Design and Perceptibility of a Wearable Haptic Device Using Low-Frequency Stimulations on the Forearm

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ABSTRACT

Upper-limb prosthetics lack the afferent feedback necessary for an amputee to know where the prosthesis is in relation to the rest of the body. To address this problem, haptic feedback devices have been explored. In this work, a prototype device, consisting of five modules, has been developed which has the capability to create touch stimulations on the forearm. Each module of the device is lightweight and compact (7.6 g, 25 x 25 x 11.5 mm) consisting of a wire coil spooled on a non-ferrous core and a neodymium magnet housed in a plastic component which can connect to the other components with Velcro. When a coil is energized, the magnet pushes against the skin creating tactile stimulation normal to the skin. In an experiment with five participants, 20 trials of individual stimulations and 20 trials of pattern stimulations were performed on each participant. Participants reported 86% accuracy in correctly locating a single stimulus and 97% accuracy in distinguishing between the four stimulation patterns.

Keywords. Haptic feedback, sensory substitution, myoelectric prosthetics.

1 INTRODUCTION

One out of every 200 people in the United States has an amputation and from 1988-1996 38.5% of those were upper-limb amputations [13]. For a myoelectric prosthetic user, efferent signals on the residual limb can be used to control the prosthesis; however, the prosthetic device does not compensate for the loss of afferent signals from the amputation. While the advent of myoelectric prosthetics have improved amputees lives [15], the high visual demand necessary to operate them has led to amputees indicating in several surveys the need for sensory feedback [15, 2, 9]. Without sensory feedback, amputees must rely on visual feedback for precise control of the prosthesis, which can be time consuming [16]. As a result, haptic feedback devices have been explored to provide this missing sensory feedback.

The need for sensory feedback in prosthetic devices was identified as early as the 1970s in Shannons work [11, 12]. He identified the main sources for creating stimulation as chemical, thermal, electrical and mechanical stimulations [12]. Thermal stimulation has been shown to have slow response and poor spatial resolution [7] making it unsuitable for sensory substitution of a prosthesis. Electrotactile stimulation has proven successful in some studies using haptic feedback for prosthetics [10], but it has a poor psychological viewing as compared with mechanical stimulation [8]. As a result, mechanical stimulation, which can be created through skin stretch, vibration, and normal (i.e. perpendicular to the skin) stimulations, has been largely explored in the literature.

This paper looks to restore tactile and proprioceptive sensations using low-frequency tactile stimulations normal to the skin. The objective was to create a haptic feedback device with the capability to relay position feedback of two hand rotations and individual finger movements to a hand amputee. The challenge of creating such a device is in generating sufficient force for stimulation while also remaining in the space constraints of the forearm. This goal is accomplished using a cross pattern of stimulators so that two-dimensions of movement can be conveyed to the user through pattern stimulations and finger movements through individual stimulations. The contribution of this paper is the finding that low-frequency touch stimulations can be used to achieve high perceptibility of point and pattern stimulations in a five cross pattern tactor array.

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2 BACKGROUND

2.1 Vibrotactile Stimulation

Vibrotactile stimulation, the use of a vibratory stimulator to stretch the skin in the shear direction, has become ubiquitous in the literature as a haptic device. Devices that utilize vibrotactile stimulation have found applications ranging from feedback for prosthetics [8] to use in the operating room [7]. Vibrotactile stimulation is often created through the use of vibrotactors. Although vibrotactors are popular and can generate vibrations with low power, they do have some negative effects. One is the audio noise which comes from them which can be up to 50 db. Prolonged vibrotactile stimulation can also lead to desensitization and users have often reported the vibration stimulus as ”annoying” if used extensively [3]. Another is when attempting to make a device that uses multiple tactors, the surface waves that travel from them due to the shear vibration can have 10% of its original amplitude left after traveling 6 cm on the surface of the skin [5]. As a result of this, many studies choose to use the C2 tactor. However, the use of the C2 tactor comes at the expense of increased size and weight. Skin stretch devices that do not use vibration have also been explored and do appear promising [16]. However, based on their size relative to the forearm they are too big to have multiple wearable devices on the forearm.

