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Haptic Perception on the Plantar Surface for Vibrotactile Feedback Insoles

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***Abstract*—The plantar surface provides critical sensory feedback during standing, balancing, and locomotion, making it a promising area for haptic augmentation. While insoles offer a practical platform for integrating vibrotactile feedback into daily activities, insole-based haptic systems must be carefully designed to complement rather than interfere with the natural sensory processing. To inform such designs, we investigate spatial acuity and perceivability ratings across the plantar surface under varying load conditions. We use a 3D-printed vibrotactile insole with four linear resonant actuators (LRAs) and conduct a user study (n=6) investigating stimulus localization and perceivability in both sitting and standing postures. We ask the participants to localize the perceived vibrotactile stimuli on a foot contour and to rate the perceivability of the stimuli. Our results reveal significant differences in localization accuracy between anatomical regions, with average localization errors ranging from 8.44 ± 0.1 mm at the great toe to 35.4 ± 7.06 mm in the metatarsal region. In addition, we find that perceivability rating significantly decreased in the standing posture compared to the sitting posture, with an average reduction of 29%. These results provide guidelines for the optimal design and positioning of vibrotactile actuators in haptic insoles interfaces.**

***Index Terms*—Vibrotactile feedback, haptic feedback, haptic perception, foot sole, plantar surface**

I. Introduction

The sole of the human foot serves as an important sensory area, continuously providing critical information about ground conditions, weight distribution, and balance during standing and locomotion [1]. Haptic feedback systems can be valuable communication tools, particularly in situations where visual and auditory channels are occupied [2]. While numerous areas of the body have been explored for haptic interfaces [2], [3], the plantar surface offers unique advantages. It naturally processes complex tactile information during daily activities and remains available when hands are occupied with other tasks. Since most people wear insoles on a daily basis, the integration of vibrotactile feedback devices seems practical [4]. Moreover, the critical role of the foot in balance and locomotion suggests that haptic feedback through this channel could be particularly intuitive and effective for applications ranging from navigation assistance [5], [6], [7] to rehabilitation [4], [8], [9], [10], [11], [12], [13], [14].

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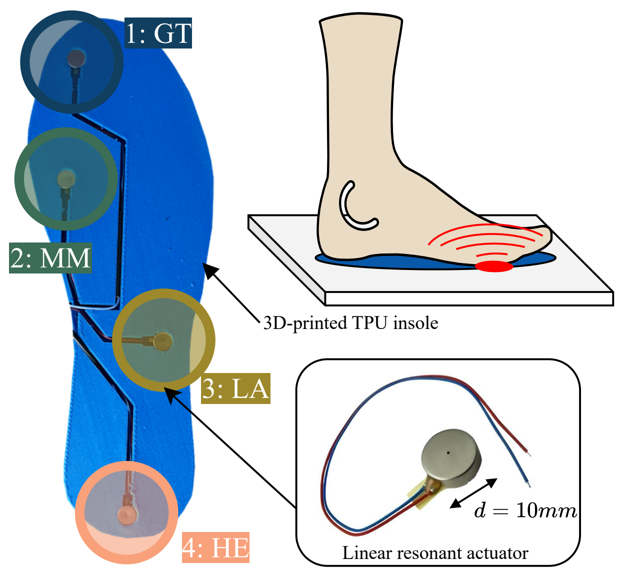


Fig. 1: We investigate vibrotactile perception on the plantar surface with a 3D-printed thermoplastic polyurethane (TPU)-based insole containing four linear resonant actuators (LRAs). Actuators 1 to 4 are placed on four selected landmarks of the plantar surface, namely the great toe (GT), medial metatarsal (MM), lateral arch (LA), and heel (HE). Each actuator has a diameter of 10 mm.

In the rehabilitation context, systems have been developed to provide real-time gait feedback [9] and to assist patients with Parkinson's disease [10]. Moreover, patients with peripheral neuropathy benefit from tactile insoles in gait rehabilitation [14]. Studies have shown that subthreshold vibrotactile stimulation can enhance balance control [8]. White noise foot vibrations can reduce stride, stance and swing time variability in the elderly [12] and vibrotactile cues delivered through an insole were effective in communicating increased fall potential [13]. Vibrotactile insoles have been used to guide squats and deadlifts by providing information about the centre of pressure [15].

