Published in final edited form as:
IEEE Trans Biomed Eng. 2006 October ; 53(10): 2047–2054. doi:10.1109/TBME.2006.881804.

Polarity effect in electrovibration for tactile display

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Abstract

Electrovibration is tactile sensation of an alternating potential between the human body and a smooth conducing surface when the skin slides over the surface and where the current is too small to stimulate sensory nerves directly. It has been proposed as a high-density tactile display method, for example to display pictographic information to persons who are blind. Previous models for the electrovibration transduction mechanism are based on a parallel-plate capacitor, in which the electrostatic force is insensitive to polarity. We present experimental data showing that electrovibratory perceptual sensitivity to positive pulses is less than that for negative or biphasic pulses, and propose that this disparity may be due to the asymmetric electrical properties of human skin. We furthermore propose using negative pulses for insulated tactile displays based on electrovibration because their sensory thresholds were found to be more stable than for waveforms incorporating positive pulses.

Index Terms

electrovibration; sensation; waveform; polarity; tactile display; haptic

Introduction

A common experience for users of ungrounded, line-powered electric appliances (especially older ones) is the sensation of vibration or texture when the skin gently slides over a smooth, metallic part that has an alternating potential. The threshold of sensation, which is sensitive to area, skin locus and condition, subject sensitivity, and frequency, has been reported as low as 2 V 0–P[1] [1] for 50-Hz excitation on the back of the knuckle. The fingertips, having thicker skin and often greater hydration, require approximately one order of magnitude greater potential.

To feel electrovibration, the outermost layer of skin (stratum corneum) must be dry so as to be relatively nonconductive, and must move relative to the metallic surface (hereafter called the

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[1]All voltages reported are zero to peak.

electrode). At power line frequencies (usually 50 or 60 Hz), the resulting capacitive displacement current is on the order of a few μA, which is 2–3 orders of magnitude smaller than the current necessary to electrically stimulate the mechanoreceptive afferent nerves responsible for touch perception [2]. Based on these observations, Mallinckrodt [3] proposed that sensation of electrovibration is due to a frictional shear force modulated by the electrostatic attraction between the electrode and the deeper, conductive layers of skin. Relative movement is required because the stratum corneum contains no mechanosensors, and therefore compression due to the electrostatic force cannot be directly detected. However, tangential movement of the skin relative to the electrode introduces a shear force that depends on the applied normal force F_a, the electrostatic force f_e, and the coefficient of dynamic friction μ:

$$f_t(t)=\mu[F_a+f_e(t)] \tag{1}$$

where f_e varies with time according to the applied potential. This shear force deforms the skin, the sensory apparatus of which is sensitive to both normal and shear vibration in the range 40–800 Hz, with sensory thresholds as low as 0.1 μm depending on frequency and contactor size [4,5]. Although the frequency dependence of electrovibration *per se* has not been extensively studied, the phenomenon has been reported detectable over at least the range 5–1000 Hz with the lowest sensory thresholds between 50 and 300 Hz [1,6, p. 169].

Electrovibration has been proposed as a transduction mechanism for hand-explorable tactile displays, which could aid persons who are blind. Such a display would be very useful for presenting graphical or pictorial information, which is often very difficult to present verbally via screen readers [7], but which lends itself to tactual presentation, usually in the form of raised-line drawings [8]. An electronic alternative would allow more rapid access to electronic information sources.

Strong [6,9] developed a prototype of such a display comprising a matrix of 180 electrodes with 2.54 mm center–center spacing, the tops of which were coplanar with an insulating substrate. A series of perceptual experiments demonstrated that subjects were able to distinguish line pairs separated by 7.6 mm with 95% accuracy. Recognition of simple outline geometric shapes (circle, square, triangle) varied (non-monotonically, but generally increasing) between 42% and 88% for discretized shapes based on grids ranging from 4×4 to 9×9 electrodes, corresponding to overall shape sizes of 10–23 mm. In a similar experiment[2], Tang [10] obtained somewhat better results for a similar electrode array which was fabricated lithographically and coated with an 8-μm layer of polyimide. The original motivation for adding the insulator, also proposed by Strong [6, p. 218], was threefold: (1) Allow greater electric potential for a stronger sensation without skin breakdown; (2) Stabilize the sensation in the presence of sweat, which can short out the skin and reduce or eliminate the sensation[3]; and (3) Allow custom surface texture to control and stabilize the frictional characteristics of the sliding interface, which we have also investigated [11].

The electrostatic force in Eq. (1) may be calculated by considering the system as a two-dielectric, parallel-plate capacitor (Fig. 1):

$$f_e(t)=\frac{\varepsilon_0 A v^2(t)}{2\left(\frac{T_s}{\varepsilon_s}+\frac{T_p}{\varepsilon_p}\right)(T_s+T_p)} \tag{2}$$

[2]One of Tang's spatial patterns (out of twelve total) was slightly different from Strong's and some experimental details differ between the two studies.

