Effects of Vibration Direction and Pressing Force on Finger Vibrotactile Perception and Force Control

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***Abstract*—This paper reports about the effects of vibration direction and finger-pressing force on vibrotactile perception, with the goal of improving the effectiveness of haptic feedback on interactive surfaces. An experiment was conducted to assess the sensitivity to normal or tangential vibration at 250 Hz of a finger exerting constant pressing forces of 0.5 or 4.9 N. Results show that perception thresholds for normal vibration depend on the applied pressing force, significantly decreasing for the stronger force level. Conversely, perception thresholds for tangential vibrations are independent of the applied force, and approximately equal the lowest thresholds measured for normal vibration.**

***Index Terms*—Active touch, finger pressing, force control, normal vibration, tangential vibration, vibration direction, vibrotactile sensitivity.**

I. Introduction

In recent years, the ubiquity of touchscreens and touch panels drove a large body of research aiming at augmenting those surfaces with rich haptic cues, such as e.g. localized vibration [17], surface friction modulation [28], or virtual button clicks [11], [35], with demonstrable effects of vibration on the perception of surface textures and sense of motion [39].

Research employing vibration to convey haptic effects in consumer electronics is however confronted with a number of technological challenges related to various hardware factors, including the limited room available for embedding powerful actuators, power requirements, and the lack of movable parts to be actuated. In this scenario, every single advancement in understanding the psychophysics of touch can be harnessed to make the most of haptic technology, as seen for example in Apple's Force Touch trackpad, which renders illusory downward clicks produced by lateral motion at the fingertip [29]. As another example, providing maximally perceivable vibrotactile cues would allow to reduce the power requirements of haptic surfaces. A possible way to achieve such goal is to deepen our understanding of the perceptual mechanisms underpinning human sensitivity to vibration along different directions, which is the object of the present work.

The perception of normal and tangential tactile stimulation has been studied by several authors, also by means of comparative experiments. Biggs and Srinivasan [7] investigated the sensation intensity arising from normal and tangential static displacements applied to the fingerpad. The participants' index fingerpad was glued to a flat-ended, cylindrical probe tip (1 mm diameter) mounted on a 3-axis positioning robot able to move in both directions. Perceptual equivalence was found for much smaller displacements along the tangential direction than the normal one, suggesting a significantly higher sensitivity to tangential displacement. A study by Paré et al., conducted under similar experimental conditions, furthermore revealed that humans are able to assess the magnitude of slowly time-varying forces applied to the fingertip along both directions [34]. Ullrich and Cruz [43] compared pulses of normal and tangential acceleration at the fingertip for three pressure levels (0.5, 2, 5 N). Their study showed that the perceived magnitude mainly depends on the acceleration's relative values: tangential pulses were perceived as slightly weaker than normal pulses for low acceleration values, whereas they were perceived up to 40% stronger for high values. However, these differences became insignificant for the lowest pressure level. Birznieks et al. [8] found that 4 Hz sinusoidal motion along the distal-proximal axis produces greater activation of tactile afferents as compared to the same stimulation along the ulnar-radial axis. Jeong et al. [21] found that the sensitivity to normal vibration at 8, 10, 25 Hz is increased by 30% when 3 Hz tangential vibration is added.

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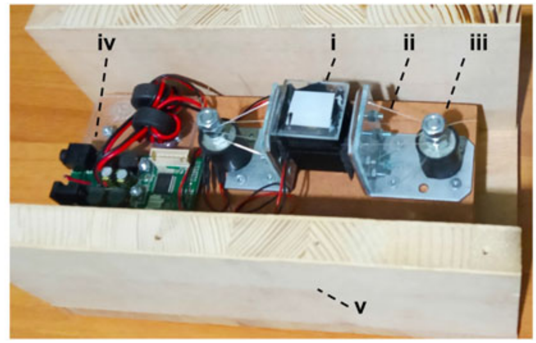
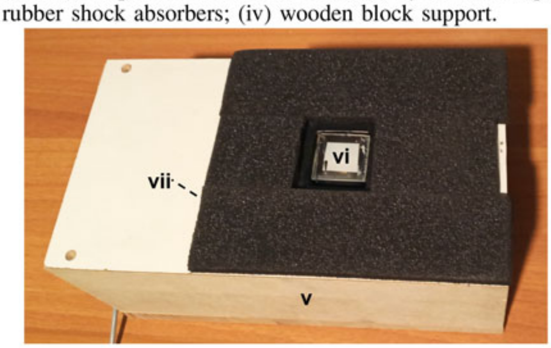
All of the above studies considered events stimulating the FA-I, or SA-I and SA-II mechanoreceptors [8], responsible for low-frequency skin motion and force grip control [22]. On the other hand, the foundational studies on human vibrotaction that deal with the frequency range targeted by Pacinian corpuscles (approximately 40-800 Hz, FA-II mechanoreceptors) [27] mainly tested vibrations in the normal direction: Among the most relevant results, the lowest sensitivity threshold for the fingertips was measured in the frequency range around 250 Hz [46], and it was found that vibration perception at high frequencies varies between humans due to several factors, including the density of receptors [2], the temperature of the skin [1], [16], and the mechanical properties of the skin related to aging effects [5]. Some effects related to gender have been identified, too, although not statistically significant, with slightly lower perception thresholds at 250 Hz in women [45]. Conversely, to the best of our knowledge, only a handful of studies have focused on the direction of vibrotactile stimuli at frequencies compatible with the Pacinian channel. In an early work, Miwa [30] reported similar thresholds along both directions for a hand in contact with a flat surface vibrating in the 3-300 Hz frequency range. More recently, Hwang et al. [19] studied the perception of vibration intensity at several frequencies between 60 and 320 Hz in users who lightly grasped a handheld mobile device mockup vibrating along the width, height, and depth directions. They found a significant, although small, effect of direction. Landin et al. [24] found that the human hand cannot easily discriminate the direction of vibration targeting the Pacinian channel, concluding that one-dimensional actuation can be as effective as complex three-dimensional vibration. In order to understand and model the propagation of shear and normal vibrations along the fingers, also involving pressing forces, Wu et al. [48] developed a finite element (FE) model of the fingertip. Their simulations show two resonances respectively at 125 and 250 Hz, independent of direction; in particular, normal vibrations produce horizontal strain of the superficial finger-pulp simulated tissues. Furthermore, their analysis indicates that the resonance magnitude at 250 Hz is independent from finger pre-compression, suggesting that perception thresholds at that frequency would not be affected by pressing force. However, subsequent *in vivo* studies partially contradicted these simulations: The measurements made by Wiertlewski and Hayward [47] on human subjects found no evidence of a resonance around 125 Hz, while other authors found that sensitivity thresholds for normal vibration at 250 Hz do decrease significantly with the pressing force [32], [33], possibly due to the fact that, for increasing forces, the Pacinian mechanoreceptors in the finger pulp are better coupled with the vibrating surface. More recently, Serhat et al. [41] developed a FE model of the fingertip, and analyzed the modes generated by forcing vibrations on normal, ulnar-radial and distal-proximal directions, predicting main resonances at 220, 122 and 164 Hz, respectively. However, the effect of pressing forces was not taken into account.