Ground texture rendering in virtual reality (VR) has emerged as an application for vibrotactile shoes or insoles in order to enhance immersion [16], [17]. Vibrotactile patterns coupled with motion can create the impression of interacting with virtual objects that exhibit compliance, elasticity, or friction [16]. In addition, foot vibrations can induce immersive virtual walking experiences [18].

Previous studies demonstrated that humans can intuitively interpret vibrotactile feedback across various body location [19], with extensive studies focusing on wrist-based applications [20], [21], [22], [23]. However, wrist-based vibrotactile feedback proved ineffective in the rehabiliation for Patients with Parkinson's disease [24], for such cases insole-based cues may be more appropriate. Early work by Velázquez et al. [25] demonstrates the feasibility of pattern recognition through the foot sole, although their evaluation was limited to seated conditions. More recent comparisons of body positions for tactile feedback suggest that foot-based stimulation can be effective and intuitive for navigational cues [5], [6].

Evidently, vibrotactile feedback on the plantar surface shows promise for sensory augmentation and as communication channel. Effective implementation requires understanding perceptual characteristics such as stimulus localization and perceivability distribution. Earlier work shows that the capability to discriminate vibrotactile stimuli spatially is dependent on the location [26]. Consequently, the actuator placement at locations with high mechanoreceptor density improves stimulus detection [27]. Further, previous research shows that standing increases perception thresholds for vibrotactile stimuli. Yet, it remains unclear how this effect scales for haptic devices and how it influences spatial acuity [28]. Prior work has explored pattern recognition rates and temporal encoding through haptic insoles [29] and application-specific effects for communication and virtual reality rendering [16], [30]. However, with respect to haptic interface design, there are still open questions regarding the usable spatial acuity and perceivability across the plantar surface.

In this work, we investigate vibrotactile perception on the plantar surface using a 3D-printed insole with embedded LRAs. We examine how perceivability varies in different anatomical regions of the foot sole. Additionally, we analyze the spatial acuity across these regions to determine their suitability for conveying spatially-encoded information. In a participant study ($n = 6$), we evaluate stimulus perceivability ratings and the user's ability to accurately localize stimuli across four anatomic regions of the foot sole. Our investigation compares performance between seated and standing postures to further our understanding of how weight bearing alters vibrotactile perception. This work contributes empirical findings on plantar surface sensitivity and practical insights concerning the optimal actuator density and placement for vibrotactile insoles.

II. Methods

A. Vibrotactile Insole System

In order to apply vibrotactile cues at the plantar surface, we design a vibrotactile insole system, which consists of three main components. A 3D-printed flexible insole, four linear resonant actuators (LRA), and custom control electronics. The vibrotactile insole is sized according to a European shoe size 42 and has a thickness of 5 mm. We 3D print the insole with flexible thermoplastic polyurethane (TPU) filament (TPU95A, Shore hardness 95A; colorFabb, Belfeld, Netherlands) using fused filament fabrication (FFF) on a Prusa MK3S printer (Prusa Research, Prague, Czech Republic). We select 20% infill ratio with a gyroid pattern. The insole shape is designed in Fusion360 CAD software (Autodesk, San Rafael, USA). The insole features cutouts for actuator placement designed to fit actuators with 10-mm diameter by press-fit mounting without adhesives. Grooves in the insole accommodate the connecting cables to the actuators (Fig. 1). The final insole weighs 123 g.

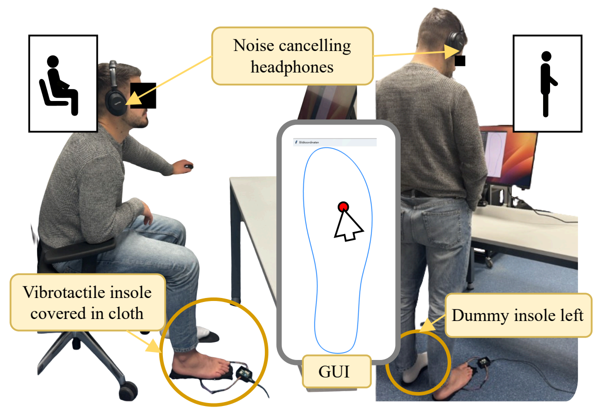


Fig. 2: Setup for testing vibrotactile perception on the plantar surface. Participants placed their right foot on a fabric-wrapped insole with actuators. The left foot rested on a fabric-wrapped dummy insole. Mouse pointer and localization dot of the graphical user interface (GUI) are enlarged for visualization.