[3]The additional insulator did not solve the problem of the sensation disappearing with sweat. We propose the following explanation. Assuming that conductive sweat coats the surfaces of both the skin and the polyimide, the electric field *at the skin–polyimide interface* will drop to a very low value. Since only the electric field at this interface (rather than across the skin and/or polyimide) can cause a modulated shear force on the skin, these sweat layers will reduce modulation of the shear force and thus reduce the sensation of electrovibration.

where

f_e Electrostatic force

ε_0 Permittivity of space

ε_s, ε_p Relative permittivities of insulators s (skin) and p (plastic = polyimide)

A Area

v Potential, total across both dielectrics

t Time

T_s Thickness of insulator s (skin)

T_p Thickness of insulator p (plastic)

In the case where there is no insulator, Eq. (2) reduces to

$$f_e(t)=\frac{\varepsilon_0 A v^2(t)}{2\left(\frac{T_s^2}{\varepsilon_s}\right)} \tag{3}$$

which was proposed independently by Grimnes [1] for this transduction mechanism and is also the standard textbook description for force on the plates of a parallel-plate capacitor[4].

While the v^2 term in Eq. (2) should make the electrostatic force insensitive to the polarity of v, we observed in earlier unpublished experiments that perceived intensity varied among waveforms having equal absolute-value-vs.-time functions. Strong [6, p. 165] also reported slight, qualitative changes in “edge sharpness,” “coarseness,” and “amplitude” when a biphasic waveform was presented with equal positive and negative magnitudes, but with varying length positive and negative phases. (The total width for both phases was always 0.8 ms, meaning that the relative split of positive and negative pulses should have been imperceptible.)

Because choice of stimulus waveform is an important design criterion for a practical tactile information display, we designed an experiment that would specifically detect polarity sensitivity in the electrovibratory transduction mechanism. Four waveforms were tested, all of which should produce an equivalent electrostatic force, and therefore sensation, if Eq. (2) holds. We found, however, that thresholds were higher for positive pulses than for negative or biphasic ones, suggesting a polarity-sensitive process in the force production. We present a tentative explanation for this behavior based on the asymmetric electrical properties of dry skin.

We did not attempt to measure $f_e(t)$ or the resulting perceived sensation magnitude directly, but instead used the method of response invariance [12, p. 33] wherein the amplitude of the electrode potential $v(t)$ (i.e., the stimulus intensity) was manipulated to achieve a sensation that subjects could just barely feel. It was assumed that the force required to achieve threshold was invariant across waveform in spite of slight qualitative differences. Since the waveforms we chose were rectangular in shape and the stimulation frequency was fixed, we may replace $f_e(t)$ and $v(t)$ with their amplitudes F and V, respectively, where F is the fixed threshold force for each subject and V is the potential required to achieve a threshold response. Applying these substitutions in Eq. (2) and solving for V, we obtain the predicted sensation threshold potential:

[4]Eq. (2) is different from that proposed by Strong [9, p. 77]; see Appendix 1.

$$V=\sqrt{\frac{2F\left(\frac{T_s}{\varepsilon_s}+\frac{T_p}{\varepsilon_p}\right)(T_s+T_p)}{\varepsilon_0 A}} \quad (4)$$

Methods

Apparatus

Electrovibratory stimuli were delivered to the skin via an insulated electrode array illustrated in Fig. 2. Electrical waveforms were produced by an arbitrary waveform generator (HP/Agilent 33120A), a high voltage amplifier (Trek PZD700, gain=100), and a current-limiter circuit described below. Fig. 3 illustrates the four waveforms, labeled A–D. The four waveforms had different polarity sequences, but the same absolute values, as functions of time. They were therefore expected to produce identical sensations and yield identical sensory thresholds.

The electrode array was fabricated at the University of Wisconsin-Madison Center for Applied Microelectronics on a silicon wafer base. The full procedure is documented elsewhere [11, 13,10]. Briefly, the bare wafer was first coated with a 0.4-μm-thick layer of chromium deposited by sputtering, followed by a 4.0-μm thick layer of polyimide (DuPont Pyralin® PI2611, ε_r=2.9), spun on as a liquid and then hardbaked. Next, layers of titanium (30 nm), copper (300 nm) and gold (50 nm) were deposited by electron beam evaporation and patterned by UV lithography to form the metal electrodes and connection leads. The electrodes were then coated with an additional layer of polyimide (7.7 μm, PI2721, ε_r=3.3) followed by a 0.3–0.4-μm thick layer of an aromatic thermosetting copolyester (ATSP, ε_r=2.3). ATSP was originally developed as an adhesion promoter in microelectronic applications [14], but was used here to prevent sweat, oil, and salt absorption by the top polyimide layer. Since the top polyimide and ATSP layers are bonded and therefore do not move relative to each other, they may be assumed to form a single insulator for our purposes. Our analyses will therefore consider a singular insulator of nominal thickness of 8 μm and refer to it as PI. The edge traces of the electrode leads were left exposed for connection to the stimulation source. The chromium layer of the wafer was grounded. The electrodes used in this study had a diameter of 2.29 mm and were arranged as a 7×7 matrix with a center-to-center spacing of 3.15 mm. Only eight of these electrodes were electrically connected, forming a square pattern with one corner missing.

