

Vibrotactile force feedback system for minimally invasive surgical procedures

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Abstract—Lack of adequate force feedback for the surgeon in minimally invasive surgery (MIS) can lead to unnecessary trauma to tissue and adverse events during surgery. The successful use of vibrotactile stimulation to augment overloaded or deficient sensory modes in the human operator in other application areas warrants an investigation into its application in MIS. A vibrotactile force feedback system was designed, and its ability to provide useful force information to subjects performing a simulated MIS task was evaluated. Results showed that the system responds as predicted against the bottom surface of the foot, and that subjects were able to perceive a linear increase in force as linear increase in vibration intensity. Furthermore, vibrotactile force information increased one's sensitivity to tissue contact (1.3 N maximum force – no vibration, 1.0 N maximum force– fine step vibration feedback; $p < 0.001$) and improved one's ability to consistently and accurately differentiate tissue softness in a simulated MIS task. Vibrotactile force feedback in MIS appears to have benefits which can lead to a decrease in trauma to tissue and adverse events.

I. INTRODUCTION

In minimally invasive and telerobotic surgery the surgeon's task is made more difficult by the lack of haptic feedback, and a distortion in force feedback. These inherent complicating factors in minimally invasive surgery (MIS) increase the number of complications and occurrences of adverse events compared to traditional open surgery [1]. Additionally, in the absence of force feedback (as in robotic surgery), or presence of distorted force feedback (as in laparoscopic surgery), surgeons apply more force to tissue, are in contact with tissue for longer periods of time, and have difficulty differentiating between materials of differing softness [2]. Tissue which is manipulated for longer periods of time, and with larger force magnitudes will incur more severe trauma, which results in more discomfort to the patient, an increase in hospital stay length, and an increase in hospital costs. The magnitude of force required to cause significant trauma to tissue, enough to prolong hospital stays and cause complications, has been identified to be approximately 5.3N [3], which is well within forces measured during laparoscopic procedures (0-12 N) [4]. Thus, it is important to decrease the magnitude of forces applied to tissue, and the time in contact with tissue.

Many attempts have been made to provide surgeons with accurate force and torque information by means of sensors and force feedback control systems [5]. Tactile sensors have also been developed to give surgeons information about the material characteristics of the tissue they are manipulating [6]. From a systems perspective, these solutions have a major drawback. That is, tactile sensors and force-torque sensors provide the surgeon with information through the visual channel that is already overloaded in MIS. The surgeon is not only at a disadvantage visually as a result of the lack of depth information (i.e., lack of binocular disparity and shadows), but must also be aware of every detail of the image to guide the surgical procedure. Thus, requiring a surgeon to infer force information via the visual sensory mode in addition to monitoring the surgical site is undesirable.

More desirable is the option of utilizing another sensory channel to provide the force information. Based on Wickens' multiple resource model, one's task performance is improved with the use of multiple sensory modalities [7]. In other words, all else being equal, the more sensory modalities a human is able to utilize while performing a task, the greater their ability is to perform a task at a high performance level. Thus, when attempting to alleviate this problem of inadequate force feedback, one should look to the under utilized sensory modalities, such as the vibrotactile sense. Studies in other application areas have shown that vibrotactile feedback can successfully provide information, and improve task performance in subjects with a deficient, or overused visual channel. For instance, one's ability to navigate was shown to improve with the addition of tactile back cues [8]. Additionally, deaf people have been able to improve their lip reading ability with the aid of a tactile hearing aide [9].

The goal of this study, motivated by the promising enhancement capabilities of vibrotactile feedback, was to design and evaluate a vibrotactile feedback system which can provide meaningful force information to surgeons. Two experiments were conducted. First, an experiment was conducted to determine the psychophysics of vibrotactile signal perception. Second, the ability of vibrotactile force information to improve one's ability to control force application, differentiate tissue softness, and perform more efficiently during MIS was investigated.

II. VIBROTACTILE FORCE FEEDBACK SYSTEM

The vibrotactile force feedback system used a sensor to first measure the forces and torques applied to tissue. The force and torque data were then passed through a function

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which converted them into a vibration amplitude. The resulting vibration signal perceived by the user was proportional to the magnitude of the force being applied to tissue.