The actuator placement on our vibrotactile insole is guided by the distribution of fast-adapting type II (FAII) mechanoreceptors and distribution of receptive fields [31]. The four actuators target specific anatomical locations (Fig. 1). The first actuator sits beneath the great toe (GT), where mechanoreceptor density peaks and receptive fields are smallest. The second actuator rests under medial metatarsal (MM), which bears significant weight during wide stance [32] and is a commonly chosen actuator position in other haptic insole prototypes [17], [29]. The third actuator aligns with the lateral arch (LA), maintaining ground contact in wide stance. The fourth actuator is located beneath the central heel (HE), despite its lower mechanoreceptor density, as it is an anatomical landmark.

Coin-shaped LRAs (VG10366002D, Vybronics, Shenzhen, China) generate the vibrotactile stimuli. Each actuator measures 10 mm in diameter and 3.6 mm in height, with a resonant frequency of 175 Hz. The actuators produce vibrations perpendicular to their housing surface, aligning with the primary direction of ground reaction forces on the plantar surface. The actuators provide rated accelerations of 1.5 g_{RMS} at their rated voltage of 2 V_{RMS}. We use haptic drivers (DRV2605L, Texas Instruments, Dallas, TX, USA) to control the four LRAs. These are mounted on a custom PCB including a microcontroller module (ESP WROOM-32E, Espressif, Shanghai, China).

B. Vibrotactile Perception Experiment

Our experiment evaluates vibrotactile perception on the plantar surface, exploring both stimulus localization and user perceivability ratings.

1) Experimental Procedure and Participants: The experiment consists of two tasks: stimulus localization and perceivability rating, with participants performing both tasks while standing and sitting. The posture order was counterbalanced across participants. Six participants (one female, five male, age 22.33 ± 2.25 years) were recruited from our lab and have provided written informed consent for this study. Only participants with self-reported European shoe size 42 were selected. All except one participant were right-foot dominant. In order to familiarize themselves with the experiment, participants first placed their foot on the insole while seated, received three randomized vibration cues, and then rated them in the GUI.

In both postures, participants placed their bare right foot on the actuated insole and their left foot on a dummy insole (Fig. 2) with same dimensions. Conducting the experiment without shoes avoided confounding effects from the curved in-shoe geometry and allowed for visual inspection to ensure repeatable foot-insole alignment. Both insoles were covered with opaque black socks to conceal actuator positions and prevent tactile discrimination between TPU and metal surface. This also reflects typical application conditions where users mostly wear socks. Participants wore noise-canceling headphones playing pink noise to mask actuator sounds and minimize environmental distractions.

For the standing condition, participants maintained an upright posture with their heels aligned to the rear edge of the insoles. Participants were instructed to maintain a natural stance without weight shifting. In the seated condition, participants sat naturally with their feet positioned identically to the standing condition.

The experimental protocol consisted of 24 total stimuli per participant (12 per posture), with three stimuli from each of the four actuators, presented in randomized order. Each stimulus comprised three 400-ms vibration bursts, separated by 150-ms pauses (total duration: 1.2 s). A pulsed pattern was selected to enhance stimulus detection [33]. The actuators operated at their rated amplitude ($2\,V_{RMS}$). Each stimulus was preceded by a visual cue from the experimenter indicating that a stimulus will follow.

2) Localization Task: During the localization task, the participants marked the perceived vibration locations on a graphical user interface (GUI, Fig. 2) that displayed the outline of the insole. After the participant marked the perceived location, the mark clears to ensure that the participant is not biased by their previous input.