Finally, Hwang et al. [20] designed an experiment aimed at identifying finger sensitivity thresholds for vibration at 150 and 280 Hz along the normal, lateral and fore-and-aft directions, while applying a constant force equal to 0.8 N. Their study reported lower detection thresholds for the normal direction; moreover, contrarily to the previous literature [46], thresholds were measured significantly higher at 280 than 150 Hz. Concerning lateral and fore-and-aft directions – both tangential – they found comparable detection thresholds; these may be explained by measurements of human finger impedance [47], which shows similar stiffness and damping coefficients along both such directions, hence suggesting an approximately equivalent sensitivity.



(a) Inside: (i) suspended actuated element; (ii) nylon wire suspension; (iii) rubber shock absorbers; (iv) wooden block support.



(b) Outside: (v) wooden case; (vi) vibrating surface of the actuated element; (vii) black foam.

Fig. 1. The experimental device.

The present study focuses on normal and tangential vibration at 250 Hz, where sensitivity is nominally highest according to the majority of classical studies [46]. Furthermore, participants had to exert either a 0.5 or 4.9 N force with their index finger, respectively representing light touch and pressing as commonly found in various everyday actions [4], [38]. In the light of existing results [20], only the lateral (ulnar-radial) direction was taken into account for tangential stimulation.

II. Apparatus

For the purpose of our study, a device (shown in Fig. 1) was designed to generate normal or tangential vibrations at 250 Hz whose amplitude is minimally affected by normal forces applied by a pressing finger, with negligible cross-talk between directions.

Fig. 2 depicts the actuated element: it consists in a $26 \times 26 \times 36$ mm 3D-printed plastic cuboid which encloses two Lofelt L5 voice-coil actuators [6] mounted perpendicularly

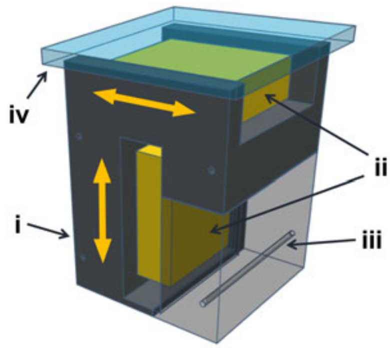


Fig. 2. Actuated element: (i) 3D-printed plastic cuboid; (ii) two Lofelt L5 actuators, vibrating along the normal and tangential direction, respectively; (iii) through-holes for suspension; (iv) 30 × 30 mm Plexiglass touch surface.

to each other, thus producing vibrations respectively along its vertical and transverse axes. The actuators are driven by a 2-channel class-T audio amplifier (Dayton Audio DTA-2). As a touch surface, a 30 × 30 mm Plexiglass panel is glued on the top of the cuboid. To maximize the actuators' efficiency, the cuboid is suspended: mechanical decoupling in fact helps reduce the actuated mass in contact with a finger pressing on top of it, meanwhile keeping actuation independent of the applied force, as detailed below. Furthermore, the suspension contributes to minimize cross-talk between normal and tangential vibrations. The resulting total mass of the actuated element is 25 gr, whereas its elastic constant (due to its suspension) is 2.6 N/mm in the normal direction.

The actuated element's supporting structure, visible also in Fig. 1(a), is outlined in Fig. 3: it consists of a wooden block (iv) housing two rubber shock absorbers (iii) that hold the actuated element (i) by means of nylon wires (ii) traversing its four through-holes (see Fig. 2). After testing several materials such as steel and copper, nylon wires were eventually selected for their good trade-off between rigidity and mechanical absorption, resulting in low dissipation particularly around 250 Hz, that is the frequency of the stimuli used in the experiment. The wooden block stands on a CZL635 load-cell sensor (v) which monitors normal forces applied to the actuated element. The analog force signal is amplified by a INA125P IC, and then sampled at 9.6 kHz with 10-bit resolution by an Arduino Mega 2560 microcontroller. The amplifier gain was set so as to read force values in the range 0-20 N with a resolution of 0.02 N.

Finally, a wooden case sized 250 × 130 × 120 mm encloses the whole system; its 2 mm plywood top is covered by a 6 mm-thick foam layer surrounding the actuator (see Fig. 1(b)).

A. Characterization and Validation

The device was characterized by measuring the resulting vibration acceleration when a human finger pressed on top of the actuated element with different forces. For this purpose, a triaxial PCB 356A17 accelerometer, whose data were sampled at 5 kHz, was fixed to the cuboid's top surface, therefore standing between the vibrating surface and a pressing fingertip. As opposed to experimental devices testing vibrotactile sensitivity

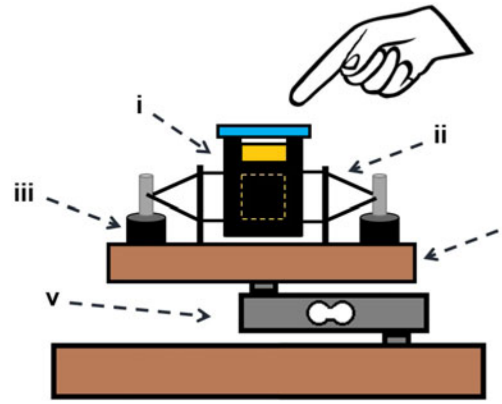


Fig. 3. Side view of the supporting structure for the actuated element: i) actuated element, ii) nylon wires, iii) shock absorbers, iv) wooden panel, v) load-cell.

under mechanically grounded vs. ungrounded conditions [18], the actuated element in our device is suspended and therefore uncoupled from its support. The mechanical conditions are thus different from those of a test device which is rigidly coupled with a immovable support or is conversely handheld.

Ten subjects (9 male, 1 female) were asked to press the index finger of their dominant hand on the accelerometer while matching a target force, and then to hold it on until the measurement was completed. The target forces were set to 0.5, 1, 2 and 4.9 N. During the procedure, participants would lay their forearm on a support and were guided by instructions appearing on a computer screen. The experimenter continuously inspected the posture of their pressing finger to ensure consistency across the measurements.