The amplifier output can produce a dangerous level of current so a redundant safety circuit limited the current and charge to the electrode array even with multiple failures (such as amplifier output short to +700-V power supply or top polyimide insulator failure, or even partial failure of the safety circuit itself). The safety network comprised a series connection of three parallel RC networks (each with R = 1 MΩ, C = 0.1 μF), inserted between the amplifier output and the active electrodes, and did not affect the stimulus waveform. Thus, the subject was protected from electric shock hazard because even multiple system failures would result in < 1 mA current and < 100 μC charge into the finger, which are safe levels [15, Chap. 11]. Therefore, the only significant risk was the possibility of momentary painful electric shock should the top electrode dielectric fail. In this event (which did not happen) a subject could simply lift his or her fingertip off the electrode surface. In future practical use of electrovibration, where precise control of electrode potential may not be necessary, a simple, intrinsically-safe driving circuit could be employed [e.g., 13].

Subjects and environment

Experimental and safety procedures were reviewed and approved by the University of Wisconsin-Madison Health Sciences Institutional Review Board (IRB). Prior to experiments, subjects signed IRB-approved consent forms explaining the nature of the experiment and potential risks.

Five adult, right-handed human subjects (2F/3M) participated in this experiment. All were University of Wisconsin graduate students, age ≥ 30, and were recruited by personal invitation. One had previous experience with electrovibration in a shape identification experiment. Subjects used the index fingertip on their right hands to scan the electrode array, which was located on a table in front of them. Subjects were comfortably seated with elbow support to reduce fatigue. Room temperature was 17.5–23.5°C with a relative humidity of 40–61%. (Excessive temperature or humidity causes increased sweating, which can interfere with the sensation of electrovibration.) Subjects washed their hands with soap and water before the experiment. Before every experimental trial (below), the fingertip was briefly wiped with isopropyl alcohol, and the electrode array was wiped with acetone and isopropyl alcohol. This was done to prevent buildup of any contaminants (e.g., skin secretions and debris) that could change the frictional properties of the sliding interface during the course of the experiment.

Familiarization

The electrovibration sensation is somewhat unusual and requires practice to acquire [1,9]. After cleaning the fingertip and electrode array as described above, subjects were allowed to practice scanning the array with their fingertip so they could learn the proper finger angle (varied, 30–90°) and force (approx. 40 mN) to apply. The force must be high enough to ensure adequate contact, but low enough to prevent stick-slip, which interferes with the sensation. Although the angle and force were not controlled, subjects quickly learned how to adjust their scanning motions to maximize their ability to detect the minute electrovibratory sensation. Subjects also learned to relax to allow fluid scanning motions and minimize sweating. The stimulation voltage was fixed at 175 V for this procedure, which is well above sensation threshold. The experimenter turned the stimulation on and off several times to allow subjects to feel the difference between electrovibration and the smooth texture of the array. During this familiarization process (10–15 min), no data were recorded.

Procedure

After becoming comfortable scanning the array, all subjects completed five blocks of sensation threshold runs using an adaptive procedure described below. Each block comprised four runs, one for each waveform. Each run yielded a single threshold measurement as described below. Subjects completed two (in one case, three) runs (15–30 min each) during each experimental session, with up to two sessions per day, sessions separated by at least 3 h. Therefore, each subject completed ten experimental sessions, each of which lasted no more than 80 (most, <60) min. During the first block, subjects learned how to perform the sensation threshold procedure and how to perceive low-intensity electrovibration, which is rather subtle. Data from this first block were examined to ensure subjects were performing the task correctly but then discarded. The final four blocks, counterbalanced by run and subject to minimize the effect of learning, contributed to the data reported here.

Each run contained a series of trials, each of which was preceded by finger and electrode cleaning. Each trial contained a pair of sequential test intervals separated by one second. Only one interval, chosen randomly, contained a stimulus. The length of each interval was determined by the subject, typically ranging between 30 and 50 s. This long time was necessary because subjects needed to adjust their scanning to feel confident about their responses. Subjects responded yes or no regarding whether or not they could determine which interval contained the stimulus[5].