3) Perceivability Rating Task: In the second part of the experiment, participants evaluated the perceivability of the stimuli. For perceivability, participants verbally rated each stimulus on a 5-point Likert scale (Tab. I) ranging from 'not perceptible' (1) to 'very easy to perceive' (5). This scale was chosen to balance differentiation capability with clear distinction between levels, following established practices in previous work [33].

Prior to data collection, participants completed a familiarization trial to understand the procedure and experience the vibrotactile stimulation. The complete experiment lasted approximately 30 minutes per participant, with the localization task preceding the perceivability task for all participants.

TABLE I: Vibration perceivability questionnaire.

Likert scale	'The vibration was ...'
1	not perceptible.
2	faintly perceptible.
3	moderately perceptible.
4	easy to perceive.
5	very easy to perceive.

C. Data Analysis and Statistics

Localization accuracy is quantified using the absolute localization error, calculated as the Euclidean distance between each actuator's geometric center and the participant-selected location. In order to analyze directional influences, we map the localizations by approximating the 2D data as bivariate Gaussian distributions forming ellipsoids across the plantar surface. Their boundaries are defined by the first and second Mahalanobis distances. For perceivability analysis, we evaluate participants' Likert scale ratings. Shapiro-Wilk tests and visual inspection of Q-Q plots [34] reveal that the data does not follow a normal distribution, necessitating non-parametric statistical methods. Due to the small sample size, we report means, standard deviations, and medians to illustrate trends, omitting interquartile ranges for clarity, although the Likert-Scale is not ordinal. Statistical analyses are performed using the *jamovi* software [35] with a significance level of $\alpha = 0.05$. We employ Friedman tests for group comparisons, followed by post-hoc analysis using Durbin tests. We report all means with standard error (SE), in figures error bars indicate standard deviation (SD) if not indicated otherwise.

III. Results

Fig. 3 shows the distribution of localization answers by the participants, which serves as the source for performance metrics such as localization error (Fig. 4).

A. Localization Accuracy

The overall mean localization accuracy is 19.5 ± 1.6 mm (SE), while the median is 13.5 mm with a standard deviation (SD) of 19.2 mm. Notably, 75 % of all localizations are within a 24 mm radius of the center of the 10 mm diameter actuator.

The distributions of localizations form a circular cluster for actuator 1. They form longitudinal clusters for actuators 2 and 3. At the heel (actuator 4), the cluster is transversely shaped when sitting but becomes longitudinal when standing (Fig. 3). The localization distributions for the respective actuators exhibit no overlap in the first Mahabalonis distance elipsoids [36]. However, when standing, the localization error variance for actuator 2 increases substantially in longitudinal direction, overlapping with localizations of actuator 1 and 3. Pronounced outliers appear in longitudinal direction. Notably, two times, a participant could not perceive a stimuli during the localization task for actuator 2.

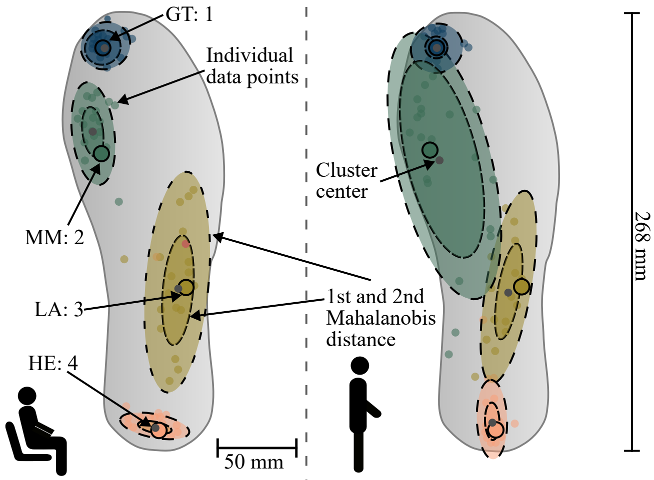


Fig. 3: Spatial distribution of perceived actuator localization in sitting (left) and standing (right) postures. Ellipsoids (dashed) summarize the cluster distributions of localization in terms of Mahalanobis distances in longitudinal and transverse direction. The localization scatter is most pronounced in the longitudinal direction of the plantar surface. In standing posture, the localization variance of actuator 2 increases substantially. Actuators 1 to 4 are placed on the great toe (GT), the medial metatarsal (MM), the lateral arch (LA) and the heel (HE).

