# Electrocutaneous Stimulation for Sensory Communication in Rehabilitation Engineering

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*Abstract*—**Physiological and psychophysical functions underlying electrocutaneous stimulation are discussed, including a comparative review of stimulus parameters and coding formats. Procedures are recommended for implementing electrotactile displays and for generating reliable, painfree sensations with a useful communications bandwidth.**

## I. INTRODUCTION

IN rehabilitation engineering, technology is employed to replace or to supplement a physiological function which is impaired. In almost all cases, rehabilitative devices include a sensory component which requires the transmission of information to the user. This information may represent the primary function of the device, as in visual or auditory prostheses, or it may represent a secondary, feedback function which augments the usefulness of the device, as in sensory feedback from a mechanical arm.

The tactile sense is an attractive alternative to the major senses of hearing and vision for transmitting sensory information. Tactile communication has advantages even when the major senses are unimpaired; tactile displays can be concealed under clothing for cosmetic acceptability, can be made lightweight and efficient, and can free the major senses for other, more important tasks.

Various techniques can be used to access the tactile sense as an alternative sensory input channel. The skin is responsive to thermal, chemical, electrical, and mechanical stimuli. However, only the last two forms of stimuli are practical for applications in sensory aids.

Electrotactile displays, as implemented today, can provide densely packed, high-resolution information with a useful spatiotemporal communications bandwidth. Unlike mechanical vibrators, electrocutaneous stimulators have no moving parts and maintain a constant contact with the skin. They are efficient in terms of power consumption, simple to fabricate, and capable of producing a sensation whose frequency and intensity can be reliably controlled.

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## II. BIOELECTRIC PHENOMENA OF ELECTROTACTILE STIMULATION

### A. Electrophysiology of the Skin

*1) The Effects of Body Locus:* The outer surface of the epidermis is composed of layers of squamous epithelial cells. These cells form in the underlying dermis and migrate outward over a period of a few days, in a continuing process of regeneration. In the transition from dermis to epidermis, the cell nucleus dies; the resulting cornified epithelial tissue, when dry, exhibits a high electrical resistance (50–200K Ω); the tissue readily absorbs moisture, however, and exhibits a resistance when hydrated of less than 10K Ω [54].

The epithelial layer varies in thickness depending upon body location and repeated mechanical contact; the palmar and plantar skin surfaces are significantly thicker than other surfaces, with a correspondingly higher electrical resistance. Therefore, this higher resistance makes constant current pulses more difficult to generate and to control. The palmar and plantar surfaces are innervated by highly specialized receptors (Meissner's and Iggo's corpuscles) associated with the dermal ridges and by Pacinian corpuscles in the deeper tissues. These receptors are maximally sensitive to deformation and exhibit a wide vibratory bandwidth, useful in the discrimination of texture [7], [56]. The palmar and plantar surfaces, although superior in two-point resolution as compared with trunk skin, have a considerably smaller available area; they are also more suited for active prehensile tasks than for passive sensory displays. For these reasons, the palmar and plantar surfaces are less appropriate for long-term electrotactile stimulation.

*2) Receptor Characteristics of Hairy and Hairless Skin:* The innervation of the skin includes both corpuscular receptors and free nerve endings. The corpuscular receptor (Merkel's corpuscle) possesses a well-defined structure of some complexity, and the surrounding tissues are often modified in some manner [7]. Free nerve endings arise from within the dermis; they are densely distributed and form multiple, branching patterns of innervation.

Hair follicles are differentially distributed in the skin. The base of each follicle is encircled by nerve endings, and is maximally sensitive to mechanical movement. Sebaceous glands, whose ducts join the common pathway of the hair follicle, provide a pathway through the epidermis, as do the electrically conductive ducts of the sweat glands. Finally, occasional breaks in the epithelial tissue permit electrical current to

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reach the free nerve endings in the dermis during electrical stimulation.

The experimental evidence and the psychophysical characteristics associated with electrocutaneous stimulation suggest that the sensation is elicited by direct stimulation of primary, myelinated afferent nerves in the dermis, rather than by unmyelinated C fibers or by afferent nerves in deeper structures such as muscles [5], [6], [30], [34].

*3) Skin Locus and Preparation:* The optimal locus for an electrotactile display is an area of skin on the trunk which is either hairless or shaved, and which has been prepared by moistening with warm water to hydrate the surface epithelial layer. The abdomen has proved satisfactory for one- and two-dimensional electrotactile displays [34]; the inner surfaces of the upper arms or of the thighs are equally satisfactory, although smaller in area. Surfaces immediately covering a major underlying nerve branch, such as the wrist, are less suitable since direct stimulation of the nerve produces paresthesias or small involuntary muscle movements. Once moistened, the skin beneath an electrotactile display tends to remain moist, and the impedance remains low, partly by the vasodilating action of the stimulating current and by the passive retention of normal skin moisture under the electrode.

*4) Electrode Materials and Geometry:* It is necessary to use a stimulating surface material which does not develop nonconductive surface compounds in a bioelectric environment. Acceptable materials include gold, silver, platinum/iridium, and similar alloys. Our electrodes are constructed either from silver or from gold-plated printed circuit boards and from silver- or gold-plated rivets. An ellipsoid active center 4 mm $\times$ 7 mm is pierced with a 4 mm diameter rivet, surrounded by a 1 mm insulating annulus, and then by a 9 mm $\times$ 25 mm concentric ground electrode. The concentric configuration increases the discriminability of adjacent points and limits current spread; leakage currents, measured at the center of an adjacent (12 mm o.c.) electrode, usually do not exceed 5-8 percent of the current at the active electrode.

