Communicating through touch for assistive technology: Macro fiber composites for vibrotactile stimulation on the abdomen

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Abstract. The sense of touch is a primary communication mode in assistive technologies for those with sight loss. Research in real-time sensory substitution systems is showing promise, as alternative senses are utilized to afford object recognition or spatial understanding, especially vibrotactile stimulation on body areas that are considered redundant and unobtrusive. Here, a novel belt, integrating a matrix of macro fiber composites, is purposed to deliver vibration stimuli to the abdomen. The design and development of the belt is presented and a systematic experimental study is conducted to analyze the impact of frequency and duty cycle. The belt is a beta precursor to a soft haptic feedback device that will enable situational awareness and obstacle avoidance through the localization of vibrotactile stimulation relative to a body-centric frame of reference.

1. Introduction

Globally, visual impairment is approaching 300 million, inclusive of those with blindness and low vision [1]. In an effort to aid those with sight loss and preserve mobility, a myriad of electronic travel aids (ETAs) have been devised and built [2-6]. These devices substitute or capitalize on intact sensory modalities for alternative communication interfaces. They provide an input based on environmental understanding and subsequently re-display an impression of pertinent objects or hazards, assisting in spatial negotiation or travel. The sensory substituted output is distilled and tailored for the other senses, such as touch and/or hearing, and represents a critical feature of the human-machine interface [7-14].

For example, the BrainPort [15, 16] utilizes stimulation on the tongue to indicate a pattern obtained through computer vision from a digital camera. Similarly, 'The vOICe' [17] and other real-time auditory sensory substitution systems [18] have been proposed, converting data from a digital camera to an auditory output that can be interpreted by the user. While these solutions may grant user's a restricted understanding of their spatial context, their practicality is finite, particularly in more complex and dynamic, ecologically-valid paradigms [19]. Deficiencies stem from both limited inputs (environmental sensing) and outputs (hazard communication) that leave vulnerabilities in real-world conditions.

To facilitate the transduction of an ETAs' output, human-machine interfaces should mitigate deterring factors of the potential wearable, including minimizing obtrusiveness, maximizing the use of redundant sensory transduction, and benefiting from favorable psychophysical properties. Vibrotactile stimulation has been suggested to address all these issues, with a plethora of body locations as potential interaction points for the interface [3], such as the abdomen [20-23] and arms [24].

Tactile stimuli can be displayed as either signals or patterns, and can be modulated in pulse frequency, amplitude or duration [23], making tactor-based vibrotactile communication a robust output for assistive technology. Based on the technical literature, the torso represents the most suitable body location to provide humans with vibrotactile communication [23, 25-27]. Localization of vibrotactile stimulation at the abdomen, for pneumatic- and vibration motor-based tactors, has been demonstrated to be highest near the navel and spine [20]. Further, the abdomen was indicated to be more sensitive for higher stimulus frequencies [20], and timing parameters related to localization performance have been investigated [28, 29].

Macro-fiber composites (MFCs) are a class of piezoceramic material that integrate structural compliance, high actuation authority, and controlled anisotropy [30]. MFCs were originally designed by researchers at NASA Langley Research Center over fifteen years ago [31, 32]. Since their inception, they have found translational applications across a number of engineering and scientific domains [33, 34]. For example, MFCs have been leveraged in morphing-wing and flapping-wing structures [35, 36], in underwater propulsion systems [37], and energy harvesting devices for humans, animals, and robots [38-40]. Here, we propose the integration of MFCs in a new vibrotactile device, consisting of a specialized belt featuring an array of MFCs for environmental reconstruction.