Each run started with a suprathreshold, 350-V stimulus which was then adjusted stepwise following each trial according to the Parameter Estimation by Sequential Testing (PEST) algorithm [12, p. 162]. The initial adjustment step size was 10 V and could vary from 2.5 V to 20 V as follows. Following each "yes" response, the level was decreased by one step and following each "no" response, the level was increased by one step. The step size was doubled following each consecutive "yes" or "no" response starting with the second one, and halved following each reversed response. The run continued until the level varied by no more than 5 V for eight consecutive trials. The mean of these values was recorded as the threshold for that trial, which represents a 50% confidence of interval detection. Figure 4 shows data from a typical run.

Results and Discussion

Waveform A had a much higher mean sensation threshold (19.4 ± 1.5 V, mean ± SE) than waveforms B, C, and D (12.7 ± 0.9, 12.9 ± 0.8, and 14.1 ± 0.4 V, respectively). A repeated-measures analysis of variance on waveform and run showed that the effect of waveform was significant, $F(12, 36) = 7.17$, $p < 0.01$. Run and the waveform–run interaction were not significant ($p > 0.1$). A Newman-Keuls post-hoc comparison showed that waveform A was significantly different from the other three ($p < 0.01$) but that B, C, and D were not significantly different ($p > 0.4$). Four of the subjects and runs showed similar effects. One subject (open squares in Fig. 5) showed no apparent effect of waveform; the reason for this is unclear. Residuals were examined and found to be normally distributed and not correlated with time or predicted values. Although we did not specifically ask subjects about qualitative differences between waveforms, one subject spontaneously identified waveform B as feeling "different" from the others, without knowing which waveform was presented at any given time.

There are two possible explanations for the observed waveform polarity sensitivity, each of which will be examined in turn. The first possibility is that some component of the sensation is due to electrical stimulation of the sensory nerves (electrotactile[6] stimulation), which is known to be polarity sensitive [2,18,19,15, p. 269]. However, because of the polyimide insulator, only capacitive displacement current can flow. (The operating potentials in this experiment were well below the breakdown potential of the polyimide, which we tested to over 500 V.) The maximal capacitance (due to the area of the eight electrodes and the upper polyimide and ATSP layers, assuming 100%, skin contact, which is not possible due to the dermal ridges) is 192 pF. This capacitance, which at the highest threshold observed in the experiment (31.3 V) results in only 6 μC of charge, is much less than the 120 nC charge required for surface electrotactile stimulation on the fingertip [20]. Therefore, direct stimulation of nerves is not possible. The additional observation that the sensation in these experiments is always that of texture, rather than that of localized tingle as in electrotactile stimulation, supports this conclusion.

The second possible reason for polarity sensitivity is that one or both of the insulating layers (skin and polyimide) may not behave as ideal dielectrics. This is very unlikely for the polyimide, an excellent dielectric operating at less than one-tenth of its tested breakdown potential[7]. The skin, however (specifically the sweat ducts and the stratum corneum, deeper layers being much more conductive) has resistive properties that are both nonlinear with current

[5]We chose this confidence-based response rather than a (two-alternative) forced-choice response because pilot experiments showed that the latter method required much longer to achieve convergence. Given the lengthy experimental trials, we decided to use the shorter confidence method to reduce subject fatigue, at the risk of possibly increasing subjective bias [12]. An earlier experiment [16, p. 85], similar to the present one but also collecting forced-choice accuracy data, showed that for the eight trials contributing to the threshold data for all subjects and runs, the mean forced-choice accuracy was approximately 73%. This indicates that while subjects chose conservative detection criteria as would be expected, it was not so conservative as to call into question the validity of the method.

[6]Strong used the term "electrotactile" in reference to sensations produced by both the electrostatic modulation of frictional shear force and the electrical stimulation of nerves. We separately name these effects electrovibratory and electrotactile, respectively, which is consistent with the usage of the other investigators cited herein. Electrotactile stimulation is frequently called electrocutaneous [e.g., 17]. One of the authors (Haase) has proposed the term "vibrostatic" as a more descriptive name for electrovibration.

[7]We measured the mean (dc) current across the top polyimide/ATSP layer by placing a layer of aluminum foil on the top surface of the electrode array while stimulating with waveform B (Fig. 3) at a potential of 19.5 V 0–P. No dc component was observed; the equipment resolution was 20 nA.

and asymmetric with polarity [21–25]. For example, Grimnes [22] showed that the resistive part of the small-signal, dry skin admittance at 50 Hz quadrupled when a negative dc bias of 5 V was superimposed on the test sinusoid, compared with no bias. Positive bias (also 5 V) did not affect admittance, and the capacitive component was not significantly affected by either polarity.

