm.

The measured and desired apparent stiffness over time, as well as displacement-force curves are illustrated in Fig. 8. For the upward stiffness step, a considerable overshoot of up to 20% (marked with a circle) can be observed for the estimated signal after the step. This is due to the fact that for an apparent softer object the TSA generates higher pulling forces. A switch to a higher stiffness then results in a lower required augmentation force of the TSA, which

Fig. 8: Step response: stiffness tracking and displacement-force curves.

in turn requires a smaller TSA contraction, and thus causes the strings to partially untwist again. The resulting slight backward movement leads to an increase in the apparent stiffness. This problem was not observed in the second case, since the TSA moves forward to increase the pulling force for a downward step. The displacement-force curves indicate that two separate regions of different stiffness can be generated.

V. PROTOTYPE DEMONSTRATOR – STIFFNESS AUGMENTATION FOR SURGICAL BONE DRILL

Drilling has a number of applications in the medical field, ranging from dental procedures to ligament reconstruction [24]. Most existing systems make use of active interfaces to provide touch feedback, see e.g. [25], [26]. We outline a prototype demonstrator for passive haptics that employs our proposed stiffness augmentation to mimic drilling into bone tissue of different stiffness. Both the mechanical design and augmentation strategies are outlined.

We propose to integrate the previously introduced mechanisms into a mock-up of a surgical drill (concept shown in Fig. 9). A separate linear position sensor will be employed to measure the displacement of the tool-tip. To reduce friction between the tool shaft and its frame, plain bearings are added. Depending on the desired maximum stiffness to be displayed, an extension spring could also be placed into the tool frame.

Human bone stiffness varies not only with gender, age, and pathology [27], but also by layer, such as with cortical and spongy bone. For simplicity, we assume only two layers of bone, a hard layer ($k_1 = 0.8 \text{ N/mm}$) with 3 mm thickness followed by a softer one ($k_2 = 0.05 \text{ N/mm}$). Note that any covering deformable tissue is not considered. Friction between tool and bone is modelled by dampers ($b_1 = b_2 = 0.3 \text{ Ns/mm}$). The computed desired forces are displayed via the enhanced surgical drill tool, while it is in contact with a rigid plastic bone. The varying forces experienced by the user while pushing the tool onto the bone thus represent the variable stiffness, and damping, perceived during a drilling process.

Fig. 9: Prototype model of a drill device with TSA. (1) motor; (2) string mount; (3) compression spring; (4) force sensor; (5) frame; (6) motor holder; (7) twisted string spring space; (8) tool shaft; (9) fixed separator; (10) bearing; (11) linear sensor

Fig. 10: Model of drilling through multilayer bone.

While drilling through the first, soft layer, the system acts as a compression spring with stiffness k_1 and a parallel dashpot with damping b_1 , as illustrated in Fig. 10. When penetration depth, i.e. compression of the assembly inside the tool, exceeds the first layer's thickness (3 mm), a stiffness k_2 and a damper b_2 is added. Thus, the rendering forces are determined as

$$F_d = \begin{cases} -k_1 \Delta X - b_1 V, & \text{phase 1} \\ -k_2 \Delta X - (b_1 + b_2) V, & \text{phase 2.} \end{cases} \quad (9)$$

Fig. 11: Two layer stiffness augmentation during drilling with different feed rates varying in the range of $[1.0, 6.0]$ mm/s.

Using this model, again displacements and forces were recorded while the 1D robot performed the simulated drilling into a two-layered bone. Data were obtained for various feed rates, ranging from $V = 1.0 \text{ mm/s}$ to $V = 6.0 \text{ mm/s}$ (chosen based on the study in [28]). The results are illustrated in

Fig. 11. It can be seen that also in this case the mechanism was capable of rendering different stiffness for different layers. Moreover, due to the damping, for higher velocities, higher forces were experienced. Still, the same artifact due to initial twisting time delay at the beginning can also be observed.

VI. CONCLUSIONS AND FUTURE WORK

In this paper, haptic augmentation was employed to enhance the applicability of passive haptics. A mechanism comprising of a light-weight, cheap and compliant TSA and a compression spring was used to display varying stiffness with a single mock-up model. Experiments were carried out to evaluate the performance of the proposed system, both for fixed and variable stiffness. As example application, a model for rendering different stiffness while drilling through different bone layers was described.

Even though the proposed mechanism shows promising performance in manipulating perceived apparent stiffness, several problems still need to be addressed. The TSA can only generate pulling forces, thus constraining the maximum apparent stiffness with this mechanism to the compression spring stiffness. In addition, the minimum stiffness is limited by the maximum twisting angle before the TSA becomes over-twisted. One solution to render larger stiffness ranges is to employ another TSA to generate forces in the opposite direction [29]. However, problems due to weight and mechanical complexity would arise, since an additional motor would be used. Other artifacts, such as initial time delay of force application as well as assembly motion during stiffness change also need to be investigated.

Furthermore, a surgeon's judgment of tissue stiffness is also influenced by auditory and vibrational feedback [30], especially while drilling through different tissues. In future work, the proposed method will be combined with other feedback to improve the realism of the simulation.

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