1) Influence of Applied Force on Vibration Amplitude: For each force level, the actuators played back – subsequently along each direction – six repetitions of sinusoidal vibration at 250 Hz lasting 3 s, starting at about 110 dB (re 10^{-6} m/s^2) RMS acceleration, whose amplitude was decremented by 6 dB at each repetition. Mean RMS values of the recorded acceleration were calculated by averaging the signals across eight subsequent time windows lasting 0.2 s, so as to attenuate the noise caused by unwanted finger movements during the acquisition. Then, for each amplitude level and direction, a linear model was fitted to the data as a function of pressing force. Fig. 4 shows the measurements in solid lines, and the resulting linear fit in dashed lines. The models show no linear dependence of the accelerations on force level.

2) Amplitude Response: Once vibration acceleration was proven independent from pressing force, its amplitude response was characterized in the range of interest of the experiment.

The starting level (nominal 0 dB input) corresponds to approximately 110 dB RMS as measured by the accelerometer along both directions.

Fig. 5 shows the resulting RMS acceleration values. The measured amplitude levels decrease consistently with the input down to 85 dB RMS, where some moderate over-attenuation appears for both directions. Cross-talk was also assessed, by checking RMS acceleration in the respective orthogonal axis. While cross-talk is generally not relevant down until

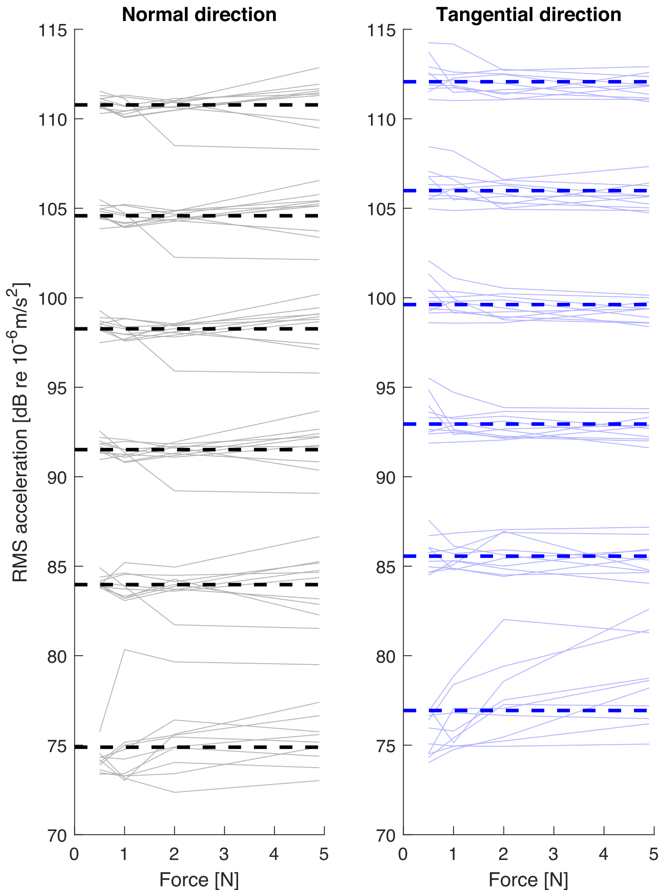


Fig. 4. Acceleration measurements for all subjects (solid lines) and respective linear fits along pressing forces (dashed lines).

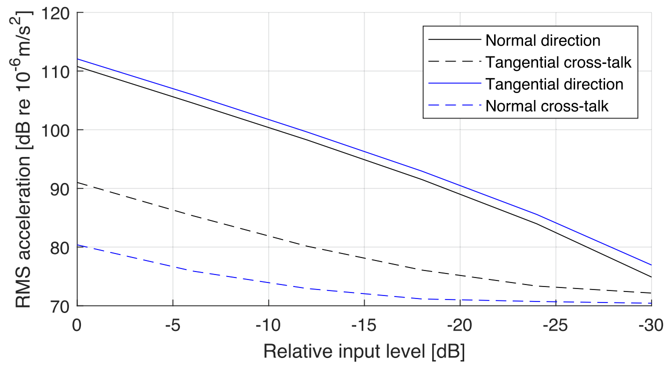
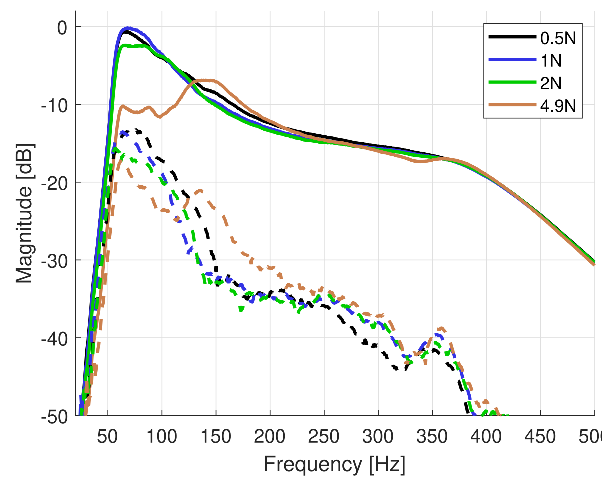
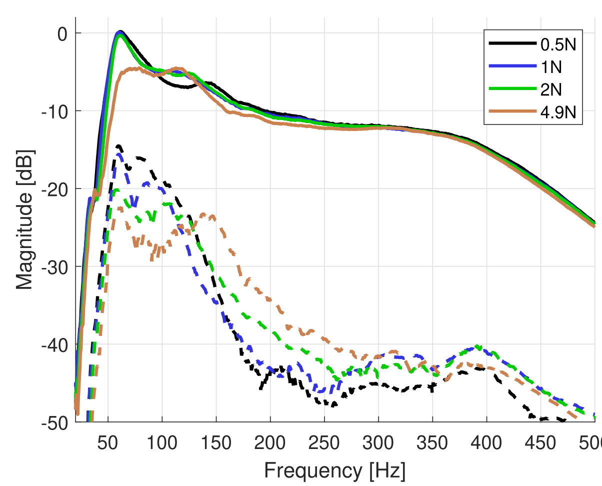


Fig. 5. RMS acceleration resulting from sinusoidal vibration input at 250 Hz as produced by the normal and tangential actuators. Solid lines represent the native direction of motion of the actuators, while dashed lines show cross-talk along the respective orthogonal axes.



(a) Normal direction.



(b) Tangential direction.

Fig. 6. Mean frequency responses for different pressing forces (solid lines) and cross-talk responses (dashed lines).

90 dB RMS, it is more prominent for the normal actuator—approximately between 15–20 dB below the native vibration direction. Also, for both directions, it becomes relatively more noticeable as the input level decreases.

3) Frequency Response: Even though our experiment focused on 250 Hz sinusoidal vibration, the frequency response of the actuated element was measured along the normal and tangential directions by exciting the device with sinusoidal sweeps between 10 and 600 Hz, lasting 15 s [12].