While obtained on the hand dorsum rather than the fingertip[8], these results suggest that the skin may be acting as a rectifier at potentials similar to those used in our study. Furthermore, although the measured conductance in [22] was very small (10–40 μS for a 56-mm^2 electrode, corresponding to 0.6–2.6 μS for our smaller, 3.6-mm^2 electrodes), the fact that there is no other dc path to the polyimide–skin interface allows even a miniscule rectification current to at least partially "short out" the skin dielectric for the negative phases of waveforms B, C, and D. This would reduce effective thickness of the overall (i.e., polyimide + skin) dielectric, which by Eq. (2) would result in a lower threshold for these three conditions, as observed.

While we did not attempt to measure the potential at the electrode– polyimide interface to directly test this hypothesis, there is indirect evidence supporting it. The ratio between the sensation threshold for waveform A and the mean threshold for the other three waveforms is 1.47, giving some indication of the relative influence of the added dielectric layer. If we make the assumption that the skin rather completely shorts out negative pulses relative to the good dielectric properties of polyimide, we would expect a similar threshold ratio for a separate experiment that manipulated not waveform or polarity, but only polyimide thickness:

$$ratio=\frac{V(PI\,thickness=8\mu m)}{V(PI\,thickness=0)} \tag{5}$$

where the numerator represents threshold voltage at the polyimide thickness in the present experiment and the denominator represents threshold with no polyimide layer at all. Unfortunately, because of the different frictional properties of polyimide and bare metal electrodes, it is not feasible to measure the latter threshold directly for purposes of comparison. However, if we measure V at any two polyimide thicknesses, we can linearly estimate[9] the threshold at any desired thickness (including zero), as earlier suggested by Strong [9].

In fact, we performed an earlier such experiment [16, p. 77] examining the effect of polyimide thickness (5 and 21 μm) on sensation threshold. The mean threshold voltages for the thin and thick polyimide layers were 31.6 and 54.5 V, respectively[10]. Extrapolating and interpolating these data to effective polyimide thicknesses of zero and 8 μm yields threshold potentials of 24.7 and 35.9 V, respectively. The ratio in Eq. (5) is then 1.45, very close to the expected value.

A similar analysis was performed using Strong's published data, which show threshold values of 73 and 100 V for insulator[11] thickness of 12.7 and 25.4 um, respectively. The predicted thresholds for the zero and 8 μm cases are then 36 and 59 V, respectively, yielding a ratio of 1.6, also relatively close to the expected value and supporting our hypothesis that the observed effect of waveform on sensory thresholds is due to the skin acting as a leaky dielectric for negative potentials.

[8]In a separate experiment, we also observed that fingertip skin has somewhat higher conductance for negative pulses than for positive ones; see Appendix 2.

[9]Strictly speaking Eq. (4) predicts a nonlinear relationship between Tp and V. However, over a wide range of hypothetical values of the (unknown) skin parameters ε_S (1–100) and T_S (0–1000 μm), this relationship is close to linear, justifying the use of a simple linear estimation. Future studies testing at least three thickness of polyimide will be necessary to determine how accurately Eq. (4) models sensory threshold.

[10]The latter value in [16] appears to be incorrectly presented as 53.9 V.

[11]Strong used polyvinylidene (Dow Saran®) as an insulator rather than polyimide.

Although one subject reported qualitative differences between waveforms, we did not investigate this phenomenon any further in this study. Such a investigation would require normalizing the perceived intensity of the waveforms in question, so that perceived quality differences were not caused by intensity variations. However, we propose that polarity sensitivity of the electrovibratory transduction mechanism could yield qualitative differences among the waveforms we studied. Consider that waveforms A, C, and D, when adjusted to threshold level, all provide (via Eq. (2), even if modified for a partially shorted skin dielectric) a force-vs.-time profile that repeats every 10 ms. In contrast, the force profile for waveform B will repeat only every 20 ms, because the alternating polarity pulses will produce different forces. The result is that the base repetition frequency of waveform B is one-half of that for the other waveforms, and tactile frequency changes of this magnitude are readily perceivable [26]. Without polarity sensitivity, the force profile of all four waveforms would repeat at a rate of 100 Hz.

Finally, although the reason for the low relative (to mean) variation of threshold for waveform D compared with the other waveforms is not obvious, we offer a conjecture. Waveform D is the only waveform containing only negative pulses. If we accept that the skin is largely shorted out for negative pulses, sensations elicited by this waveform will be less subject to variations in the skin's dielectric properties (e.g., due to changes in subject sweating, room temperature, and humidity) than the other waveforms, i.e., the stable dielectric dielectric properties of polyimide would dominate the transduction mechanism for waveform D. This would presumably result in more stable thresholds than for the other waveforms, which would be more subject to skin conditions. Because of this finding, we recommend using negative stimulation pulses for insulated[12] tactile displays based on electrovibration.