2.2 Normal Stimulation

Normal (perpendicular to the skin) stimulation is a form of stimulation that acts perpendicular to the skin. In the literature this type of stimulation has been created through voice coil devices [14], servo motors [1], and other pushing mechanisms [6]. Although, for voice coil devices the authors did not find any display worn on the forearm, many have been used for stimulation on the fingers [14]. One notable forearm haptic display for prosthetics using normal stimulation was created by Antfolk *et al*. The device consisted of five actuators driven by servo motors, and when a servo motor was operated, it depressed a plastic button into the skin for normal stimulation on the ventral side of the forearm. The actuators were placed in the orientation of an open palm to create a more natural feel for the user. In a study with two participants, one participant reported 48% accuracy from 25 trails and the second 81% accuracy from 55 trials of determining stimulation site from the five servo motors [1].

Devices using normal stimulation are relatively unexplored in the literature compared to skin stretch and vibrotactile devices. This may be a result of Shannon and Biggs' *et al*. work which declared shear to be more effective than normal stimulation [11, 4]. Shannon qualitatively assessed that a scratching stimulation evokes more sensation than a patting one [11]. On the other hand Biggs'

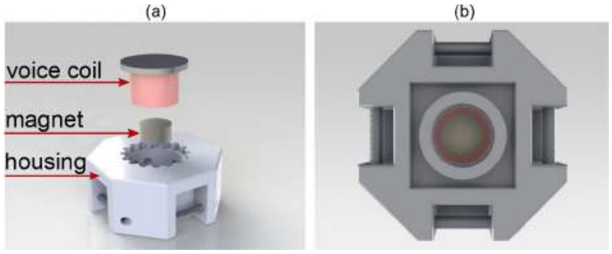

Figure 1: Design concept of wearable voice coil assembly. (a) Exploded view and (b) bottom view.

study was to determine if shear or normal displacement of the skin would be more effective for producing tactile sensations. The study showed for stimulations on the forearm that more displacement and force were required for normal displacements to have the same feeling of intensity as tangential ones [4]. While Shannon and Biggs' claims are sound, they were not entirely conclusive and do not completely address all the constraints of the amputee-prosthesis-feedback system.

3 Normal Stimulation Prototype Device

The objective of this work was to create a haptic feedback device with the capability to relay position feedback of two hand rotations and individual finger movements to a hand amputee. To accomplish this a wearable haptic device worn on the forearm was developed with five stimulation sites. With five stimulators, haptic feedback from the device could be used for an amputee to represent movement of the digits if stimulated individually or gross hand rotations through stimulation patterns. In the literature when using multiple tactors is required, vibrotactors are most typically employed. Although the vibration they create can be noisy and if the tactors are placed too close together they can become difficult to differentiate [5]. Therefore, a device using normal stimulation was created to be compact for multiple sites to be used effectively. Using normal stimulation, the device would be less noisy, not generate surface waves like vibrotactors, and perhaps more comfortable [3]. To achieve normal stimulation, it was decided to use a voice coil design, as seen in some other normal stimulation haptic displays [14].

Although normal stimulation devices have been used in haptic devices before, to the authors' knowledge few to none have been used on the forearm for providing haptic feedback. The concept for the device was to house the voice coil in a rigid shell so that when energized it would repel a magnet normal to the skin's surface (Fig. 1). Using the magnet as the repelling element, a non-ferrous core would need to be used. With this setup, force of the stimulation could be controlled by the voice coil current and stimulation frequency could range from very small to large frequencies unlike vibrotactors which only operate at large frequencies.

In the design of the device, an optimization between generated force and size of the device was required. Since force required from a normal stimulation on the forearm was not readily available in the literature, the authors experimentally determined that a force on the order of 1 N using the magnetic repulsion should be sufficient for effective normal stimulation. Regarding size of the device, since the end goal was for the device to be used with hand amputees, the constraint was to be able to fit five of the devices on the forearm in a cross-pattern. Other important factors to consider were minimizing weight to not add excessive load to the user's forearm and to minimize power consumption. The design criteria for the normal stimulation devices can be seen in Table 1.