Fig. 6 shows means of these responses averaged over all subjects, for the various pressing forces tested. A peak is noticeable around the native resonance of the actuators (64 Hz), especially for forces up to 2 N. The respective frequency responses then gather, particularly above 200 Hz, showing that the system is effectively independent of force in the tested range [0.5, 4.9 N], and confirming the vibration amplitude measurements acquired for a 250 Hz sinusoidal input signal shown in Fig. 4.

Mean cross-talks in frequency were also measured, which are more than 20 dB lower than the frequency responses above 200 Hz.

III. Experiment

The experiment compared sensitivity to normal vs. tangential (ulnar-radial axis) vibrations at the fingertip, for pressing forces compatible with those occurring during everyday interactions [4], [38]. Namely, two force levels were considered: 0.5 N (*weak*) and 4.9 N (*strong*).

Fig. 7. Setup: a participant presses on top of the actuated element before reporting whether he felt vibration via the red (no) or green (yes) button on the console; on the computer screen, a horizontal colored bar shows the target pressing force (green range), while a white cursor moving along it represents the force currently exerted.

The following two hypotheses were tested:

H1: Sensitivity is higher for tangential than normal vibration.

H2: Sensitivity increases with finger pressing force.

Vibrations were presented only in half of the trials and, similar to what was done in [20], at each trial participants had to decide whether vibrations were present or not. Such forced binary choice kept the task simple and the session time predictable. Furthermore, binary choice results in a guess probability equal to 0.5, which penalizes decision bias, similar to two-alternative forced choice methods [31].

Separate psychometric functions – expressing the proportion of correct responses as a function of vibration acceleration – were estimated for the two force levels.

A. Participants

Twenty-two participants (15 males, 7 females) aged between 22 and 50 (M = 31, SD = 7.8) were recruited among students of the University of Udine and employees of Electrolux Professional SpA. They participated on a voluntary basis and were not paid.

B. Setup

In a quiet room, the experimental device described in Section II was placed on a table before a monitor and next to a console hosting two buttons, as shown in Fig. 7. An armrest, positioned in front of the device, allowed participants to keep their forearm comfortably aligned with it, and ensured that a consistent hand posture was used throughout the procedure. Moreover, for consistency across participants, the experimenter continuously checked that they touched the vibrating surface with their fingertip rather than the flat fingerpad. Participants wore earmuffs (Uvex K1) to prevent them from hearing any noise possibly coming from the actuators.

Vibration signals driving the actuators were generated by a RME Babyface PRO USB audio interface, reproducing digital waveforms from a laptop. Force signals from the load-cell were sampled by the Arduino microcontroller as explained in Section II, and sent to the same laptop via USB. The microcontroller also received and forwarded messages from the console buttons. A Python script running on the laptop controlled the entire procedure.

C. Stimuli

Stimuli were generated by reproducing sinusoidal waveforms at 250 Hz and lasting 1 s, i.e. a duration sufficient for the Pacinian system to decode a stable vibratory event [44]. Seven vibration amplitudes were presented along each of the two directions (normal and tangential) for both the weak (0.5 N) and strong (4.9 N) force level, for a total of twenty-eight combinations. Therefore, twenty-eight blocks of stimuli were prepared for a single session, each containing six repetitions of one specific vibrating combination, alongside six non-vibrating stimuli. In each session, the 28 blocks $\times$ 12 stimuli $=$ 336 trials were presented in randomly balanced order, both within and between blocks to avoid possible adaptation effects [14].

D. Vibration Amplitude Ranges

Due to the known variability of touch sensitivity [16], before the actual experiment a pilot test was performed in equivalent conditions with eight participants (6 male, 2 female), aimed at assessing convenient vibration intensity ranges to be used in the main test.

Initially, participants were presented with a 250 Hz vibration at 120 dB RMS acceleration (re 10^{-6} m/s^2), that is at a level which would be easily perceived by the majority of people [33], [46]. Stimuli were then decreased by 3 dB at each subsequent trial, until becoming imperceptible. Four sequences of trials were played, one for each combination of direction and force. Participants had to adjust the exerted pressing force until matching a given target as shown on the computer screen (see Fig. 7). At each trial, they had to report whether vibrations were perceivable or not by operating the console buttons (red for no, green for yes).

The resulting perception thresholds are summarized in the four boxplots of Fig. 8, one for each factor combination.The ranges of vibration amplitudes for use in the main test were then chosen so as to largely cover those of the measured pilot data, and especially extend to smaller amplitudes. Seven vibration levels were finally selected in each range, evenly distributed at 3 dB from each other. The pilot test participants did not take part in the main experiment.

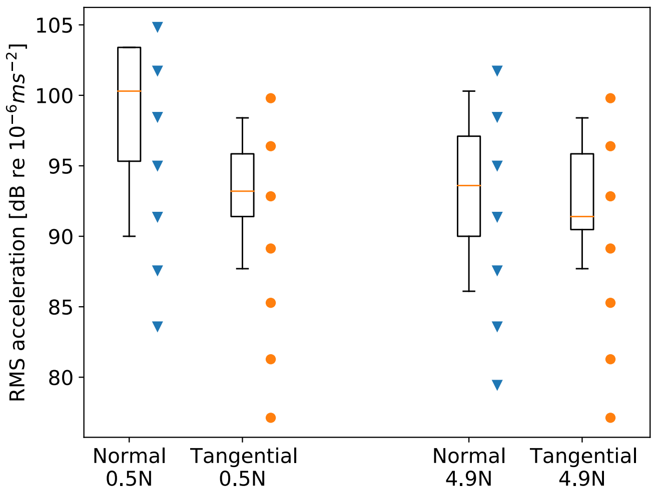


Fig. 8. Boxplots of perception thresholds measured during the pilot test, for each combination of pressing force and vibration direction. The seven amplitudes selected for use in the main test are marked in blue and orange, respectively for normal and tangential vibration.

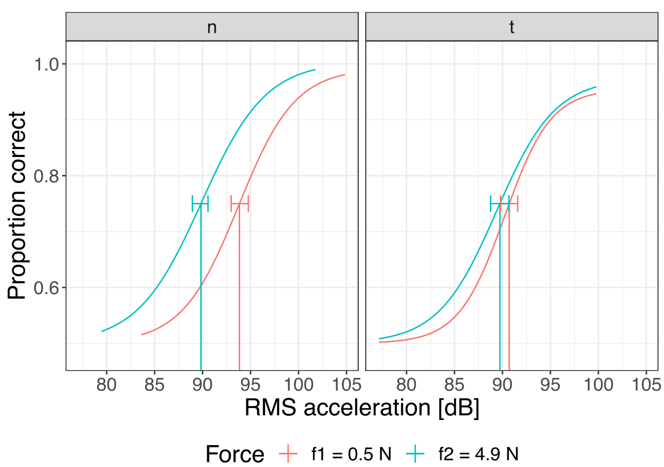


Fig. 9. Estimated psychometric functions of the RMS acceleration for normal (left plot) and tangential (right plot) vibration.