The overall difference between the standing and sitting condition is not statistically significant. Nevertheless, the standing condition exhibits increased IQR, standard deviation, and overall variability (Fig. 4a). The average localization error varies across actuator locations (Fig. 4b) from 8.47 ± 0.7 mm (actuator 1, GT) to 28.4 ± 3.97 mm (actuator 3, LA).

The localization accuracy is significantly influenced by the actuator position [$\chi^2(3) = 11.0$, $p < 0.012$]. The Durbin post-hoc comparisons indicate significant localization differences between the actuator 1 and 2 ($p = 0.003$), 1 and 3 ($p < 0.001$), whereas there is no significant difference between actuator 1 and 4 ($p = 0.210$). Between 4 and 2 ($p < 0.037$) and 4 and 3 ($p = 0.010$), there are also significant differences.

The localization error distributions for actuator 2 changes substantially with posture (Fig. 3). The mean localization error increases for all actuators except 4. Actuator 2 showing the largest increase, though this difference does not reach statistical significance [$\chi^2(1) = 0.667$, $p = 0.414$]. The standing position leads to increased variation in localization errors for actuator 2, evidenced by enlarged IQR and SD (Fig. 3, 4c). For actuator 3, while SD remains relatively constant, the IQR decreases in the standing position. Actuator 4 has a reduced average localization error distribution while standing but exhibits more frequent large localization mismatches. The shape of its error distribution stretches longitudinally when standing. However, none of these posture-related effects reach statistical significance. Further, a Kruskal-Wallis test shows no significant difference [$\chi^2(5) = 5.15$, $p < 0.398$] in localization accuracy between participants.

B. Vibration Perceivability Rating

Analysis of the perceivability rating of the vibration finds comparable ratings across actuators 1-3, ranging from 3.36 ± 0.25 to 3.67 ± 0.16, while actuator 4 exhibits the lowest perceivability rating of 2.61 ± 0.18 (Fig. 5). A Friedman test demonstrates marginally non-significant differences between actuator locations [$\chi^2(3) = 7.27$, $p < 0.064$]. However, subsequent exploratory post-hoc pairwise comparisons using Durbin tests reveal significant differences in perceivability rating between actuators 1 and 4 ($p = 0.023$), and between actuators 3 and 4 ($p < 0.010$).

Regarding perceptibility thresholds, actuator 1 maintains consistent detectability across all trials, while actuators 2, 3, and 4 are rated as 'not perceptible' in five, one, and six instances, respectively. Notably, actuator 3 receives the highest frequency of 'easy to perceive' ratings. Actuator 2 is most often rated as 'very easy to perceive'. Posture significantly influences perception, with a Friedman test revealing reduced average perceivability ratings in the standing condition compared to sitting [$\chi^2(1) = 6.00$, $p < 0.014$, Fig. 6a].

Average perceivability drops significantly (-29%) from sitting to standing. While all stimuli are consistently detected in the sitting condition, standing yields 12 instances ($\approx$17%) of 'not perceptible' ratings. Further analysis reveals significantly decreased perceivability during standing ($p < 0.001$) for all actuators (2-4) except at the GT (Fig. 6b), accompanied by increased variability in perceivability ratings.

Although participant-specific effects are identified through a Kruskal-Wallis test [$\chi^2(5) = 13.2$, $p = 0.021$], post-hoc analysis reveals only minimal inter-participant differences, with a single significant comparison ($p = 0.048$) between two participants.

IV. Discussion

A. Localization Accuracy

Localization accuracy varies significantly across the plantar surface, with the great toe and heel showing superior performance compared to the medial metatarsal and lateral arch regions. These findings align with previously reported two-point discrimination capabilities, where the great toe shows 15 mm discriminatory distance, while medial metatarsal and lateral arch regions range from 25 to 35 mm [26].

The localization clusters form longitudinal ellipsoids, especially for actuators 2 and 3 (Fig. 3). Since the Euclidean distance calculation does not take direction into account, this clustering explains the increased variance for these actuators (Fig. 4).