Further experiments will be necessary to confirm that the electrical properties of the skin cause polarity sensitivity for electrovibration. One approach might be to place a thin, flexible, grounded, conductive, moisture-impermeable layer over the fingertip, allowing it to receive mechanical forces at the interface, yet removing the electrical properties of the skin from the interface. We would expect the observed polarity effect to disappear, supporting the role of the skin as its origin. This treatment would also remove the deleterious effect of sweat on the interface, stabilizing both the mechanical and electrical properties of the transduction mechanism. Sensory thresholds and suprathreshold percepts that are less subject to variations in skin condition (e.g., hydration) would likely result, potentially increasing the practicality of electrostatic-based tactile communication devices.

Acknowledgements

This work was supported by the National Science Foundation, grant IIS-0081201, and the National Eye Institute, grant R01-EY10019. We appreciate the assistance of Jacqueline Braun in designing the adaptive threshold measurement procedure.

We also thank the anonymous reviewers for several useful comments, including the suggestion to perform the additional experiment described in Appendix 2.

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[12] We would not make this recommendation for uninsulated electrovibration displays, such as those proposed by Strong, because this would drive a net direct current into the skin, possibly causing skin irritation.

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Appendix 1

Our Eq. (2) differs from that published by Strong [9, p. 77], who proposed

$$f_e(t)=\frac{\varepsilon_0 A v^2(t)}{2\left(\frac{T_s}{\varepsilon_s}+\frac{T_p}{\varepsilon_p}\right)^2}.$$

When the plastic thickness T_p is reduced to zero, this equation does not match Eq. (3), which the standard formulation for capacitor plate force. We therefore believe Strong's equation is in error. Eq. (3) was derived from first principles using a method equating electrostatic force to the ratio of differential work performed by moving a plate a differential distance [e.g., 27, p. 821], assuming distensible insulators.

Appendix 2

The following experiment was conducted following the first submission of this manuscript. Four consenting subjects (different from those used earlier) participated in a preliminary test to determine whether fingertip skin displayed asymmetric electrical properties with respect to polarity, as earlier observed on the hand dorsum [22]. Each subject gently scanned a flat titanium plate (stimulation electrode) which received waveform B (see Fig. 3) with a potential of 19.53 V 0–P, similar to the threshold potentials observed in the main study. The ipsilateral forearm rested on a large-area titanium plate (return electrode) soaked with tap water and connected to the signal generator common through a 1-kΩ current-sampling resistor. A 0.22-μF capacitor in parallel with the resistor reduced high-frequency noise. Each subject performed three trials, one each with the index, middle, and ring finger; each trial lasted approximately one minute. The fingertips and stimulation electrode were first cleaned with isopropyl alcohol.

An Agilent model 54624A oscilloscope set to averaging mode (128 samples) recorded the potentials on the fingertip electrode and across the current sampling resistor. Oscilloscope data (1600 readings for a single 200-ms waveform period) were postprocessed in Microsoft Excel. The current through the return electrode was found to be a rectangular pulse with an initial overshoot that varied between finger and subject (20–80%, 1–5 ms decay time) indicating differing electrical characteristics (resistance and capacitance) for the fingertip electrode–skin interface. The mean pulse current varied from 6 to 121 μs depending on finger and subject.

Finally, we calculated the dc stimulation electrode potential and return electrode current by computing the mean values across one averaged (128-sample) waveform period, for each trial, to determine whether the electrode–skin interface preferentially conducted positive or negative pulses. The dc potential was +31 mV for each trial, indicating a slightly asymmetric but repeatable stimulation waveform. The dc current varied from −2.47 to +0.01 μA over the twelve trials, with a mean value of −0.64 μA. Only one trial showed a positive current. These values were then expressed as percentages of the mean currents during the positive and negative sections of the waveforms (100 ms each, including the inter-pulse periods) to normalize any polarity asymmetry according to the skin conductance for that trial so that high-conductance trials did not dominate the mean. The mean of these twelve percentage values was −6.4%, indicating net negative dc. The one-tail t-value for this percentage data compared with the null hypothesis (zero dc current) was 2.65, which yields a p-value of 0.011 (11 dof). We therefore conclude that under stimulation conditions similar to those used in the main experiment, including large variations in skin impedance, the skin conductance is significantly greater for negative pulses compared with positive ones.

