

## VIBROTACTILE FEEDBACK ENHANCES FORCE PERCEPTION IN MINIMALLY INVASIVE SURGERY

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Distorted force feedback in minimally invasive surgery causes the procedure to become more difficult for the surgeon. A simulated tissue probing task was designed to test the hypothesis that vibrotactile feedback can enhance one's ability to differentiate tissue softness, and control the forces being applied to tissue. The two independent factors in the study were vibration feedback, consisting of four levels (continuous, fine-step, crude-step, and no vibration), and audibility, consisting of two levels (on and off). The results demonstrated that with the aide of vibration the absolute probing depth error was reduced (5.7mm – no vibration, 3.65mm – fine step), and the average maximum force applied was reduced (1.32 N – no vibration, 1.04 N – fine step). Additionally, the normalized time to detection (0.93s/s audible, 1.10s/s non-audible) and maximum force (1.16 N – non-audible, 1.08 N – audible) was reduced in the audible condition. These results indicate that vibrotactile stimulation is a viable substitute for force feedback in simulated minimally invasive surgery.

### INTRODUCTION

There are limitations to the technique of minimally invasive surgery (MIS) which can make it more difficult and less safe (Targarona, 1998). One such limitation is the distortion in force feedback that the surgeon receives. A major cause of force feedback distortion is the friction force present between the trocar and the surgical tool [Picod, 2005]. It masks the true force being applied to the tissue, desensitizing a surgeon's force perception. Other sources of force feedback distortion, which result in similar consequences, are: a reactionary torque from the abdominal wall, the lever created at the point of entry, friction in the surgical tool's grasping mechanism, and inherent levers in the surgical tool mechanism (Picod, 2005, Macfarlane, 1999). Distorted force feedback makes it difficult for surgeons to differentiate between tissues of differing hardness, causes higher forces to be placed on the target tissue, and results in longer time in contact with tissue (Perreault & Cao, 2006).

In devising a solution to distorted force feedback in minimally invasive surgery, task performance as a function of alternate sensory augmentation was investigated. Sensory augmentation, by means of haptic feedback, is a technique that has been used in other areas of

application to enhance human performance. For instance, haptic feedback has been studied extensively in teleoperation tasks, such as controlling unmanned vehicles. Persons who control unmanned vehicles find it arduous to track remote objects as a result of the vehicle's movements in remote space, delay in feedback, and low image resolution. However, when continuous haptic information was applied to the joystick, operators were better able to track objects, demonstrated by a reduction in their tracking error (Korteling, 1998). Furthermore, the lateral position error and driver effort in controlling automobiles have been shown to decrease when an active steering wheel, which provides lateral vehicle position information by means of haptic feedback to drivers, is utilized (Schumann, 1994).

One specific form of haptic feedback is vibrotactile feedback, which has been shown in other applications to improve task performance. For example, there is evidence that vibrotactile feedback can be used as a tool to improve one's navigational ability (Hong, 2003). Therefore, we propose that vibrotactile feedback can be used to provide surgeons with additional force information. Our goal was to examine the effect of vibrotactile stimulation as a source of force information on an operator's performance in a simulated MIS task. In particular, the performance measures of interest

were force application control, ability to differentiate between tissue masses of differing compliance, and time to task completion. It was hypothesized that additional force information in the form of vibration would result in the same or better performance in force application control and differentiation. It was further hypothesized that an intermittent form of vibration signal is more useful for differentiating between tissue softness in a probing task than a continuous signal.

## METHODOLOGY

### Simulated Surgical Task

A task was designed to simulate tissue probing or needle driving during suturing in MIS. Subjects were required to locate a small round opening in a flat horizontal surface (approximately 2.5 mm in diameter) with a blunt needle, in a laparoscopic trainer box. Once located, subjects penetrated the double-layer silicone gel mass beneath the surface with the needle. The top gel layer was the “soft layer,” whose compliance may be compared to fatty tissue. The bottom layer was the “hard layer,” with compliance comparable to that of human liver. Five different gel sets with different top layer thicknesses were used in the experiment to avoid learning by the subjects. The subjects were instructed to halt penetration once they felt the harder second layer. A modified surgical tool, fitted with a blunted needle and force/torque sensor (ATI mini-40), was used to perform the task. An endoscope was used to display the image of the workspace on a video screen positioned in front of the subject.