The limiting factor for the design was the voice coil and magnet sizes. To keep the device according to the objective size, the

Table 1: Prototype device design objectives and actual parameters

	Objectives	Prototype Specs
Size	< 25 x 25 x 25 mm	25 x 25 x 11.5 mm
Peak Force	1 N	0.44 N
Weight	< 20 g	7.6 g
Coil Resistance	< 10 Ω	4 Ω

maximum magnet diameter was chosen to be a 6.35 mm x 3.18 mm magnet (K&J Magnetics, Product #: D42-N52, 5233 Gauss surface strength). To determine the optimal voice coil outer radius (inner radius should be as small as possible) and length, a relation for a cylindrical magnets' magnetic strength was used:

$$B = \frac{B_r}{2}\frac{d+l}{\sqrt{(d+l)^2+r^2}} - \frac{d}{\sqrt{d^2+r^2}} \qquad (1)$$

where B_r is the magnet's residual induction, r is the radius of the magnet, l is the thickness of the magnet, and B is the magnetic strength of the magnet at an axial distance d from the magnet. Knowing the surface strength of the magnet, the residual induction can be determined allowing for a plot of magnetic strength versus axial distance (Fig. 2). Using this information, a voice coil design was arrived at that optimized its dimensions for producible force, while keeping the resistance low. As can be seen in the figure, around 7 mm the magnetic strength has reached 90% of its cumulative sum. Increasing the voice coil beyond this would result in a small return in force produced for the extra power required, due to the extra wire resistance. As a result, the dimensions for the voice coil were chosen to be 9.5 mm in diameter, limited by space on forearm, and 6.35 mm in thickness, limited by Fig. 2.

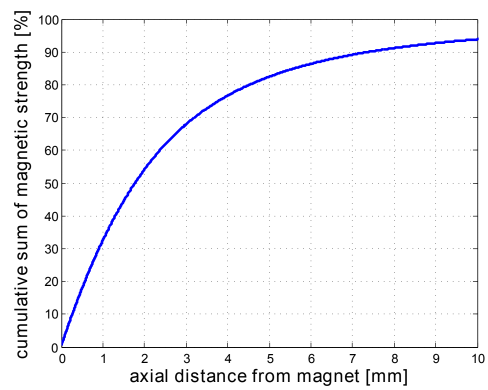

Figure 2: Plot used for determining voice coil length.

To verify that these voice coil dimensions could produce a perceivable stimulation, the equation for the force between a coil and magnet could be used. By assuming that (1) can also be used for the radial component of magnetic strength, an estimate of producible force could be obtained. The force could be determined from the equation for the force produced between a coil and magnet:

$$F = 2\pi I \sum_{i=1}^{L}\sum_{j=1}^{R} r_{i,j}B_{i,j} \qquad (2)$$

where R is the number of radial coils, L is the number of axial coils, I is the current through the voice coil, B is the magnetic strength, and r is the distance from the coil. From this, the theoretical measured force was given by:

$$F_{th} = 0.239I \quad (3)$$

where F_{th} is the theoretical force (N) and I is the current (A). Thus, it was estimated that at 1.5 A through the wire, the produced force of the magnet against the skin would be 0.36 N which is on the order of 1 N. This relation was experimentally verified through measuring the current at which the magnet could no longer hold an object of known weight. The experimental relation between force and current through the voice coil was found to be:

$$F_{exp} = 0.292I \quad (4)$$

where F_{exp} is the experimental force (N) ($R^2 = 0.9988$). As a result, the voice coil parameters were suitable for the desired task since the peak force (determined by allowable current through 32 AWG used in the coil) was 0.44 N at 1.5 A.

The fabricated normal stimulation tactors are shown in Fig. 3. The voice coil based design uses a magnet to repulse against the skin and was used for its compactness and light weight as opposed to using motors [1]. A 3D printed ABS plastic shell houses the voice coils and modules are connected to each other with Velcro. The final specifications for the modules are 25 x 25 x 11.5 mm, 7.6 g, 4.3 Ω, and can produce a peak force of 0.44 N at 1.5 A (Table 1).