In comparison to traditional tactors (linear resonant actuators (LRAs), pneumatics, or dc motors) used for vibrotactile stimulation, MFCs offers the advantages of an actuator alternative with smaller size and lighter weight, ease of conforming to various geometric shapes, and a wide operational frequency range. Specifically, MFCs have an area density of 0.16 g/cm^2 and may be fabricated to various shapes as long as the electrodes can placed in an interdigitated pattern [41]. On the other hand, pneumatics and dc motors are often limited to available off-the-shelf sizes, which may hinder the integration in personalized, haptic devices. Beyond their nearly free-form fabrication, MFCs have a substantially wide operational frequency range from 0 to 10 kHz [41]. In contrast, LRAs are only operational within narrow high-frequency bands. Building on the compelling literature on MFC applications, we seek to explore their use as an alternative to existing tactors.

The specific objective of this proof-of-concept study is to psychophysically examine the MFC-based prototype towards the future integration of this novel haptic interface in a smart wearable for situational awareness in the setting of low vision. Specifically, we examine the localization sensitivity of vibrotactile stimulation on the abdominal region for various combinations of stimulation frequency and duty cycle. Experiments are conducted on a group of 48 subjects, toward a first quantification of the sensitivity of vibrotactile stimulation, which will inform an iterative design cycle, where the number and density of MFCs will be systematically varied.

The rest of the paper is organized as follows. In Section 2, the hardware design and communication configuration of the belt is presented. In Section 3, the experimental scheme for the human subjects testing and the statistical data analysis are detailed. The results of statistical analysis from the pilot study are presented in Section 4, while a discussion of the result is offered in Section 5. Conclusions and future research directions are summarized in Section 6.

2. Belt Design

2.1. Hardware

The belt utilizes four 38 mm × 20 mm M2814-P1 (Smart Material Corp., Sarasota, FL, USA) MFC actuators driven by two AMT2012-CE3 amplifiers (Smart Material Corp., Sarasota, FL, USA); two actuators are driven by each amplifier. The MFCs are arranged at the corners of a rectangular 85 mm × 140 mm region. Each actuator is mounted on a three-dimensional (3D) printed casing fabricated on the Makerbot Replicator (MakerBot Industries, LLC, Brooklyn, NY, USA) to form a tactor unit. The MFC is fixed on both sides and slightly bent to fit within the 37 mm length of the 3D printed casing.