The localization performance can be compared to other body locations, where vibrotactile spatial acuity has been quantified and used in prior work [3], [19]. The average localization error (19.5±1.6mm) is similar to upper arm performance, with great toe accuracy exceeding that of the wrist, and heel accuracy falling between forearm and upper arm measurements. While actators 2 and 3 exhibit slightly lower accuracy than stomach-placed actuators, direct comparison is limited by different actuator types (LRAs vs. eccentric rotating mass motors) [3]. However, the mechanoreceptor density in these compared regions on the upper body are significantly lower than on the plantar surface [37]. This indicates that mechanoreceptor density is not the sole determinator for localization capability.

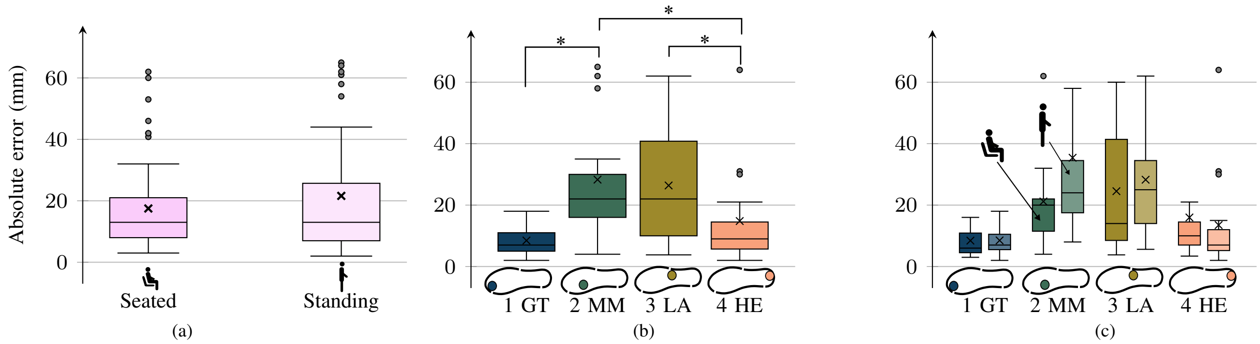


Fig. 4: Localization errors vary across postures and actuator locations. While posture shows no significant overall effect on localization accuracy (a), significant differences exist between actuator locations (b). Toe (1) and heel (4) actuators show higher accuracy compared to metatarsal actuators (2,3). When comparing postures by location (c), average accuracy decreases substantial at the medial metatarsal (2) during standing, while other locations maintain similar performance levels despite increased variability. Mean values are indicated by 'x'. Asterisks indicate statistical significant differences (Friedman test and post-hoc pairwise comparisons according Durbin tests $p < 0.05$). Outliers denote samples beyond 1.5 interquartile range above the third quartile. Actuators 1 to 4 are placed on four selected landmarks of the plantar surface namely the great toe (GT), the medial metatarsal (MM), the lateral arch (LA) and the heel (HE).

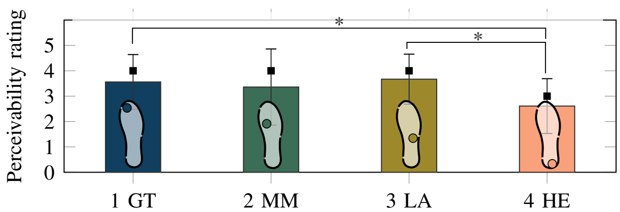


Fig. 5: Average vibration perceivability ratings show comparable levels for actuators 1-3, while heel stimulation (actuator 4) is perceived as significantly less clearly. Black squares indicate median and error bars indicate standard deviation. Asterisks indicate statistical significant differences (Friedman test and post-hoc pairwise comparisons according Durbin tests $p < 0.05$). The actuators 1 to 4 are placed on four selected landmarks of the plantar surface namely the great toe (GT), the medial metatarsal (MM), the lateral arch (LA) and the heel (HE).