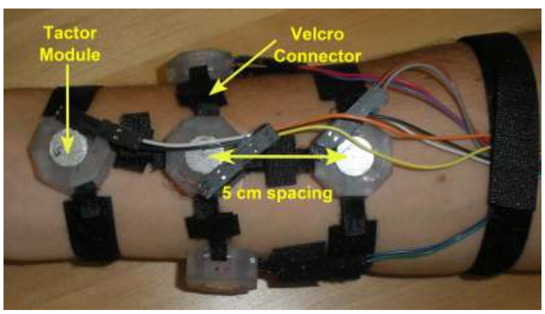

Figure 3: Prototype five tactor array donned on the forearm.

4 Pilot Study Experimental Setup

A study was conducted at the University of Massachusetts-Amherst to evaluate the effectiveness of five normal stimulation modules spaced 5 mm apart in a cross pattern on the ventral side of the forearm (Fig. 3). MATLAB Real-Time Workshop was utilized to control each module which were current controlled through servo amplifiers. Two National Instrument PCI-6229 data acquisition cards were used for communicating with the servo amplifiers and for data collection. Approval to experiment with the haptic feedback device was given by the University of Massachusetts-Amherst Institutional Review Board.

An experiment with five healthy participants, 3 male and 2 female (μ=25 yr, σ=3.9 yr), was performed. The duration of the experiment lasted 30 minutes. Participants first donned the haptic feedback device, and then lay their arm on the table with the palm facing up as shown in Fig. 3. During the experiment, participants were asked to keep their arm flat and not move it; however, they were allowed to move their arm around between trials. During trials, participants view of the device was blocked to ensure that they would rely on the haptic sensations alone for locating the stimulations.

The first test was for participants to differentiate between individual stimulations from the voice coil actuators. A five-minute training period was given to familiarize participants with the stimulations, but not make them fully accustomed to them. The five stimulation locations (top, bottom, middle, left, and right) can be seen in Fig. 4. The stimulations consisted of repeating a pulse with a 1 s period and 10% duty cycle three times. A total of twenty simulations with varying stimulation location and amplitude (0.15 N, 0.29 N, and 0.44 N) were performed. Participants were asked to indicate which actuator had caused the stimulation and were allowed to respond any time after the first pulse occurred, but had to respond after the three pulses.

The second part of the experiment was to test participants with four patterns (Fig. 4). A training period of 5 minutes was given for each participant so they would feel comfortable with the patterns. Twenty trials of varying the pattern while also varying stimulation force between 0.15 N, 0.29 N, and 0.44 N were tested on each participant. During experimentation, the patterns were a 1 s phase delay between stimulations with 10% duty cycle for each stimulation and a 5 s overall period repeated up to 3 times. In both experiments participants were allowed to report the perceived site anytime after the first stimulation and they were not given feedback on their results between trials.

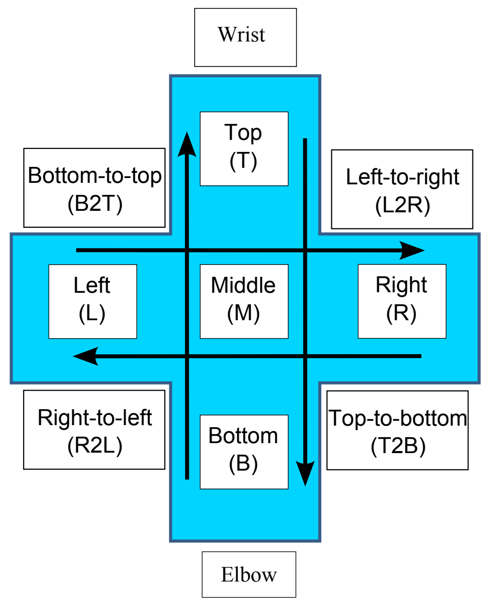

Figure 4: Individual stimulation sites and stimulation patterns.