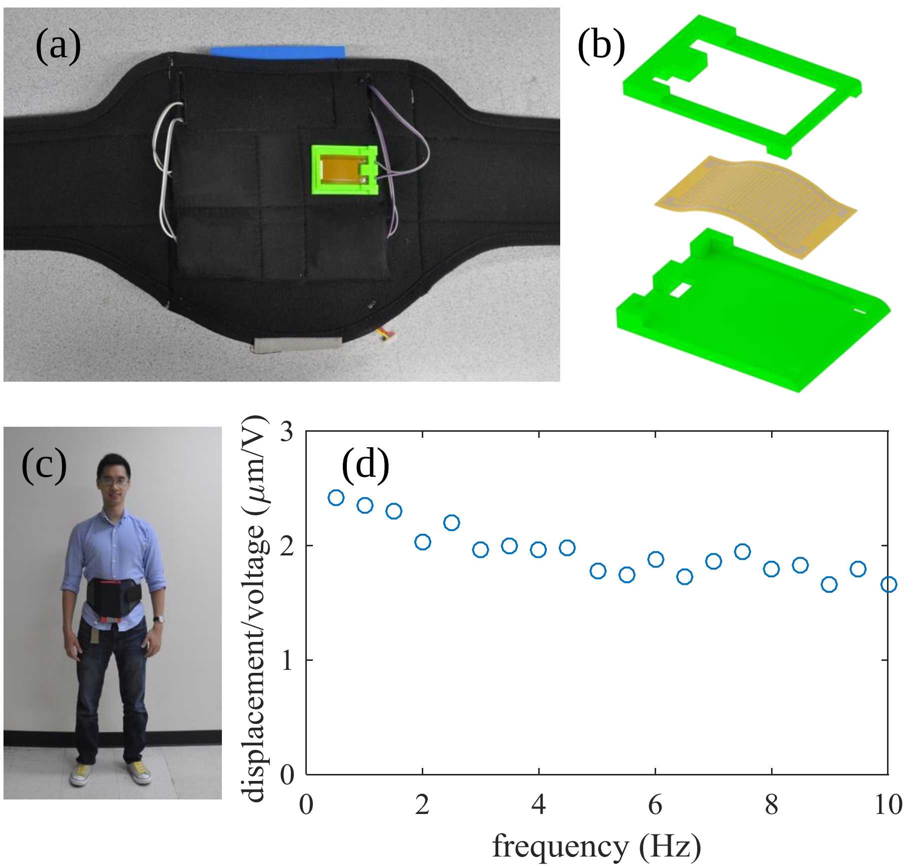

Figure 1. (a) Arrangement of the MFC tactors (one of the tactors and its 3D printed casing is shown explanted from the woven pocket); (b) computer-aided design illustration of exploded view of MFC and 3D printed casing; (c) picture of the belt as worn on the abdomen and centered on the umbilicus (the first author Paul Phamduy wears the belt); and (d) response of the MFC tactor, computed as the ratio between the peak-to-peak displacement at the midspan and the peak-to-peak voltage input to the power supply.

Vibratory stimulation is created by providing an alternating square wave with a peak-to-peak voltage of 5.0 V and a DC offset of 2.5 V to the amplifier, which translates the signal to an alternating square wave with a peak-to-peak voltage of 2.0 kV and a DC offset of 0.5 kV across the electrodes of the MFC tactor. The applied square wave voltage elicits bending deformation of the MFCs, due to MFC prebending that causes the piezoelectrically-induced axial stress to elicit bending.

The response of the MFC tactor was characterized using a Labview (National Instruments, Austin, TX, USA) program was utilized along with a LB-70 laser displacement sensor (Keyence Corp., Itasca, IL, USA) to capture the supplied input voltage and the output deflection at the MFC midspan. The amplitude of deflection of the MFC was approximately constant, between 0.009 and 0.013 mm peak-to-peak vibrations, for frequencies between 1 and 10 Hz.

The tactors are held in place with woven pockets that were sewn into a commercially available lumbar back support belt, see figure 1. During positioning on the end user, the expanded support brace of the belt is rotated 180 degrees and centered on the umbilicus.

The belt is powered by an 11.1 V 1400 mAh three-celled lithium-polymer battery (Traxxas, McKinney, TX, USA). A 5 V step-down voltage regulator is utilized to distribute the power to the amplifiers and microcontroller. On a single charge of the battery, the belt can operate for approximately 20 minutes.