### Vibrotactile Feedback System

While subjects performed the probing task in the vibration conditions, they received vibrotactile feedback, through a vibration device, with an intensity proportional to the force at the needle tip-tissue interface. The incoming force data (pounds force) from the force torque sensor were converted into a square wave function (Volts), which served as the input for the vibration device. The function which converted the force data into a square wave amplitude was determined through a dynamic analysis of the vibration device, and an experiment which determined estimates for amplitude voltages corresponding to vibration intensity transition points. With the transfer function, subjects

perceived a linear increase in force as a linear increase in vibration intensity.

The vibration device (a modified VBW32 skin transducer by Audiological Engineering, Inc) was mounted on an aluminum piece which protruded slightly above the surface of a flat platform. During the experiment, subjects were asked to stand with one foot on the platform such that the vibration device was in contact with the skin in the arch area of the foot. The exact position of the device was not specified due to differences in foot geometry, fat content, and skin compliance between subjects. Subjects were instructed to position their foot in a manner in which they were most sensitive to the vibration output.

### Experimental Design

The experiment was a 4 (vibration: continuous signal, fine step signal, crude step signal, no vibration) x 2 (audibility: on, off), pseudo randomized, repeated measures design. The fine step vibration signal was a discontinuous signal, in which 12 evenly spaced input force ranges corresponded to a single vibration output. The crude step signal was similar to the fine step, with only 6 possible vibration magnitudes instead of 12. The audibility “on” condition meant that one was able to feel and *hear* the vibration signal from the vibrating device, while in the audibility “off” condition, the signal was not audible. Each of the twelve subjects was required to go through a training session to become familiarized with the setup. The experiment consisted of 10 trials in each of the eight conditions. Additionally, one of five different silicone gel sets with varying silicone gel layer thicknesses was randomly used for each trial to eliminate learning from memory. The dependent variables of the experiment were the penetration depth, maximum force applied, time to detection, and average slope of the force profile.

## RESULTS

### Penetration Depth

The error in penetration depth, defined as the deviation from the depth of the hard layer, was calculated from the penetration depth data. A two-way analysis of variance showed that there was a significant effect in the vibration factor ( $F(3, 952) = 3.13, p = 0.025$ ). A post-hoc Tukey test revealed a

significant difference between the fine step and the no-vibration conditions.

Since the task was a localization task, positive and negative error values were able to cancel each other, and tended to bring the means close to zero. Consequently, the absolute value of the error and variance in the gross error were also analyzed (see Table 1). T-tests showed a significant difference between the continuous audible and no-vibration audible ( $t(238)=3.82, p<0.001$ ), continuous non-audible and no-vibration non-audible ( $t(238)=3.43, p=0.001$ ), fine step audible and no-vibration audible ( $t(238)=3.97, p<0.001$ ), fine step non-audible and no-vibration non-audible ( $t(238)=4.52, p<0.001$ ), crude step audible and no-vibration audible ( $t(238)=2.95, p=.0018$ ), and crude step non-audible and no-vibration non-audible ( $t(238)=3.46, p<0.001$ ) conditions.

Table 1: Absolute error in penetration depth

Absolute Error in Penetration Depth (mm)					
	Continuous	Fine Step	Crude Step	No Vibration	Mean
Audible	3.76	3.72	4.11	5.60	4.30
Audible Off	4.11	3.59	4.01	5.80	4.37
Mean	3.94	3.65	4.06	5.70	

Paired t-tests were conducted on each of the four vibration conditions across the audible factors. There were no significant differences found.

An ANOVA was also conducted on the variance in the error. A significant effect was found in the vibration factor ( $F(3, 952)=18.07, p<0.001$ ). The variance in error was much larger for the no vibration condition ( $42.43 \text{ mm}^2$ ), compared to all other vibration factor levels, but most notably the fine step level ( $17.44 \text{ mm}^2$ ). A post-hoc Tukey test indicated a significant difference between the no-vibration and each of the three vibration conditions.