Localization patterns appear linked to anatomical structures, mechanoreceptor distribution and differences in skin-thickness. The superior performance at the great toe correlates with its high mechanoreceptor density [31], while the heel's accuracy likely stems from its role as an anatomical landmark [38] rather than mechanoreceptor density. In the medial metatarsal and lateral arch regions, localization patterns have distinctive longitudinal distributions. Besides, given the variable and generally thicker nature of plantar skin, vibrotactile cues may propagate more extensively across the surface and penetrate deeper to activate FAII receptors.

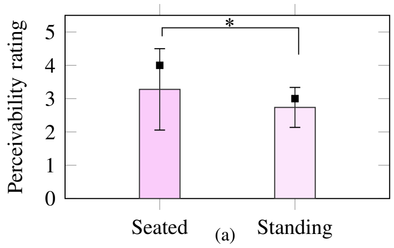
Mechanoreceptor receptive field sizes likely play a role in localization performance. Our vibration stimulus, at a frequency of 175 Hz, primarily excites FAII receptors [39], which have the largest receptive field sizes among the classes of mechanoreceptors. Their receptive field size areas correspond to ellipsoids with diameters ranging from 7.07 mm to 81.54 mm [31]. Large receptive fields integrate stimuli over their area, limiting the spatial resolution. Our results regarding localization accuracy are in the same order of magnitude as the range of receptive fields.

The spatial distribution of individual localization responses for actuators 2 and 3 (Fig. 3) is spread along the metatarsal bones that span the mid-foot from the arch to the base of the toes. This is particularly pronounced in the lateral arch region [40]. It is likely that vibrations propagate along these bone structures, resulting in stimulation of mechanoreceptors across a broader area. This appears to reduce the localization precision of the participants, as indicated by the larger absolute errors in these areas. Moreover, the resonant frequency of bone (approximately 200 Hz) being close to the actuators frequency (175 Hz) further suggests that bone conduction plays a role in vibration propagation and subsequent perception [41].

The compact arrangement of the calcaneus (heel bone) and surrounding tarsal bones may contribute to the improved localization performance at HE, contrasting with the elongated metatarsal structures. However, longitudinal localization errors during standing suggest that bone conduction may play a less significant role at this site. The localization errors increase in magnitude and variation when the participants are standing. However, the effect is not statistically significant, due to low statistical power. Further investigation is required here. However, the increased scatter of localizations for actuator 2 (MM) is substantial (Fig. 3) and is likely connected to the increase in pressure in that region during standing. When plantar pressure is increased, multiple mechanisms can influence localization accuracy. Increased pressure may improve propagation of vibration into the bone structure, which stimulates a larger area of mechanoreceptors. In addition, the increased variance may also be explained by the participants shifting their weight slightly during the standing condition.

Analysis of localization distributions (Fig. 3) reveals opportunities for optimizing actuator placement. In the seated condition, non-overlapping distributions in the toe and metatarsal regions indicate an increased spatial acuity, which suggests an increased usable actuator density. However, standing posture exhibits reduced localization accuracy, particularly in the longitudinal direction, indicating that fewer, strategically placed actuators may be more effective. The lateral metatarsal region and toe region, showing minimal distribution overlap, can accommodate an additional actuator. Conversely, the significant longitudinal scatter and outliers in the lateral arch region indicate that increasing actuator density here likely does not enhances haptic interface functionality.

Participant feedback suggests that the blank foot outline used for reporting stimulus locations may have introduced additional difficulties, as the absence of anatomical landmarks



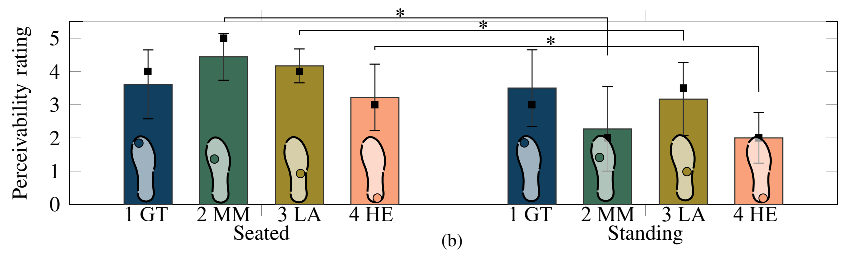


Fig. 6: Posture significantly affects vibration perceivability ratings, with a general decrease during standing (a). This reduction is most prominent at actuators 2 and 4, while actuator 1 maintains similar mean perceivability ratings across postures (b). Yet, the median perceivability rating shifts by one point in the Likert scale. This indicates that when standing the general perception threshold increases. Squares denote the median values. Error bars denote standard deviation. Asterisks indicate statistical significant differences (Friedman test and post-hoc pairwise comparisons according Durbin tests $p < 0.05$). The actuators 1 to 4 are placed on four selected landmarks of the plantar surface namely the great toe (GT), the medial metatarsal (MM), the lateral arch (LA) and the heel (HE).