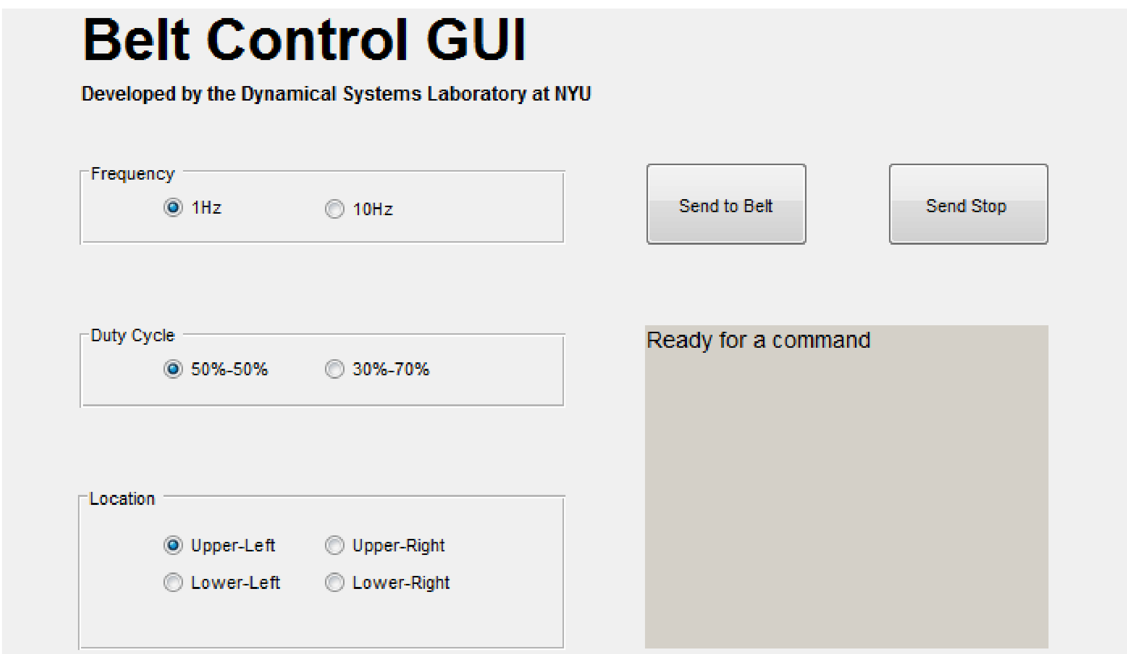

Figure 2. Graphical user interface (GUI) to control the frequency, duty cycle, and location of the vibration of the belt. The radio buttons can be utilized to set frequency, duty cycle, and location of the vibration, while the buttons on the top right can be utilized to start or end the stimulation. Vibration parameters currently in use are displayed on the textbox.

2.2. Communication and Control

The signals sent to the tactors are controlled by a Moteino microcontroller (LowPowerLab LLC, Canton, MI, USA) through an array of TXB0104 voltage level shifters. The microcontroller can independently control the frequency and duty cycle of each tactor. Similar to the communication network in [42], the radio transceiver on the microcontroller receives a command through a base station utilizing a user datagram protocol (UDP). A custom graphical user interface was created using MATLAB (MathWorks, Natick, MA, USA); commands entered into the GUI were communicated over the UDP to manually control the frequency and duty cycle and to switch between the four tactor locations, see figure 2.

3. Experimental Scheme

3.1. Human Subjects Testing

The data collection for this study follows the regulations of the University Committee on Activities Involving Human Subjects of New York University (IRB-FY2016-57). All participants were either undergraduate or graduate students recruited from New York University Tandon School of Engineering and were tested separately for each trial.

To explore the feasibility of the MFC tactors as a stimulus generator, we collected data on the psychophysical performance of vibrotactile stimulation under various combinations of duty cycle and frequency. Specifically, we scored if the subject was able to discriminate the location of the stimulus, by correctly identifying its location relative to the prescribed stimulation parameters. The tested frequencies were 1 and 10 Hz and duty cycles were 50% and 30%.

A total of 48 individuals were recruited in the experiment and each trial was conducted by: i) collecting the individual's age and sex; ii) instructing the individual on how to wear the belt on the abdomen; iii) randomly turning on only one of the four tactors while the individual is wearing the belt; and iv) querying the individual on the location of the stimuli and collecting the following scripted responses: 'top left', 'top right', 'bottom left', 'bottom right', 'I don't know'. If the location of the stimulus matched the individual's response of the perceived location, the query was scored as correct. A mismatch of perceived and actual stimulation location was scored as incorrect. 'I don't know' responses were treated as lapses, and these trials were discarded for analysis. This is repeated for an individual until all four tactors were tested at least once. That is, the query of the location of the stimuli was repeated four times per individual. Although not analyzed, lapses are also presented in what follows for completeness.

As there were two frequencies, two duty cycles, and four tactor locations for testing, there were four possible combinations of frequency and duty cycle pairs and 24 permutations for the order in which the tactors were tested per individual. Six permutations were then randomly assigned to each of the four pairs of frequency and duty cycle. A full permutation of frequency, duty cycle, and testing order would require 96 individuals. However, as only frequency and duty cycle were examined for vibrotactile localization, their pairs were randomized onto the 24 permutations for the testing order to reduce the number of individuals needed for the study. The set of 24 permutations was tested on two groups of 24 individuals each, with one permutation per individual. Thus, the permutations were completed twice to account for the 48 individuals in the experiment and there were 12 individuals assigned to each frequency and duty cycle pair. Of note, there were no female participants assigned for the condition in which the frequency of stimulation was 10 Hz at a 50% duty cycle secondary to the randomized assignment of participants to conditions.