A. Force Sensor

The ideal force sensor for the vibrotactile system must be able to sense an appropriate range of forces, and have a suitable resolution. The forces encountered during MIS have been measured to range, on average, from 3-6 N, but may reach magnitudes as high as 20 N [7]. Thus, for these experiments, a mini-40 force/torque transducer from ATI-IA, Inc., was used. It is capable of sensing forces between 0-65 N with a resolution of 0.003N. This force/torque transducer was chosen for its applicability to all MIS procedures. For more specific, extremely delicate surgical tasks, a sensor with a higher resolution and smaller force range would be more appropriate.

The force/torque sensor was integrated into the shaft of the surgical tool, near the bottom, so that only the forces at the tool-tip would be measured. It should be noted that additional care must be taken when integrating the sensor into the surgical tool if it is necessary to utilize the grasping mechanism. In that case, one must take into account an additional force from a rod which moves in and out of the outer shaft to actuate the grasping mechanism. As this rod is pushed and pulled, it puts tensile and compressive forces on the outer shaft at the grasping mechanism pin joint (figure 1).

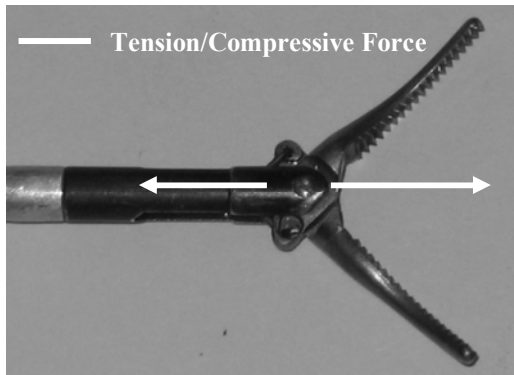


Fig. 1: Tension/Compressive Force on outer shaft due to actuation of grasper mechanism

B. Vibration Signal

Humans can perceive both frequency and intensity changes in a vibration signal. Consequently, it is possible to vary the frequency or the intensity of the vibration signal as a function of force to present the force information to the surgeon. However, data show that humans are better at discriminating between different vibration intensities than vibration frequencies. A 30% increase in frequency is necessary for humans to perceive a change in frequency [8], as opposed to a 0.4 - 2.3 dB increase in intensity [9]. As a result, the vibrotactile system was designed to vary only the vibration intensity as a function of force. A 250 Hz square wave with a small amount of stochastic resonance was used to drive the vibration device. It has been shown that humans are most

sensitive to vibration at this frequency [9], and addition of stochastic resonance improves sensitivity to vibration [10].

C. Output Location

The location for attaching the vibration output device must be considered carefully. The fingers, hands, and forearm have the lowest vibration intensity thresholds. Ideally, the vibration device would deliver a vibration stimulus to this area of the body. However, it has been shown that one's sensitivity to vibration is reduced when that area of the body is active [11]. If a vibrotactile signal were applied to the fingers, hands or forearm, not only would the surgeon be desensitized to the vibration while busy performing the surgery, but it would distract the surgeon during surgery. The next most sensitive, and practical, area of the body is the foot. Consequently, the vibration signal was applied to the bottom of the foot in the arch area.

D. Vibration Output Device

The vibration transducer used in the vibrotactile system was a modified Tactaid II by Audiological Engineering, Inc. (figure 2). The device consisted of a U-shaped steel member rigidly fixed to a lower plastic housing. Adhered to the underside of the top portion of the steel member was a magnet. Below the magnet was a coil of very thin wire. As an alternating current was applied to the coil of wire a magnetic field was generated, inducing an alternating force on the magnet, and causing the steel member to vibrate. In its original form there was an upper plastic housing which enclosed the steel member, magnet, and coil of wire. In order to effectively apply the vibration to the arch area of the foot, the upper housing was removed and the lower housing was rigidly fixed. As a result, the steel member, which would now be in direct contact with the foot, would apply the vibration signal. A platform, which the subjects can stand on, was designed to support the foot (figure 2).