E. Procedure

Before each session, the following data were collected from participants: age, gender, dominant hand, and finger temperature (using an infrared thermometer). Then, they got briefed about the experimental protocol. A short initial training phase was carried out, presenting stimuli at the highest intensity measured in the pilot test. In addition to letting participants familiarize with the task, the training phase also made sure that all of them could perceive the starting vibration level, thus preventing potential flooring effects in individual data.

During the main test, participants pressed the tip of their dominant index finger on the actuated element and adjusted the exerted force so as to match a given target. Targets were represented as the range spanning ±20% around the respective nominal force value (0.5 or 4.9 N), and depicted as a green bar on the computer screen (see Fig. 7). As soon as the exerted force was held within such range continuously for at least 1 s, then a stimulus was presented. If the exerted force deviated from the target value by more than ±20% (i.e. if the cursor got out of the green bar) during stimulus presentation, then the trial was repeated. At each trial, participants had to report whether they had perceived a vibration or not, by respectively pressing the green or red button on the console. A new trial started automatically 0.8–1.2 s (interval picked randomly to prevent anticipation) after one of the response buttons was pressed and the finger was lifted from the actuated element.

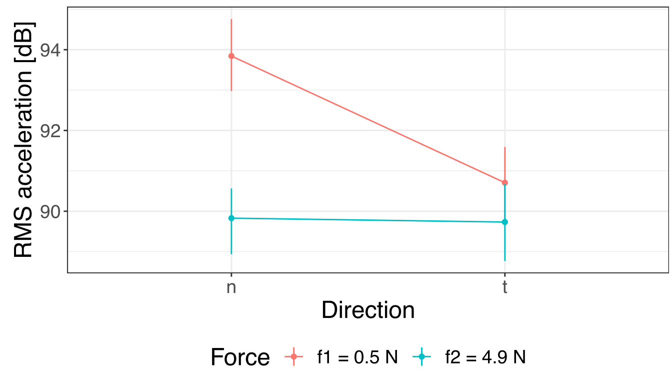


Fig. 10. Estimated vibration perception thresholds (75% correct).

Participants were allowed to rest between blocks of stimuli. Each session lasted 40 to 60 minutes. None of the participants reported fatigue in completing the session.

IV. Results

Each trial generated a record containing the following data: factor combination, decision, time to make the decision, and mean force applied during the stimulus. Proportion correct p_c was computed on each block for all factor combinations and participants, as follows [31]:

$$p_c = \frac{\text{hits} + \text{correct rejects}}{\text{trials } (=12) \text{ in a block}}$$

Data were analysed using the software R 4.0.2 with the quickpsy [25] and brms [10] packages.

A. Psychometric Functions

Psychometric functions were estimated by fitting to each factor combination a sigmoidal (i.e. s-shaped) logistic curve of the form [42]:

$$p_c(x) = \gamma + (1 - \gamma - \lambda) \cdot \frac{1}{1 + e^{-k(x - x_0)}}, \tag{1}$$

where γ is the guess probability (0.5), λ is the estimated lapse rate, while k and x_0 are the estimated slope and estimated midpoint of the sigmoid, respectively. The explanatory variable x represents vibration amplitude.

The estimated functions are presented in Fig. 9, while Fig. 10 shows the estimated perception thresholds at 75% correct, corresponding to the midpoints of the curves. Differences are apparent between the thresholds, while the slopes seem stable across conditions. For tangential vibrations, the effect of pressing force is small, with threshold estimates of 90.2 dB and 89.4 dB RMS acceleration for weak (f1) and strong (f2)

TABLE I
POPULATION-LEVEL EFFECTS OF THE NONLINEAR MIXED EFFECTS MODEL

Effect	Estimate	l-95% CI	u-95% CI
η intercept	-0.32	-1.35	0.68
η amplitude	3.82	3.11	4.68
η direction-n	-1.22	-1.66	-0.80
η strong force	0.47	0.06	0.90
η direction-n:strong f.	1.35	0.78	1.96

force values, respectively. By contrast, the effect of force is more evident for normal vibrations: the respective thresholds in fact amount to 93.7 dB and 89.9 dB RMS acceleration.

B. Statistical Analysis

The psychometric functions suggest that perception thresholds are affected by vibration direction for the lower pressing force level, and by force for the normal direction, speaking in favor of the partial validity of hypotheses H1 and H2.

In order to estimate the effects of both predictors, we standardized the continuous acceleration amplitude variable $\mathrm{zamplitude}$, then fitted the following nonlinear mixed effects model, and finally estimated its parameters by Bayesian inference [10], [23]:

$$\text{proportion correct} \sim 0.5 + (1 - 0.5) \cdot \frac{1}{1 + e^{-\eta}}$$
$$\eta \sim 1 + \mathrm{zamplitude} + \mathrm{direction} * \mathrm{force} + (1 + \mathrm{zamplitude}|\mathrm{subject}) \quad (2)$$

where the parameter η of the logistic function depends on amplitude, direction, force, and the interaction direction:force. In addition to these population-level effects, individual intercept and slope of the amplitude effect were estimated for each subject. A normal prior was used for η. Because the nonlinear link function is given in the prediction formula, the distribution of the response variable was specified as binomial with an identity instead of a logistic link in the function call.

1) Population-Level Effects: The estimates for population-level effects and their 95% credible intervals (CIs) are given in Table I. For the nominal factors, the baseline condition is weak force (f1) with tangential vibrations. Note that the estimate values refer to the nonlinear parameter and are therefore hard to interpret in terms of proportions correct; however, positive and negative coefficients respectively imply an increase and a decrease in the proportion correct relative to the baseline condition. Since the CIs do not contain zeros for any main effect nor the force:direction interaction, one can conclude that all these effects are significant. In line with the psychometric function estimates, strong force (f2) has a subtle effect in combination with tangential vibrations (the baseline case in Table I), and a more prominent effect in combination with normal vibrations.