3.2. Data Analysis

The discrimination performance of the MFCs was examined for the entire group and then for subgroups to explore potential effects of age, gender, and combinations of frequency and duty cycle. Specifically, the discrimination performance for the subgroups were examined between: i) males and females, ii) a younger and older half of the population, and iii) variations of frequencies and duty cycles.

Discrimination performance was modeled as a binomial response variable with unknown rate (θ) of correct responding. Rates were estimated using a Bayesian estimation procedure with an uninformed Jeffreys prior [43]. The shortest interval containing 100 (1-α)% of the probability mass in the distribution over the unknown rate (θ) provide a Bayesian analog of the traditional significance test. For example, a confidence interval that does not contain the 'blind guessing' rate of θ = 0.25 provides evidence for a significant level of discrimination in that experimental condition. Further, lack of overlap between the confidence intervals derived from the posterior rate estimates of two experimental conditions shows a significant difference in the discrimination rates induced by two experimental conditions.

All analyses were performed using MATLAB with a significance threshold of $\alpha = 0.01$. To examine potential bias in vibrotactile localization of the four quadrants, a chi-square test was performed with the null hypothesis that there is no difference in the discrimination performance based on the quadrant.

4. Results

All undergraduate and graduate students participating in the study were from New York University Tandon School of Engineering. In total, 48 subjects participated in the study (13 female), ranging from age 18 to 37 (one participant did not indicate age) with a median age of 23. Age 24 was used as the cutoff to explore the potential effect of age on vibrotactile stimulation, by partitioning the population into two subgroups, older versus younger participants.