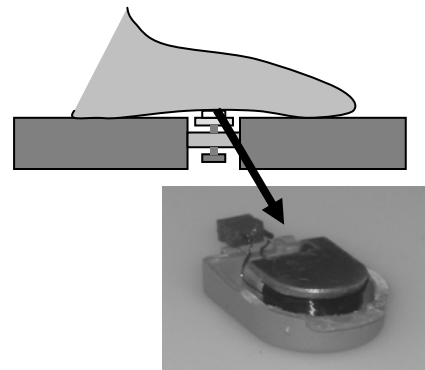


Fig. 2: Modified vibration transducer and platform schematic

III. VIBROTACTILE PERCEPTION

The successful implementation of a vibrotactile force feedback system in MIS is highly dependent on one's perception of the vibrotactile signal. A linear increase in force should be perceived as a linear increase in vibration intensity. Achieving the desired perception is not trivial for several reasons. First, it is known from Weber's law that the

magnitude change of a stimulus necessary to perceive a change in that stimulus increases with the magnitude of that stimulus. Thus, perception is not a linear phenomenon. Second, each surgeon's foot geometry and skin properties will be different, which may result in vibrotactile perception differences. Several steps were taken to achieve the desired vibrotactile perception.

A. Vibrotactile Device Response

First, the response of the vibrotactile device was investigated by placing a B&K 4393 accelerometer on top of the steel member. The vibration device was then driven at 250 Hz, while the voltage amplitude of the square wave was varied from 0-2V in 0.05V increments. Performing a fast Fourier Transform on the acceleration data, and dividing by the frequency (radians) squared, the displacement output for each data point was calculated. The results showed that the response of the device was linear up to approximately 1V. Second, a frequency sweep from 10-1000 Hz was performed at a constant amplitude of 0.2V, peak to peak. From this data a spring constant and electro-mechanical constant (Kt) of the system were extracted.

B. System Model

A simplified model of the system was developed to predict the gain of the system in actual experimental conditions. First, the mechanical system was modeled as a simple spring mass system, with the steel member acting as a spring, and the accelerometer and upper portion of the steel member, the mass. The electro-mechanical constant, Kt, was then massaged and an additional constant (C) was added to the system so that the model better matched the experimental data in the high frequency range (100-550 Hz). Next, a spring and damper in parallel was added to the system to model skin. It is known that the skin is highly non-linear. However, modeling it as a linear system was a reasonable starting point. The electrical part of the system was modeled as a system resistance and inductor in series. However, the resistance of the system was much larger than the inductance, thus the impedance due to the coil of wire was ignored. The transfer function of the model was:

$$\frac{X(s)}{V_{Applied}} = \frac{K_t}{R_{sys}} \left[\frac{C}{M_{sys}s^2 + (K_{sys} + K_{skin}) + b_{skin}s + \frac{K_t^2 s}{R_{sys}}} \right] \quad (1)$$

where X(s) is the displacement output in meters, V_{Applied} is the amplitude of the square wave driving the device, K_{sys} is the spring constant of the steel member, M_{sys} is the mass of the upper half of the steel member, K_{skin} is the spring constant of skin, and b_{skin} is the damping coefficient of skin.

From table 1, one can observe that the output of the model in experimental conditions was evaluated over a range of different skin constants. The skin constant data were extracted from a study which determined spring and damping coefficients of in-vivo pig liver under two different pre-load conditions. The contactor used had a 5 mm² area [12].

Table 1: Model constants

K _t (V/m/s)	C	R _{sys} (Ω)	K _{sys} (N/m)	M _{sys} (kg)	K _{skin} (N/m)	b _{skin} (N-s/m)
3.62	2.2	32	2631	0.0034	25-3400	1-17

In order to utilize this data it was assumed that the skin constants would vary with the pre-load and linearly with contact area. However, neither the position, nor the pre-load of the vibrotactile device was controlled for testing of the system. This was because of the expected variability in foot geometry, body fat percentage, and skin elasticity, which was assumed to affect the skin constants. Thus, subjects were instructed to position their foot in a manner in which they were most sensitive to the vibration output of the device. As a result of the expected variability of skin constants between subjects a range of constants were examined, assuming that as the skin constant increased the damping constant increased proportionally. The ranges of gain values at 250 Hz, determined by the model, were then calculated. These values were used, along with vibrotactile intensity just noticeable difference (JND) data, to predict at which voltage amplitudes subjects would perceive a change in vibration intensity (figure 3). As previously mentioned, research has shown the JND to be between 0.4 and 2.3 dB. For the purpose of this study, a conservative estimate of 1.5 dB was used.