required additional mental spatial mapping. This abstraction step might have impacted localization accuracy.

B. Vibration Perceivability Ratings

Vibrotactile perception is significantly reduced during standing compared to sitting, consistent with previous research [28]. This posture-dependent effect on perception likely stems from a combination of peripheral and central neural mechanisms that modify tactile sensitivity when the foot is under load. The most pronounced decrease in perceivability ratings occurred at the medial metatarsal (actuator 2) and heel (actuator 4) regions, which bear the majority of weight during standing, while the great toe (actuator 1) maintained relatively consistent perception across postures. The observed reduction in perception can be attributed to several peripheral factors. Under increased pressure during standing, the mechanical properties of the plantar skin undergo substantial changes, including alterations in thickness and hardness that affect vibration transmission to mechanoreceptors [41], [42]. These mechanical changes are particularly pronounced in weight-bearing regions [32], where skin compression may impede the propagation of vibrotactile stimuli.

The decreased vibrotactile perception during standing likely reflects an adaptive mechanism where enhanced pressure input from postural control leads to attenuated processing of additional vibrotactile stimuli [28], [43], [44]. This implies a trade-off between balance-related sensory processing and external vibrotactile perception in weight-bearing postures.

The reduced perceivability ratings during standing suggests two potential design strategies for haptic insoles: First, implementing actuators with stronger vibration amplitudes to overcome the dampening effects of weight-bearing. Second, while spatial resolution limitations discourage dense actuator placement in standing conditions, simultaneous activation of multiple actuators could enhance stimulus detection when reliable perception is critical.

C. Methodological Considerations and Future Work

The main limitation of our study is the relatively small sample size ($n = 6$), which limits the statistical power of our findings. In particular, the posture effects showed clear trends but did not reach statistical significance. Additionally, the participants stood on an isolated insole, whereas the application within shoes may yield different vibration characteristics. Also, the insole design can be improved, as the routing grooves in the TPU insole may have created unintended vibration transmitting paths. Testing was also limited to the right foot of young healthy participants, potentially overlooking age-related or clinical differences in tactile perception or bilateral asymmetries. The applicability of our results to dynamic locomotion requires further investigation, as prior work suggests vibration perception thresholds may further increase during walking [29].

Future studies can examine whether increased vibration amplitude can compensate for reduced perception in standing postures. Moreover, the effect of different actuator frequencies on the plantar surface can be explored, as the frequency used in this study (175 Hz) does not align with the peak sensitivity of FAII-mechanoreceptors (250-300 Hz). In addition, exploring LRAs with direction of vibration parallel to the skin surface, rather than only vertical, may yield different perceptual responses.

V. Conclusion

This work experimentally evaluated vibrotactile perception on the plantar surface. We show that haptic perception on the plantar surface varies significantly with posture and location. Standing reduces the stimulus perceivability ratings compared to sitting, except at the great toe. Spatial distribution mapping reveals that vibrotactile stimuli are only coarsely resolved in the lateral arch and metatarsal regions, with localization accuracy further reduced during standing due to the role of the plantar surface in weight bearing and balance. These findings have important implications for the design of haptic insoles, particularly concerning the number and placement of actuators. Future research should explore optimal actuator placement in regions with minimal perceptual overlap. Given the significant effect of posture, examining vibrotactile perception during dynamic movements, such as walking, could provide valuable insights into the influence of dynamic loading on haptic perception. Collectively, these findings, concerning the mapping of the vibrotactile perception of the plantar surface, offer guidance for developing effective vibrotactile insoles for rehabilitation, navigation, and virtual reality.

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