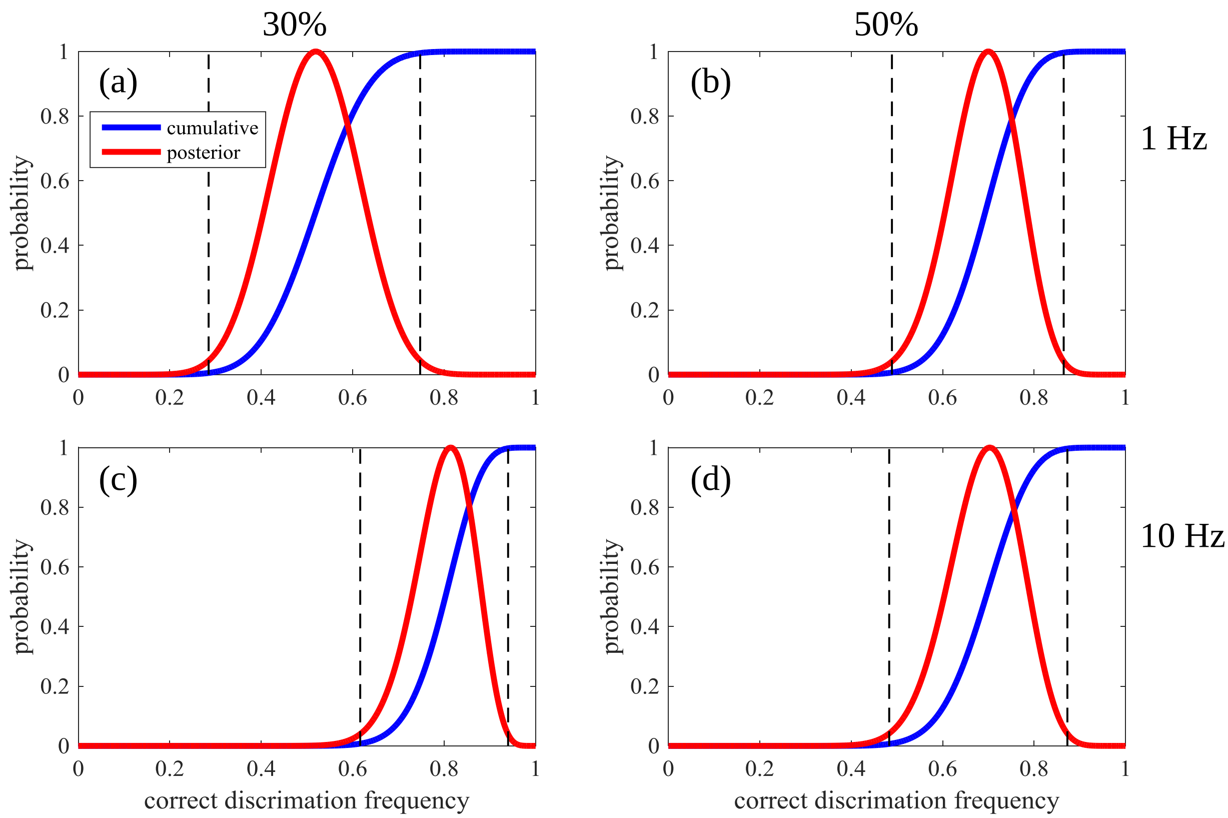

Figure 3. Probability density and cumulative probability functions showing performance in each condition. Solid red curves are posterior probabilities for discrimination frequency, blue lines are the corresponding cumulative probability functions, and vertical dashed black lines show the shortest 99% posterior confidence interval. This interval does not encompass the ‘blind guessing’ rate of 0.25 in any of the four conditions. The four frequency and duty cycle pairs examined are (a) 1 Hz, 30%, (b) 1 Hz, 50%, (c) 10 Hz, 30%, and (d) 10 Hz, 50%.

Table 1. Discrimination performance and confidence interval for the entire populations and subgroups formed to study the effects of gender, age, and across frequency and duty cycles.

conditions	1 Hz, 30%	1 Hz, 50%	10 Hz, 30%	10 Hz, 50%
overall (range)	0.52 (0.29 – 0.75)	0.70 (0.49 – 0.86)	0.81 (0.62 – 0.94)	0.70 (0.48 – 0.87)
female (range)	0.68 (0.32 – 0.93)	0.77 (0.41 – 0.97)	0.86 (0.51 – 0.99)	--- (---)
male (range)	0.39 (0.13 – 0.71)	0.67 (0.42 – 0.87)	0.80 (0.55 – 0.95)	0.70 (0.48 – 0.87)
age<24 (range)	0.45 (0.14 – 0.80)	0.80 (0.54 – 0.95)	0.76 (0.51 – 0.93)	0.70 (0.39 – 0.92)
age>=24 (range)	0.57 (0.27 – 0.84)	0.54 (0.22 – 0.84)	0.95 (0.62 – 1.00)	0.72 (0.41 – 0.93)
conditions	**30%**		**50%**	
across frequencies	0.69 (0.53 – 0.82)		0.70 (0.55 – 0.83)	
conditions	**1 Hz**		**10 Hz**	
across duty cycles	0.62 (0.46 – 0.77)		0.76 (0.61 – 0.87)	

Table 2. Lapse counts.

conditions	1 Hz, 30%	1 Hz, 50%	10 Hz, 30%	10 Hz, 50%
overall	21	12	12	15
male	13	8	8	15
female	8	4	4	---
age>=24	16	6	8	7
age<24	5	5	4	8
conditions	**30%**		**50%**	
across frequencies	33		27	
conditions	**1 Hz**		**10 Hz**	
across duty cycles	33		27	

We estimated the frequencies of correct discrimination (θ) of the stimulated quadrant given the number of total trials (N) and successful localizations (S) using a Bayesian measurement procedure. The binomial likelihood, in conjunction with an uninformative Jeffreys prior over possible discrimination frequencies, defines the posterior as

$$p(\theta|S \cdot N) \propto p(\theta)p(S|\theta \cdot N) = [\theta(1-\theta)]^{-0.5}[\theta^S(1-\theta)^{N-S}] \quad (1)$$

This yielded the probabilities shown in figure 3. The plots indicate that while all four experimental conditions produced discrimination performance above chance (discrimination rate of 0.25), the condition in which the frequency of stimulation was 10 Hz at a 30% duty cycle yielded the best overall discrimination rate, of about 0.81.