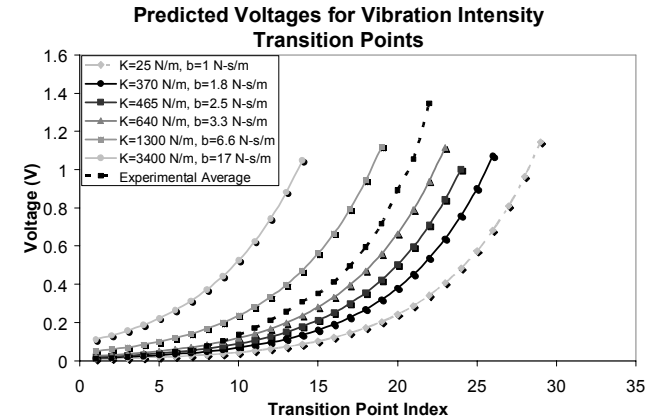


Fig. 3: Predicted vibration intensity transition points

C. Psychophysics Experiment

An experiment was conducted to determine at which voltages the vibration intensity transition points actually occurred. The experiment consisted of exposing subjects to incremental vibration intensity changes, and receiving verbal feedback from them when they were able to detect the vibration intensity change. An experiment began with a baseline voltage of 0 volt. The voltage would then be increased from 0 to some value and the subjects would give their feedback. If they did not feel a change then the voltage would be brought back down to 0 volts and it would be increased by a slightly higher amount. If they did perceive a difference the voltage would be returned to zero and the next change in intensity would be slightly less than what was applied before. The experiment would proceed in this manner until the smallest voltage change at which an intensity

difference was perceived was found. This would be the first transition voltage. The experiment would then proceed as before with the first transition point as the new baseline voltage. The experiment was conducted on four subjects. The number of transition points ranged from about 15 to 25, with an average of about 22 transition points (figure 4).

The variance between subjects was very similar to what was predicted. It is speculated that differences between pre-load and physical characteristics of each individual's foot contributed to the range in transition points observed.

The average of the experimental data was then used to create a function to convert the force measured by the sensor into a waveform amplitude. This was accomplished by dividing the force range expected during a procedure into 22

augmentation would result in the same or better performance in force detection and differentiation, and that an intermittent form of vibration signal would be more useful for attenuating force application than a continuous signal.

A. Task

The task was designed to simulate needle driving in laparoscopic suturing. The task required subjects to first locate a small target (3 mm diameter) with a needle, and then penetrate a double layer silicone gel mass consisting of a soft upper layer and harder lower layer, until they perceived the harder layer. The soft layer was equivalent to fatty tissue, while the harder layer was equivalent to human liver in compliance.

B. Experimental Design

The independent variables of the experiment were vibration feedback and audibility (the vibration device also generated a low audible sound). There were four levels of vibration feedback (continuous, fine step, crude step, no vibration) and two levels of audibility (on, off). Continuous vibration feedback consisted of feedback in which the vibration intensity was a continuous function of the force measured. The step conditions consisted of feedback in which the vibration intensity increased in steps as a function of force. For the fine step condition, there were 12 different vibration levels, while for the crude step condition, there were 6 different vibration levels. The two levels of audibility were on and off. The on level meant that subjects were able to feel and hear the vibration output. In the off level, sound deadening ear muffs were placed on subjects to substantially reduce the audibility of the vibration output. The dependent variables were maximum force, penetration depth, and time to detection. The experiment was a pseudo-randomized, Latin square design. Twelve untrained subjects performed 10 trials in each of the eight possible conditions. Five different gel masses with varying layer thicknesses were presented randomly to the subjects to eliminate a learning effect.