5 Pilot Study Results

Participants reported 86% accuracy (σ=9.6) in locating individual stimulation sites and 97% accuracy (σ=2.7) in differentiating between the four stimulation patterns. Table 2 shows a confusion matrix for the individual stimulation trials. When participants reported incorrect responses for bottom, middle or top, 70% of the time their response was either bottom, middle or top. The confusion matrix for pattern stimulations can be seen in Table 3. One trial of individual (0.15 N left) and pattern (0.15 N L2R) resulted in the participant indicating that no stimulation was felt.

6 Discussion

Participants reported 86% correct responses in differentiating between the five individual stimulation sites, which was greater than the recognition between stimulation from five servomotors on the forearm by Antfolk *et al*. [1]. The recognition of patterns was 97%, indicating that patterns were easier to differentiate than individual sites. Individual stimulations are more difficult to differentiate since they rely solely on the user's ability to recall where the

Table 2: Confusion matrix for individual stimulations

		Stimulated Site				
		Top	Middle	Bottom	Right	Left
Perceived Site	Top	17	2	0	1	0
	Middle	2	19	1	0	0
	Bottom	0	2	12	0	2
	Right	0	0	2	15	0
	Left	1	0	0	0	23
	None	0	0	0	0	1
	Σ	20	23	15	16	26

stimulation should be in relation to the users' arm. Pattern stimulations are easier to differentiate since the pattern can be determined without need of a fixed reference. When a stimulation occurs, the user only needs to know where the next stimulations are in relation to the first to correctly identify a pattern.

The most difficult part of the individual stimulations was differentiating between the three actuators located down the forearm. From Table 2, it can be seen that most of the errors for individual stimulation occurred from bottom, middle or top stimulation sites. Seventy-one percent (10 trials) of the incorrectly located individual stimulations were from those stimulations and 70% of the time the user was able to determine it was one of those sites. When activated, top, middle, and bottom can appear to be on the same spot whereas left and right are easier to differentiate due to their being on the edge of the ventral side of the forearm. While participants reported difficulty in differentiating between the stimulations down the forearm, they only had a five minute training period. If more time was spent using the device, the ability to differentiate them might increase and should be tested in future experiments.

No significant correlation between force level and accuracy were found in the experiment. Errors in reporting stimulations occurred almost equally for all force levels. To obtain clearer results more trials would be necessary with more participants. If the correct conclusion is that any of the forces used yield similar results, than power could be saved by using 0.15 N stimulation forces. Although it is worth noting that of the 200 trials, two instances occurred of a participant indicating they could not feel the stimulation and they were both with 0.15 N stimulation forces.

7 Conclusion

A prototype haptic feedback device that is modular, lightweight, and small has been created and presented. The device operates through the use of normal stimulation, which is a relatively unexplored means of stimulation in the haptics field on the forearm, created through repulsion of a magnet by a voice coil. This artificial stimulation could benefit the user by serving as a sensory translation mapping between mechanical stimulations and prosthetic movements. A pilot study, where five devices were used in a cross pattern on the ventral side of the forearm, illustrated the ability of the device to relay low-frequency individual and pattern stimulations to the user. The voice coils create a more subtle push stimulation on the skin and create little noise compared to vibrotactors. On the other hand vibrotactors create more urgent stimulations and suffer from desensitization as a result of using high frequency vibrations for stimulation. A drawback of the voice coils is their larger power consumption. Future experiments using surface electromyography (sEMG) signals from the participant will be used to examine the performance impact with closed-loop feedback. With the new study, the differences in the devices ability for use as a haptic feedback device for an upper limb amputee should be made clear. Future studies should test the device with amputees to determine its effectiveness for sensory feedback.

Table 3: Confusion matrix for pattern stimulations

		Pattern Stimulation			
		T2B	B2T	R2L	L2R
Perceived Pattern	T2B	20	0	0	0
	B2T	0	20	2	0
	R2L	0	0	23	0
	L2R	0	0	0	34
	None	0	0	0	1
	Σ	20	20	25	35

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