The discrimination performance and confidence interval for various stratifications of the data are shown in table 1. Even across groups of males, females, and the full range of participant ages, discrimination performance was mostly above chance (0.25). However, some of the 99% posterior confidence interval for subgroup conditions overlapped with a discrimination rate for chance (0.25) , such as for: i) males for whom the frequency of stimulation was 1 Hz at a 30% duty cycle, ii) participants age 24 and older with a 1 Hz stimulation frequency at 50% duty cycle, and iii) participants under age 24 and a 1 Hz stimulation frequency at 30% duty cycle. The highest discrimination performance was achieved by the participant group of age 24 and older for the frequency of stimulation of 10 Hz at a 30% duty cycle.

A chi-square test did not detect significant bias in the tendency of subjects to choose any of the four quadrants over others ($p > 0.01$). The lapses for each condition are listed in table 2.

5. Discussion

In this study, we presented the design of a specialized belt featuring a matrix of MFCs, together with its experimental validation for vibrotactile stimulation on the abdomen. We demonstrated that vibrotactile stimulation delivered by MFCs yields perceptually discriminable localization performance, which is modulated by the tested combinations of frequency and duty cycle.

5.1. Overall Discrimination Rates

Although the highest performance in the overall discrimination reaches approximately 81%, the confidence interval ranges from 29% to 94%, across the four frequencies and duty cycles, indicating high variability in the perception of vibrotactile stimulation for MFCs. This large variability suggests that the stimulation produced by MFCs was differentially perceived in the population, with a fraction of the population showing limited ability to localize the stimulation and a fraction achieving even higher performance than existing technologies [20, 21]. Specifically, the rate of localization for motor and pneumatic-based tactors, such as eccentric rotating mass motors, LRAs, and pneumatic-based tactors, was found to range between 62% and 98% [20, 21].

Performance difference between existing technologies and MFCs may be related to the mechanisms utilized to provide the vibrotactile stimulation, e.g., piezoelectric actuation versus a moving mass or compressed air, and/or the size and density of the tactors used in the design [20, 21]. Further, differences in discrimination may also be ascribed to the spatial arrangement of the tactors, the intensity of vibration, and individual differences in skin sensitivity, which can be confounded by disease, medication, and specific anatomic location(s) tested [20, 24, 44, 45]. Although other tactor types may exhibit superior localization performance on average, the high performance achieved by MFCs for specific subjects, coupled with their potential for implementation in a wearable form-factor [46] offer compelling evidence for future research on the proposed belt.

5.2. Effects of Age on Performance

Although our results did not reveal an overall observable influence of age on the localization of vibrotactile stimulation, one of the two highest differences in vibrotactile discrimination was recorded between the 10 Hz/30% and 1 Hz/50% conditions for participants age 24 and over. This evidence is in line with [24], where younger participants displayed better overall performance in the localization of vibrotactile stimuli (e.g., in those aged 18-33, as compared to 60-85), without attaining statistical significance. Age-related differences in localization performance should expected given the well-demonstrated decrement in absolute vibrotactile sensitivity that occurs with increasing age (e.g., comparing cohorts aged 18-30 and 69-79) [47].

Our experiments involved a population of relatively narrow age range, which may have masked the overall influence of chronological age on localization of stimulation. Another factor which could have mitigated our observations may relate to the demonstrated independence between age and the ability of performing spatial summation or detecting vibration from tactors of varying size [24, 47]. Specifically, one may hypothesize that subjects of different age were differentially sensitive to the vibration, but they similarly performed in the localization due to analogous abilities to process the stimulus.