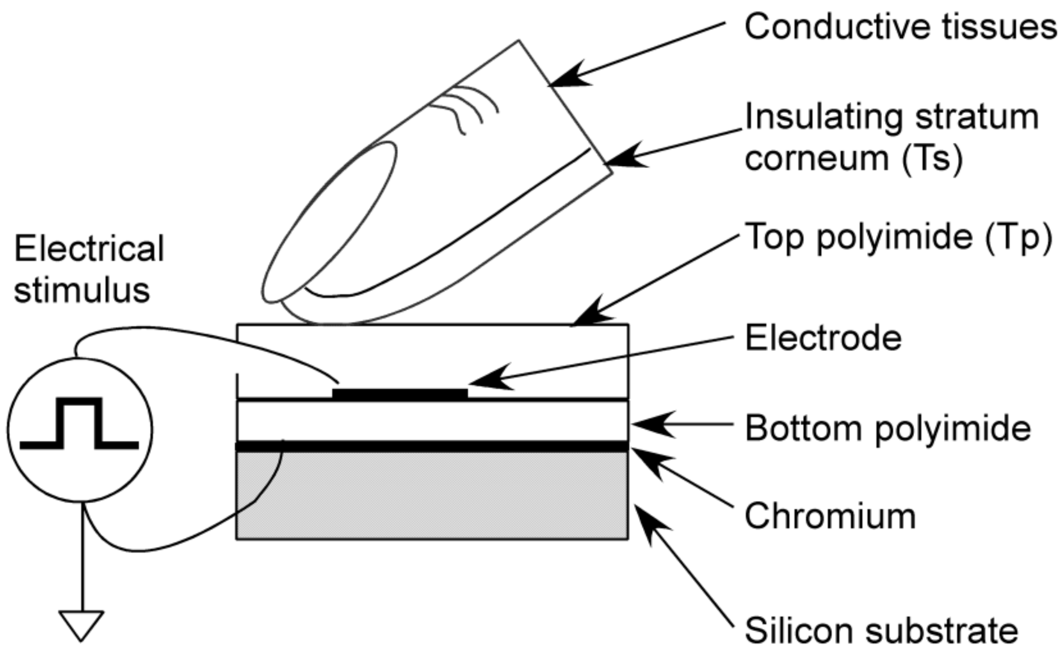

Figure 1.
Schemetic illustration of the electrode–skin interface. An electrostatic force develops between the metal electrode and the conductive skin layers beneath the relatively non-conductive stratum corneum (see text). T_s and T_p are the respective thicknesses of the stratum corneum and the polyimide layer covering the electrode. The human body is not grounded but, because of its large stray capacitance, assumes a small potential relative to ground. The results of the experiment suggest that the skin dielectric may be effectively shorted out (i.e., $T_s \approx 0$) for negative stimulation pulses, due to a polarity-sensitive conductive pathway in the stratum corneum (see discussion). According to Eq. (4), this would lower the sensation threshold potential for stimulation waveforms containing negative pulses, as observed.

Figure 2.
Photograph of finger on electrode array.

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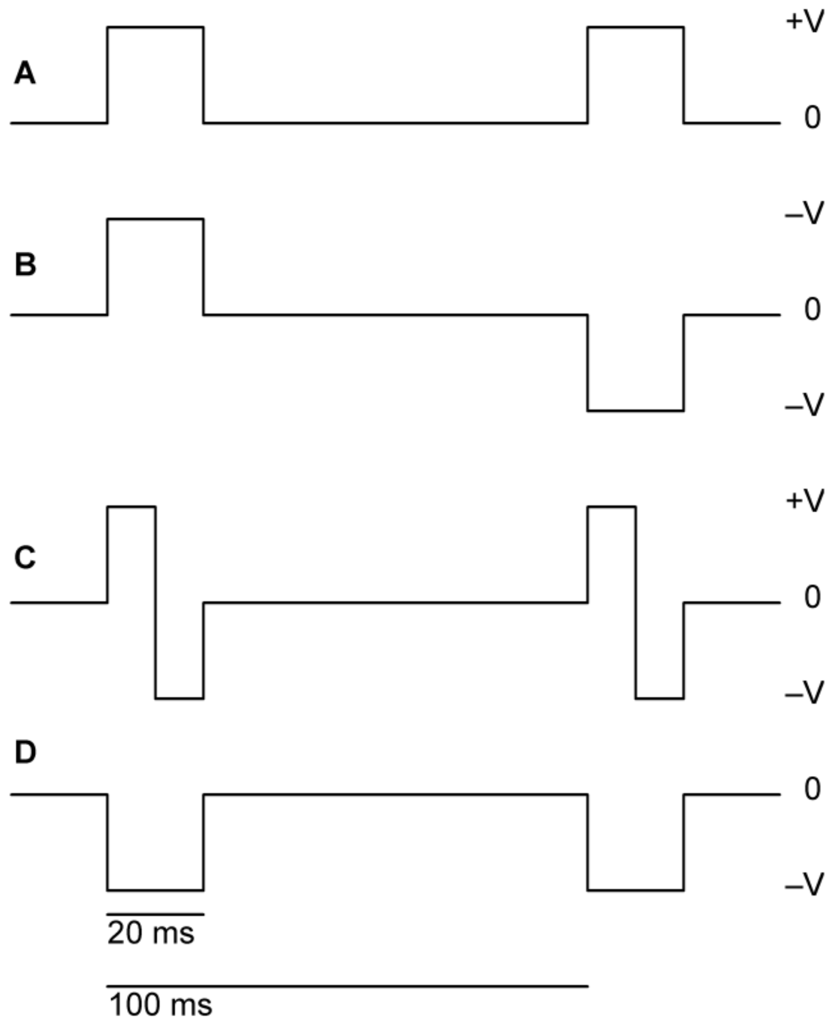