2) Random Effects and Static Predictors: Participants were treated as a random variable as specified in the last term in (2). The estimated random effects produced subject-specific deviations from the group-level intercepts and coefficients dependent on amplitude. These subject-specific effects might be explained

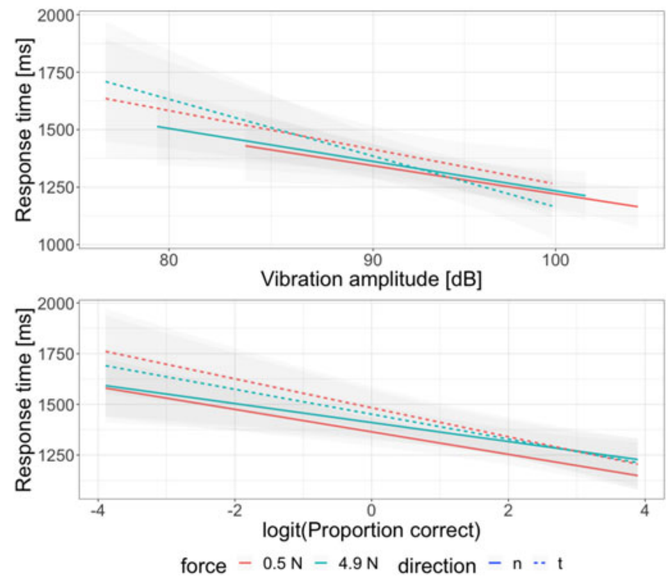


Fig. 11. Chronometric functions (top) and speed-accuracy functions (bottom) for combinations of force and vibration direction: fitted lines and 95% confidence intervals.

by the static predictors that were recorded for each participant, namely age, gender, and finger temperature. In such case, including them as predictors in the statistic model would significantly improve the fit. This was not the case, however, so the static predictors were excluded from the model.

We investigated if the subject-specific effects were correlated with any of the static predictors. The age and temperature values were standardized, and linear models were fitted between the three predictors and both random intercepts and slopes. The associations were generally not significant, as expected. A nearly significant association was observed between gender and the random slopes: female participants had a positive average random effect of amplitude (+0.63), while male participants had a negative one (-0.28), suggesting that female in general produced slightly higher proportions correct. However, a t-test between the gender groups was not significant on the $\alpha = 0.05$ level ($\mathrm{t} = -1.931, \mathrm{p} = 0.068$).

C. Response Time

According to Piéron's law [36], [37], in detection and decision tasks response time decreases with increasing stimulus intensity or choice discriminability. In the current detection task, where guessing produces 50% correct, response time is predicted to decline linearly as a function of stimulus intensity, when intensity is expressed on a logarithmic scale [9]. These chronometric functions [26] are presented in the top panel of Fig. 11. The bottom panel of Fig. 11 presents the speed-accuracy functions (SAF), the relationship between response time and logit-transformed proportions correct [9]. Although the response times generally behave as predicted, the psychometric functions fit the data better than the chronometric or SAF functions. This may be understood, however, as participants were not informed about response time measurement, nor instructed to make fast decisions.

TABLE II
MEAN ERROR AND STANDARD DEVIATION OF FORCE CONTROL

Force (N)	Vibrations	Mean error (%)	Std (%)
f1=0.5	Yes	1.4	3.3
f1=0.5	No	2.2	3.5
f2=4.9	Yes	-4.1	3.4
f2=4.9	No	-3.4	3.6

D. Pressing Force Control

Pressing force was recorded during each observation interval, whether vibrations were present or not. Force-control error was then calculated as the difference between a target force and the respective force exerted by participants.

Table II reports the relative mean error (accuracy) and the relative standard deviations (precision) of the force-control error. The presence of vibration caused lower pressing forces both in the case of a positive and negative error. Accuracy was lower for the strong force: in particular, this resulted in an average (i.e. taking into account both vibration on and off) mean error of -3.7%, highlighting a general undershooting trend, whereas, with an average mean error of 1.8%, the trend for the weak force was slightly overshooting. Precision, instead, was relatively stable, as standard deviations of the control error ranged from 3.3% to 3.6%.

A three-way repeated measures ANOVA was performed to test whether the applied force, the presence of vibrations in the stimulus, and its intensity had an effect on force control during the observation interval. The effect of force was significant ($F(1, 682) = 61.8$, $p < .001$), as well as that of the presence of vibrations in the stimulus ($F(1, 12) = 10.1$, $p < .01$). Conversely, stimulus intensity had no significant effect ($p > .05$). The interactions between factors were not significant, too.

V. Discussion

All participants but one were able to clearly perceive at least the most intense stimulus for each factor combination, confirming what found in the pilot test. The differences in sensitivities across subjects were not negligible, although the static predictors we collected did not significantly affect the models; these findings are however in agreement with previous studies [46].

The significantly higher sensitivity measured for the tangential direction is partially explained by the related literature. According to Biggs and Srinivasan [7], equivalent displacements at the fingerpad along the tangential or normal directions produce greater forces in the former case. Moreover, as simulated by Wu et al. [48], normal vibrations at 250 Hz are expected to produce a shear strain on the first skin layer: this may result in an attenuation of the energy reaching the Pacinian receptors, with correspondingly lower sensitivity.

Papetti et al. [33] as well as Oh et al. [32] reported that the sensitivity to normal vibrations increases with pressing force. Inasmuch as the mechanical behavior of our test device did not change under the considered pressing force levels (see Section II-A1), we suggest that the lower thresholds found for the normal vibration and strong force may be only attributed to the varying fingertip biomechanics during its contact with the vibrating surface. Indeed, according to [15], the contact force against a surface increases stiffness and damping of the finger pulp, making mechanical waves propagation toward the Pacinian receptors more efficient. Yet, the same study shows a slower increase of both parameters when the index finger is in abduction, with a consequently more flattened damping ratio in the 2–5 N range: if this trend was found to be similar also below such range down until 0.5 N, then the corresponding biomechanical model would not contradict our main finding, that is, perception of tangential vibrations is essentially not affected by pressing force. Further partial agreement with our finding in fact comes from biomechanical measurements made on the index fingertip in the 0.25–2 N force range, showing a slightly smoother increase of damping in the medial-lateral finger orientation than the proximal-distal one, both following a 1/3-power law in that range [47]. However, taken together such two studies would still suggest a steeper psychometric function along the tangential direction.

Similarly difficult to explain in light of our results are the findings by Hwang et al. [20], who reported lower thresholds for the normal direction. As our study suggests that sensitivity to normal vibration does change with the pressing force, this discrepancy may depend on the finger force exerted by their participants during the task [40]: however, the amplitude response of their experimental device as a function of the applied force was not reported. Likewise, the grasping force is not reported in Hwang et al.'s study based on a vibrating mobile mockup [19] which, along with the differences in the setup's ergonomics, make the small effect of vibration direction on vibrotactile sensitivity found in that study difficult to relate directly to our results.

With regard to the ability of our participants to control the exerted pressing force during the task, the weak level was generally easier to match than the strong level. Oh et al. [32] found that participants generally overshoot when asked to keep a force amounting to 0.2 N, whereas they progressively undershoot while holding force levels between 2 and 10.8 N. According to their study, our weak level amounting to 0.5 N may occupy a sweet spot in which participants could keep the force more accurately near the prescribed level, on average producing a relatively small overshoot. At any rate, Table II shows similar standard deviation values in all cases, suggesting that our participants completed the task under all conditions with comparable precision. In front of the possible need to move to a different attention level when the pressing force had to be changed, participants may have focused more on the visual feedback from the computer screen when targeting the weak force level, whereas they likely concentrated more on sensorimotor cues with the strong force.