The dependent measures were time to detection, error in penetration depth, and maximum force applied.

C. Results

The maximum force results showed that there was a significant effect found in the vibration feedback factor ($F(3, 932) = 21.93, p < 0.001$). Subjects reached higher force levels in the no vibration condition compared to all other conditions (figure 5).

Furthermore, the error in penetration depth, which is the difference between the penetration depth and the depth of the first gel set layer, was calculated and analyzed. The results showed a significant effect in the variance of the penetration depth error across the vibration feedback factor ($F(3, 952) = 18.07, p < 0.001$). The variance in the no vibration feedback condition was much larger compared to all other vibration feedback condition (figure 6).

The average absolute value of the error was greater in the no vibration condition compared to the vibration feedback conditions (figure 7). Statistical t-tests showed significant differences between the no vibration and continuous ($t(238) = 3.82, p < 0.001$), fine step and no vibration ($t(238) = 3.97, p$

Vibrotactile Perception Transition Points

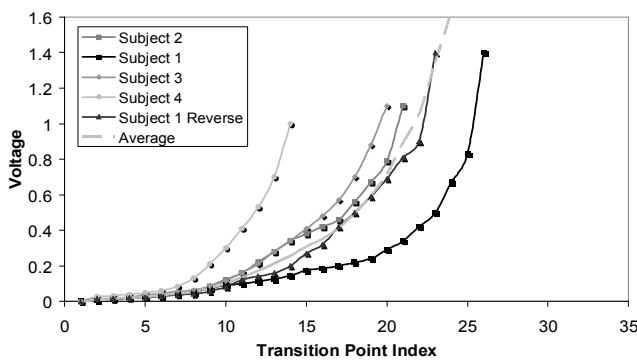


Fig. 4: Experimental vibration intensity transition points

(average number of transition points) data points. The voltage amplitudes at which a transition point occurred were then plotted as a function of the 22 force data points. A polynomial curve fit was performed to determine the function.

Despite the fact that some subjects did not perceive a linear increase in force as a linear increase in vibration intensity due to individual differences, this method created the desired approximate perception for each subject. To ensure that each subject would perceive a linear increase in force as a linear increase in vibration intensity, additional studies would need to be conducted, and additional parameters controlled. How skin constants depend on factors such as contactor area, pre-load, the area of the foot, as well as the non-linear nature of foot tissue would need to be known. A different force-amplitude conversion function would need to be determined for each subject depending on the response of his/her foot to the vibration device. Additionally, the area of the foot that the device is in contact with, as well as the pre-load, should be more precisely controlled.

IV. TASK PERFORMANCE

A task was specifically designed in the second experiment to determine if a subject's ability to control force application, differentiate between tissue softness, and perform more efficiently was improved with the aide of vibrotactile feedback. It was hypothesized that the vibrotactile sensory

< 0.001), and crude step and no vibration ($t(238) = 2.95, p = 0.0018$) conditions.

There was no difference between the audible and non-audible conditions.