Figure 3.
Stimulus waveforms applied to the electrodes. *V* represents the 0–peak potential of each waveform.

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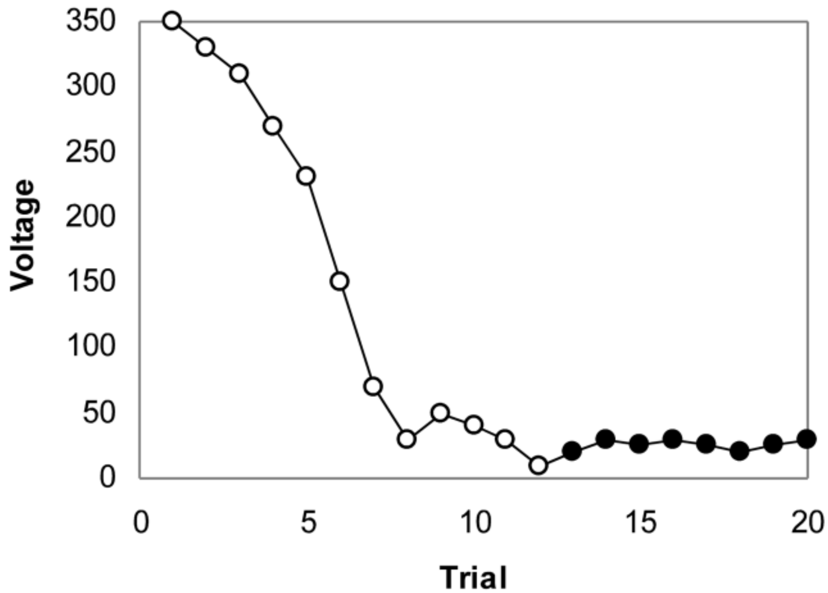

Figure 4.

Potential data from a typical sensation threshold run. The potential *V* is adjusted according to a confidence rating procedure described in the text. The mean of the final eight values (filled circles) is the threshold value for the run.

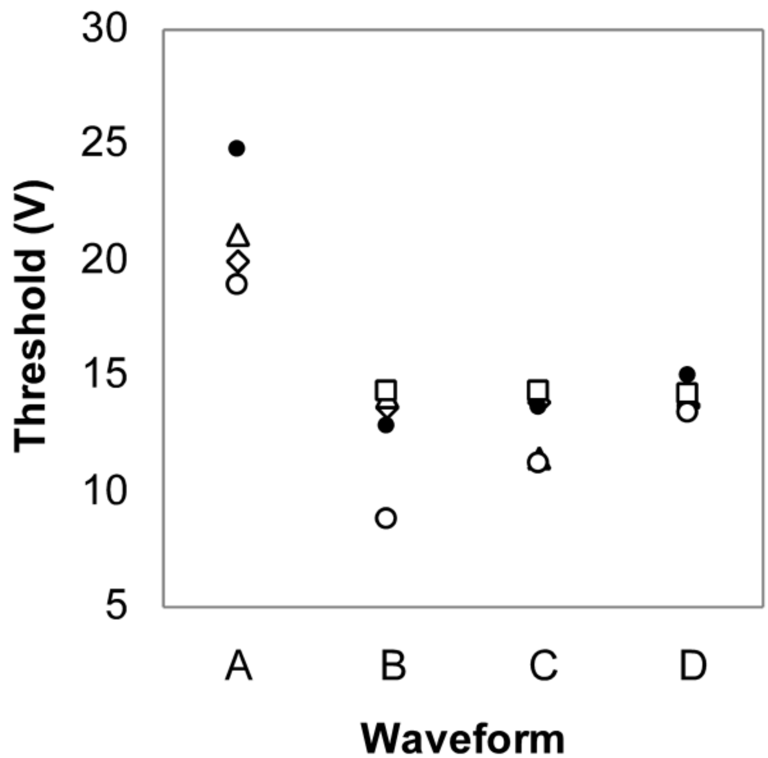

Figure 5.
Sensation thresholds for each waveform. The five symbols represent data from the five subjects.