At a deeper inspection of the same table, they seem to press with less force in presence of vibrations, meanwhile with a slightly more stable action, testified by the corresponding smaller standard deviations (i.e. increased precision). It is tempting here to associate the presence of vibration in a stimulus to correspondingly improved force control, at least for the lower force level which, with vibration, resulted both in improved accuracy and precision. This assertion may find a suggestive ground in works on the somatosensory role of vibrations for improving postural control with aging [3]. In particular, Galica et al. [13] showed that subsensory vibratory noise applied to the soles of the feet can reduce gait variability not only in participants with reduced tactile sensitivity, but to a less significance also in individuals with normal somatosensory acuity. Our experiment did not test whether our participants' sensitivity to vibration was biased by the attention needed to keep the force within the allowed range; on the other hand, this bias might have been alleviated by the vibrations themselves, whose presence also at subsensory level made force control easier.

VI. Conclusion

Our two experimental hypotheses about the effect of vibration direction (H1: sensitivity is higher for tangential than normal vibration) and the effect of finger pressing force (H2: sensitivity increases with the applied force), were confirmed, in spite of the limited role of pressing forces for perceiving tangential vibration. Our study, hence, suggests that haptic surfaces rendering vibration patterns tangential to the fingertip are likely to convey more stable cues, provided sufficient independence of their actuation from the applied pressing force. However, the production of vibrotactile stimuli in the tangential direction poses some additional technical issues as compared to the normal direction: for instance, a specific mechanical design may be needed to allow for the lateral displacement of touch surfaces; moreover, the possible slip effect between a fingertip and the surface's material may weaken the vibration effectively reaching mechanoreceptors [7].

Expanding on the present work, in the future our experimental device may be used to investigate finger sensitivity at different frequencies and force levels, resulting in a dataset useful for the validation of bio-mechanical models of the fingertip. Further experiments, involving a large sample of participants, could also address the effects of static predictors in more detail than the present study. The combination of concurrent stimulation along multiple directions is also envisioned. Taken together, the presented results have the potential to inform the design of haptic surfaces displaying rich virtual textures and materials in response to finger tapping and pressing, with possible applications to advanced user interfaces and the personal extended reality.

References

[1] G. A. Gescheider, J.M. Thorpe, J. Goodarz, and S.J. Bolanowski, "The effects of skin temperature on the detection and discrimination of tactile stimulation," *Somatosensory Motor Res.*, vol. 14, no. 3, pp. 181–188, 1997.

[2] A. Abdouni, G. Moreau, R. Vargiolu, and H. Zahouani, "Static and active tactile perception and touch anisotropy: Aging and gender effect," *Sci. Rep.*, vol. 8, no. 1, Dec. 2018, Art. no. 14240.

[3] A. Aboutorabi, M. Arazpour, M. Bahramizadeh, F. Farahmand, and R. Fadayevatan, "Effect of vibration on postural control and gait of elderly subjects: A systematic review," *Aging Clin. Exp. Res.*, vol. 30, no. 7, pp. 713–726, 2018.

[4] J. Alexander, J. Hardy, and S. Wattam, "Characterising the physicality of everyday buttons," in *Proc. ACM Int. Conf. Interactive Tabletops Surfaces*, 2014, pp. 205–208.

[5] E. Amaied, R. Vargiolu, J. M. Bergheau, and H. Zahouani, "Aging effect on tactile perception: Experimental and modelling studies," *Wear*, vol. 332, pp. 715–724, 2015.

[6] A. Berrezag and D. Büttner, "Vibrating actuator,", U.S. Patent: 10,218,250, Feb. 26, 2019.

[7] J. Biggs and M. A. Srinivasan, "Tangential versus normal displacements of skin: Relative effectiveness for producing tactile sensations," in *Proc. IEEE 10th Symp. Haptic Interfaces Virtual Environ. Teleoperator Syst.*, 2002, pp. 121–128.

[8] I. Birznieks, P. Jenmalm, A. W. Goodwin, and R. S. Johansson, "Encoding of direction of fingertip forces by human tactile afferents," *J. Neurosci.*, vol. 21, no. 20, pp. 8222–8237, 2001.

[9] C. Bonnet, J. F. Ars, and S. E. Ferrer, "Reaction times as a measure of uncertainty," *Psicothema*, vol. 20, no. 1, pp. 43–48, 2008.

[10] P.-C. Bürkner, "Advanced Bayesian multilevel modeling with the r package brms," *R. J.*, vol. 10, no. 1, pp. 395–411, 2018.

[11] Y. D. Pra, S. Papetti, F. Fontana, H. Järveläinen, and M. Simonato, "Tactile discrimination of material properties: Application to virtual buttons for professional appliances," *J. Multimodal User Interfaces*, vol. 14, no. 3, pp. 255–269, 2020.

[12] A. Farina, "Advancements in impulse response measurements by sine sweeps," in *Proc. Audio Eng. Soc. Conv. 122. Audio Eng. Soc.*, 2007, pp. 1–21.

[13] A. M. Galica et al., "Subsensory vibrations to the feet reduce gait variability in elderly fallers," *Gait Posture*, vol. 30, no. 3, pp. 383–387, 2009.

[14] A. K. Goble and M. Hollins, "Vibrotactile adaptation enhances amplitude discrimination," *J. Acoustical Soc. Amer.*, vol. 93, no. 1, pp. 418–424, 1993.

[15] A. Z. Hajian and R. D. Howe, "Identification of the mechanical impedance at the human finger tip," *J. Biomechanical Eng.*, vol. 119, pp. 109–114, 1997.

[16] C. Hatzfeld, S. Cao, M. Kupnik, and R. Werthschützky, "Vibrotactile force perception–absolute and differential thresholds and external influences," *IEEE Trans. Haptics*, vol. 9, no. 4, pp. 586–597, Oct.–Dec. 2016.

[17] C. Hudin, J. Lozada, and V. Hayward, "Localized tactile feedback on a transparent surface through time-reversal wave focusing," *IEEE Trans. Haptics*, vol. 8, no. 2, pp. 188–198, Apr.–Jun. 2015.

[18] I. Hwang and S. Choi, "Effect of mechanical ground on the vibrotactile perceived intensity of a handheld object," in *Proc. Int. Conf. Hum. Haptic Sens. Touch Enabled Comput. Appl.*, 2012, pp. 61–66.