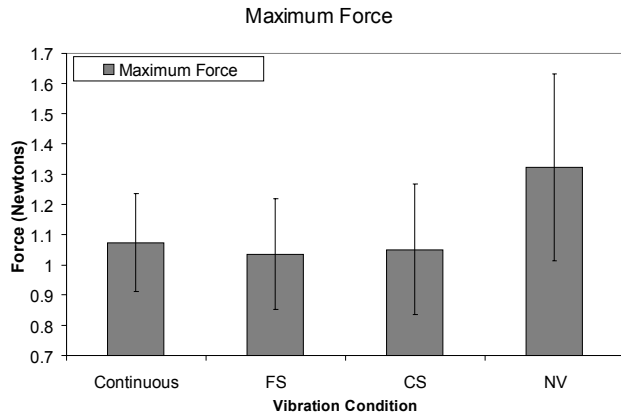


Figure 5: Average maximum force

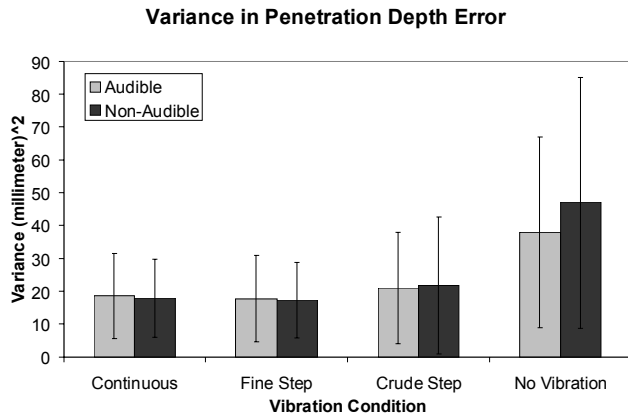


Figure 6: Variance in penetration depth error

D. Discussion

The results indicate that subjects were better able to control their force application with the addition of vibrotactile force feedback. They were also more accurate and consistent in determining the location of the second, harder gel layer, suggesting that the vibrotactile force feedback increased their sensitivity in tissue differentiation. However, further inspection of the data revealed that this may not be true. If one inspects the maximum force reached for each vibration condition and gel set, it shows that the average force was approximately the same, and close to the force which produced the maximum vibration output (1.1N). Presumably, if subjects were using a rate change in vibration intensity to determine the location of the harder layer, then one would expect the maximum force to increase as the first gel layer increased in thickness. However, this was not observed, suggesting that subjects predominantly relied on the absolute magnitude of the vibration intensity to determine the approximate location of the harder layer. It is believed that the task design contributed to this result. A frictional force between the needle and gel, in addition to inconsistent penetration techniques may have led to a gradual intensity increase at the gel layer interface rather than a sharp increase. Thus, subjects were unable to determine precisely where the

harder layer was, but were able to use the absolute magnitude of the intensity to determine an approximate location. Thus, although it appears that a subject's ability to differentiate between tissue softness was improved, a definitive conclusion cannot be made.

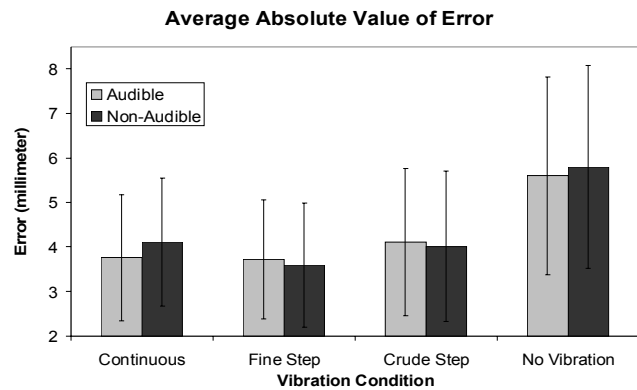


Figure 7: Absolute value of error

V. CONCLUSION

The vibrotactile force feedback system presented in this paper was able to deliver usable force information to its users. Subjects were able to accurately perceive, make sense of, and react to a vibrotactile signal which varied with the forces encountered at the surgical tool tip. This was supported by the subjects' improvement in force application control, and their ability to more accurately and consistently locate the harder layer when receiving vibration feedback.

Despite the improvements shown by this study, further studies should be done to validate and build on these results. Specifically, a much more in depth analysis of how the vibrotactile device response varies between subjects, and *why*, could be conducted to ensure each subject perceives a linear increase in force as a linear increase in vibration intensity. Furthermore, additional studies with different tasks should be conducted to further validate that one's force application control is improved with vibration feedback.

In conclusion, vibration force feedback has some benefits in MIS. Better force application control can reduce the probability that surgeons apply forces above a magnitude which can cause unnecessary tissue trauma. A reduction in tissue trauma can lead to shorter hospital stays and less discomfort for the patient, as well as a reduction in the probability of an adverse event. Other applications in which this technology may be beneficial include telerobotic surgery where the surgeon currently receives no force feedback, extremely delicate surgical tasks, and navigational procedures, such as a stent placement procedure.

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