5.3. Effects of Gender on Performance

Our results did not reveal significant gender differences in participants' localization of vibrotactile stimulation utilizing the belt, although one of the two highest differences in vibrotactile discrimination was found between the 10 Hz/30% and 1 Hz/30% conditions in male participants. A stronger effect of gender could have been expected based on previous work, indicating an effect of gender on the localization of vibrotactile stimuli, when adjusting for body surface area [48], such an adjustment was implemented by accounting for body type (i.e., size, weight, and body fat) and body area tested (i.e., forehead and arm).

Here, we did not take into consideration human physiological factors that might influence localization, including body height and weight [49], waist size [20], body fat percentage or visceral fat [50], humidity [45], temperature [51], hairy/non-hairy (non-glabrous/glabrous) skin [44], and clothing [20, 52]. Perhaps, adjusting our results by utilizing these factors could have strengthened the dependence of localization performance on age. Future studies will seek to test this hypothesis through experiments in which we will record salient physiological factors and explore their correlation with localization performance.

5.4. Effects of Vibrotactile Stimulation Parameters on Performance

Of the four stimulation conditions, the highest rate of overall discrimination was achieved for a frequency of stimulation of 10 Hz and a 30% duty cycle. While no data is available for higher frequencies, this finding suggests that further improvement would be afforded by increasing the frequency of stimulation beyond the narrow range considered in this study. Here, the selected test range between 1 and 10 Hz was a 'pulsed' light touch approach to create a foundation on which to explore higher frequencies that are more typical of vibrotactile stimulation.

Frequencies tested on other vibrotactile systems range from 25-320 Hz at the abdomen [20, 25] and 100-250 Hz on the arm [24]. MFC could operate up 10 kHz, affording a much wider range of exploration for the advancement of vibrotacticle stimulation. Frequency-dependent sensitivity for stimulation of the Pacinian corpuscle, a fast-adapting nerve receptor on the skin, sensitive to high-frequency vibration, is expected to be maximized in the range of 100-1000 Hz [53]. By testing a higher frequency range, the rate of vibrotactile localization and discrimination frequency could thus increase. Alternatively, the Meissner's corpuscles have been suggested as a mechanoreceptor for light touch modalities with vibration in lower frequency ranges, below 50 Hz [44, 53, 54]. Future work will seek to examine the role that various mechanoreceptors might play in the detection of vibrotactile stimulation from MFCs.

Other work has sought to use bursts, pulses, and patterns in their tests [28], whereby interburst and burst durations for various vibrations patterns on an array of tactors were examined for perception of the duration and spatial distribution of the vibration stimuli. Such vibrations patterns have been demonstrated to be easily recognizable and accurately provide navigational cues to mobile participants [25]. Although the bursts considered in the literature are differently defined than pulses, these findings confirm that timing parameters play a role in the perception of vibrotactile stimulation [28, 55]. Future work will seek to clarify this possibility by systematically exploring variations in the duty cycle, beyond the two values selected in this first study, in light of modulating the spatial density of the MFC actuators.

6. Conclusion

In this paper, we presented the design of a novel belt utilizing MFCs along with a feasibility study to administer vibration stimuli to the abdomen. Future work for this specialized belt will seek to improve localization performance of the vibrotactile stimuli by examining participant responses over wider frequency and duty cycle ranges. Further, we will examine other design parameters such as tactors' spacing and number, while exploring multi-element stimulation paradigms.

The incorporation of the belt with a sensing system, such as an array of multi-modal sensors, will be investigated to create assistive technologies that communicate to end-users through haptic-based human machine interfaces [22]. To further enhance the effectiveness of the proposed system, we will explore its integration with other emerging assistive technologies, such as wearable text reading systems [56] that may benefit from novel finger motion adaptive algorithms for braille reading [57], toward increasing spatial and contextual awareness.

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