[19] I. Hwang, J. Seo, M. Kim, and S. Choi, "Vibrotactile perceived intensity for mobile devices as a function of direction, amplitude, and frequency," *IEEE Trans. Haptics*, vol. 6, no. 3, pp. 352–362, Jul.–Sep. 2013.

[20] J. Hwang and W. Hwang, "Vibration perception and excitatory direction for haptic devices," *J. Intell. Manuf.*, vol. 22, no. 1, pp. 17–27, 2009.

[21] H.-Y. Jeong, M. Higashimori, and M. Kaneko, "Improvement of vibration sensitivity by tangential vibration," *J. Robot. Mechatron.*, vol. 21, no. 4, pp. 554–562, Aug. 2009.

[22] R. S. Johansson and G. Westling, "Signals in tactile afferents from the fingers eliciting adaptive motor responses during precision grip," *Exp. Brain Res.*, vol. 66, no. 1, pp. 141–154, 1987.

[23] J. K. Kruschke, *Doing Bayesian Data Analysis - A Tutorial With R., JAGS, and Stan*. New York, NY, USA: Academic Press, 2nd ed., 2014.

[24] N. Landin, J. M. Romano, W. McMahan, and K. J. Kuchenbecker, "Dimensional reduction of high-frequency accelerations for haptic rendering," in *Proc. Int. Conf. Hum. Haptic Sens. Touch Enabled Comput. Appl.*, 2010, pp. 79–86.

[25] D. Linares and J. López-Moliner, "Quickpsy: An r package to fit psychometric functions for multiple groups," *R. J.*, vol. 8, no. 1, pp. 122–131, 2016.

[26] S. W. Link, *The Wave Theory of Difference and Similarity*. Mahwah, NJ, USA: Lawrence Erlbaum Associates, Inc., 1992.

[27] W. R. Loewenstein and R. Skalak, "Mechanical transmission in a Pacinian corpuscle. an analysis and a theory," *J. Physiol.*, vol. 182, no. 2, pp. 346–378, 1966.

[28] D. J. Meyer, M. A. Peshkin, and J. E. Colgate, "Dynamics of ultrasonic and electrostatic friction modulation for rendering texture on haptic surfaces," in *Proc. IEEE Haptics Symp.*, 2014, pp. 63–67, .

[29] M. Minsky and S. J. Lederman, "Simulated haptic textures: Roughness," *ASME Dyn. Syst. Control Div.*, vol. 58, pp. 421–426, 1996.

[30] T. Miwa, "Evaluation methods for vibration effect part 3. Measurements of threshold and equal sensation contours on hand for vertical and horizontal sinusoidal vibrations," *Ind. Health*, vol. 5, no. 3-4, pp. 213–220, 1967.

[31] A. N. Macmillan and C.D. Creelman, *Detection Theory: A User's Guide*. Mahwah, NJ, USA: Lawrence Erlbaum Associates, 2nd ed., 2005.

[32] S. Oh and S. Choi, "Effects of contact force and vibration frequency on vibrotactile sensitivity during active touch," *IEEE Trans. Haptics*, vol. 12, no. 4, pp. 645–651, Oct.–Dec. 2019.

[33] S. Papetti, H. Järveläinen, B. L. Giordano, S. Schiesser, and M. Fröhlich, "Vibrotactile sensitivity in active touch: Effect of pressing force," *IEEE Trans. Haptics*, vol. 10, no. 1, pp. 113–122, Jan.–Mar. 2017.

[34] M. Paré, H. Carnahan, and A. M. Smith, "Magnitude estimation of tangential force applied to the fingerpad," *Exp. Brain Res.*, vol. 142, no. 3 pp. 342–348, 2002.

[35] C. Park, J. Kim, D.-G. Kim, S. Oh, and S. Choi, "Vibration-augmented buttons: Information transmission capacity and application to interaction design," in *Proc. CHI Conf. Hum. Factors Comput. Syst.*, 2022, pp. 1–13.

[36] H. Piéron, "Recherches sur les lois de variation des temps de laten- ce sensorielle en fonction des intensités excitatrices," *L'Ann'e Psychol.*, vol. 20, pp. 17–96, 1914.

[37] H. Piéron, *The Sensations*. New Haven, CT, USA: Yale Univ. Press, 1950.

[38] M. Riddle, J. MacDermid, S. Robinson, M. Szekeres, L. Ferreira, and E. Lalone, "Evaluation of individual finger forces during activities of daily living in healthy individuals and those with hand arthritis," *J. Hand Ther.*, vol. 33, no. 2, pp. 188–197, Apr. 2020.

[39] C. P. Ryan, G. C. Bettelani, S. Ciotti, C. Parise, A. Moscatelli, and M. Bianchi, "The interaction between motion and texture in the sense of touch," *J. Neuriophysiol.*, vol. 126, no. 4, pp. 1375–1390, 2021.

[40] A. S. Schmelt, V. Hofmann, E. C. Fischer, M. C. Wurz, and J. Twiefel, "Design and characterization of the lateral actuator of a bimodal tactile display with two excitation directions," *Displays*, vol. 54, pp. 34–46, 2018.

[41] G. Serhat and K. J. Kuchenbecker, "Free and forced vibration modes of the human fingertip," *Appl. Sci.*, vol. 11, no. 12, 2021, Art. no. 5709.

[42] B. Treutwein and H. Strasburger, "Fitting the psychometric function," *Percept. Psychophys.*, vol. 61, no. 1, pp. 87–106, 1999.

[43] C. Ullrich and M. Cruz, "Perceived strength of lateral acceleration display for touchscreen emulation of mechanical switch transients," in *Proc. Int. Conf. Hum. Haptic Sens. Touch Enabled Comput. Appl.*, 2008, pp. 331–336.

[44] R. T. Verrillo, "Temporal summation in vibrotactile sensitivity," *J. Acoustical Soc. Amer.*, vol. 37, no. 5, pp. 843–846, 1965.

[45] R. T. Verrillo, "Comparison of vibrotactile threshold and suprathreshold responses in men and women," *Percep. Psychophys.*, vol. 26, no. 1, pp. 20–24, 1979.

[46] R. T. Verrillo, "Vibration sensation in humans," *Music Perception: An Interdiscipl. J.*, vol. 9, no. 3, pp. 281–302, 1992.

[47] M. Wiertlewski and V. Hayward, "Mechanical behavior of the fingertip in the range of frequencies and displacements relevant to touch," *J. Biomech.*, vol. 45, no. 11, pp. 1869–1874, 2012.

[48] J. Z. Wu, D. E. Welcome, K. Krajnak, and R. G. Dong, "Finite element analysis of the penetrations of shear and normal vibrations into the soft tissues in a fingertip," *Med. Eng. Phys.*, vol. 29, no. 6, pp. 718–727, 2007.

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