**Maximum Force**

A three way ANOVA, with vibration, audibility, and gel set as the three factors was conducted. Significant effects were found in the vibration ( $F(3, 932)=21.93, p<0.001$ ) and audibility factors ( $F(1, 932)=10.06, p=.0016$ ) (see Figure 1). The mean maximum force was lower in the audible condition ( $1.08 \text{ +/- } 0.4 \text{ N}$ ) compared to the non-audible condition ( $1.16 \text{ +/- } 0.51 \text{ N}$ ). The mean maximum

force was largest in the no vibration condition ( $1.32 \text{ +/- } 0.62 \text{ N}$ ), and least in the fine step condition ( $1.04 \text{ +/- } 0.37 \text{ N}$ ). A post-hoc Tukey test indicated that there was a significant difference between the no-vibration condition and each of the three vibration conditions.

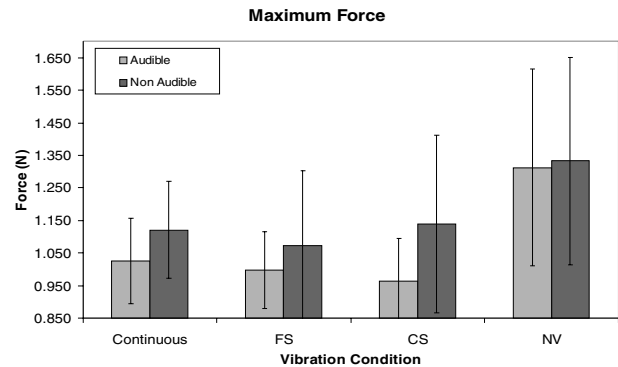


Figure 1: Maximum Force

**Detection Time**

The detection time data was normalized to eliminate large variances between subjects due to different penetration techniques. Each subject's data were normalized by dividing each data point by the average time to detection for that subject. A 2-way ANOVA showed significant effects in the audibility factor ( $F(1, 932)=37.26, p<0.001$ ), and in the gel set factor ( $F(4, 932)=8.82, p<0.001$ ). The normalized mean detection time for the audible condition was  $0.93 \text{ s/s}$  ( $SD=0.41 \text{ s/s}$ ), and  $1.11 \text{ s/s}$  ( $SD=0.49 \text{ s/s}$ ) for the non-audible condition.

**Average Slope**

The average slope is the maximum force applied divided by the time spent to reach that force minus the time spent in removal of the tool and recovery of the force magnitude lost during removal. A 3-way ANOVA showed a significant effect in the vibration ( $F(3, 932)=32.15, p<0.001$ ) and gel set factors ( $F(4, 932)=10.67, p<0.001$ ). The means of the average slope data for the continuous, fine step, crude step, and no vibration conditions were  $0.34 \text{ N/s}$  ( $SD=0.16 \text{ N/s}$ ),  $0.32 \text{ N/s}$  ( $SD=0.15 \text{ N/s}$ ),  $0.33 \text{ N/s}$  ( $SD=0.13 \text{ lbf/s}$ ), and  $0.46 \text{ N/s}$  ( $SD=0.28 \text{ N/s}$ ), respectively. The means of the average slope data for gel set 1, 2, 3, 4, and 5 were,  $0.43 \text{ N/s}$  ( $SD=0.25 \text{ N/s}$ ),  $0.38 \text{ N/s}$  ( $SD=0.17 \text{ N/s}$ ),  $0.34 \text{ N/s}$  ( $SD=0.16 \text{ N/s}$ ),  $0.34 \text{ N/s}$  ( $SD=0.17 \text{ N/s}$ ), and  $0.32 \text{ N/s}$  ( $SD=0.19 \text{ N/s}$ ), respectively.

## DISCUSSION

In general, our results show that with the aide of vibrotactile feedback a subject's ability to control force application and differentiate between materials of differing softness is improved. In addition, subjects were better able to control their force application and improve their time performance when they were able to feel *and* hear the vibration signal. Closer examination of the data in the vibration conditions suggests that subjects may have relied predominantly on the magnitude of the signal rather than the intensity rate change of the signal. For instance, when one examines the average maximum forces measured for each of the vibration conditions and gel sets (see Figure 2) it can be observed that the average maximum force did not significantly change across gel sets.

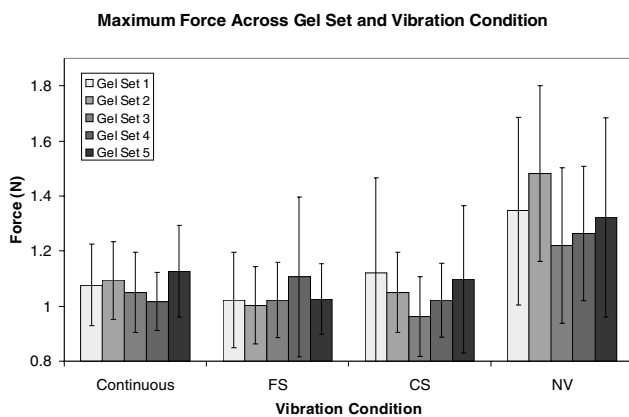


Figure 2: Maximum force across gel set and vibration condition

The average for each condition in which subjects received vibration feedback was very close to the force which produced the maximum vibration intensity output (1.1 N). Additionally, subjects consistently penetrated too deeply into shallow gel sets (1, 2), and not deeply enough into gel sets with deeper top layers (4, 5) (see Figure 3). Such a phenomena is what would be expected if subjects relied on the absolute magnitude of the intensity instead of a rate change. We speculate that inherent factors of the task contributed to this. For example, the friction between the gel and needle, along with inconsistent penetration techniques between trials, may have contributed to creating a smooth intensity transition from the soft layer to the hard layer. The smooth intensity transition, rather than a sharp rate change in intensity, may have created some uncertainty as to the exact location of the hard layer.

Despite this, subjects were able to use the absolute magnitude of the vibration intensity to gauge approximately, but not precisely, where the hard layer was. Therefore, vibrotactile feedback allowed subjects to perform better, and with more consistency, compared to when no vibrotactile feedback was available.

The results obtained in this study is supported by Wickens's multiple resource model (Wickens, 2002). Increasing the number of sensory resources improves human task performance. In this case, vibrotactile haptic feedback and audible feedback supplement the deficient force feedback in the minimally invasive surgical task, thus the improvement in performance. In this study, the visual sensory mode was not utilized. In actual laparoscopic procedures, the visual mode is typically used to deduce force information based on tissue deformation, except in cases when anatomy occludes one's view of the point of interest. It would be advisable, in a future study, to determine if vibrotactile feedback can improve one's task performance in the case when visual feedback is also available.

Other applications of this technology include robotic surgery, in which surgeons receive no force feedback, and surgical training systems. Future studies could investigate the effect of vibrotactile force information on residents in training. For instance, residents may be able to learn how a dangerous amount of force applied to tissue *looks* and *feels* by correlating it with the large vibrotactile feedback they experience. When the vibrotactile feedback is taken away, would they be able to control their force application below a prescribed force magnitude?

Furthermore, studies could be conducted to validate the results found here. For instance, a differentiation study may be conducted of materials in parallel rather than in series, to eliminate the factors which may have contributed to a blurred transition from soft to hard. Also, studies with simulated tasks, specifically designed to test force application control should be conducted to validate the promising results observed here. Last, it would be interesting to see if varying the vibration pulse frequency with the force at the surgical tool tip, rather than the vibration intensity, would produce different results.

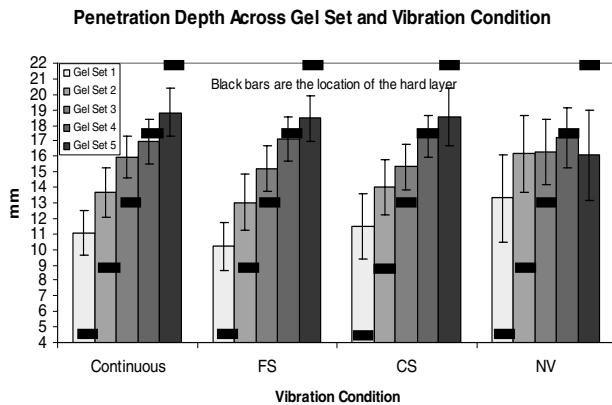


Figure 3: Penetration depth across gel set and vibration condition. Solid dark bars represent the location of the hard gel layer

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