The electrode array should be mounted on a carrier material which maintains a positive, uniform, and stable contact with the skin during normal body movements.

*5) Electrophysiology of Single Electrodes:* Assume an electrode of the above geometry, capacitatively coupled to a constant current source. If a monophasic 10 mA current pulse is applied to moistened skin, the resulting voltage pulse at the active electrode is seen as a negatively accelerated charging curve which stabilizes at an impedance of 5–7 k$\Omega$. This impedance is initially higher (10–15 k$\Omega$) when the skin area is stimulated for the first time, when the electrode has been moved, or when the skin is inadequately moistened. The impedance will fall during the first 30 s of stimulation. We believe that this warm-up effect is a result of vasodilatation or activation of sweat glands by the passage of current. During warm-up the quality of the sensation changes from "punctate" to "blunt" as the current density stabilizes; comfort is maximized by limiting the compliance of the current source to $\pm$50 V.

The polarity of the capacitatively coupled current pulse does not affect the skin impedance. The width of the pulse also does not affect impedance once the initial charging portion is exceeded. The current amplitude, however, affects impedance in a marked manner; an increase in current is associated with a decrease in impedance. As the current approaches the pain threshold for a given pulse width, the impedance falls as a linear function of current while the active electrode voltage remains constant like a Zener diode. This condition is associated with marked skin reddening and pain [33].

By intentionally electroplating the skin [33], the multiple pathways underlying an electrode have been made visible. These pathways occur at an approximate density of 1 per mm$^2$, and are associated with hair follicles, sweat glands, or breaks in the epithelial surface. If a large (e.g., 1 cm$^2$) active electrode were to be employed, it would cover some 100 pathways, each in parallel with respect to the current source. If by chance one pathway had a lower initial impedance, more current would flow through it, which would further lower its impedance, until nearly all the current would be flowing through that one spot. This runaway condition is characterized by a sudden sting, a breakdown in the skin surface, and marked reddening. The condition can be prevented by limiting the area of the active electrode to less than 15 mm$^2$, thereby reducing the number of available parallel paths and by limiting the current amplitude to approximately 1 mA/mm of active electrode area.

For the above reason, a positive and stable contact is necessary. If a 12 mm$^2$ electrode is tipped on edge, its current density rises about tenfold, and the sensation becomes prickly and uncomfortable. Body contours and movements make it more difficult to maintain proper contact between the skin and large two-dimensional arrays [10]. Stable skin contact using linear belts, however, appears feasible [35].

*6) Intradermal Waveforms During Stimulation:* The epidermis is electrically separated from the underlying dermis by a layer of bipolar lipid cells [54]. In order to measure activity within the dermis during electrotactile stimulation, voltage pulses were recorded via a coaxial semimicroelectrode implanted in the dermis during conventional stimulation. The 0.1 mm diameter central recording tip was exposed 0.25 mm in length and encased within a 0.25 mm diameter epoxy-insulated shaft. A #21 gauge hypodermic needle (0.75 mm o.d.) was introduced into the skin of the ventral forearm at an angle of 30° and to a depth of 3 mm. The microelectrode was inserted through the needle into the dermis and the needle was withdrawn. A conventional capacitatively coupled stimulating electrode was fixed in place above the recording electrode on the epidermis. Voltages were measured relative to the external ground plane, during 10 mA, 10 $\mu$s, 100 Hz constant current pulses with alternating polarity, yielding a sensation level just above threshold [34], [37].

The observed external skin impedance for these pulses was 6 k$\Omega$, while the internal impedance was 500 $\Omega$. The continuous power dissipation required to attain a sustained threshold sensation (10 mA across 6 k$\Omega$ at a duty cycle of 1/50000) was 12 mW. At pulse onset the external voltage rose at a negatively accelerated rate, while the intradermal pulse rose linearly, as in the constant-current charging of a capacitor. At the end of the pulse the external voltage pulse decayed to zero, while the intradermal voltage remained within 10 percent of its peak until discharged by a following pulse of the opposite polarity. It therefore seemed appropriate to employ biphasic pulses, in order to prevent a direct current component (and the resulting electrochemical changes) within the dermis.

*7) Long-Term Effects of Electrical Stimulation:* In a study designed to evaluate the safety of indwelling cortical electrodes, Brummer and McHardy [4] evaluated the amount of electrical charge which could be safely passed in one direction without inducing irreversible electrolytic effects. The amount of charge passed from the electronic conductor (the electrode) to the ionic conductor (the dermis) is found as the product of current amplitude and pulse duration. For a 10 mA, 10 $\mu$s pulse, the charge transferred is therefore 100 nC; for a 25 mm$^2$ active electrode, the charge density is therefore 4 nC/mm$^2$. If safety is defined as the avoidance of irreversible reactions, then only double layer charging and surface redox reactions can accomplish charge transfer without changing the solution composition [3]. Specifically to be avoided are electrolysis of water, metal dissolution, and saline oxidation. These irreversible reactions do not occur until substantial quantities of charge have been transferred, on the order of 4 $\mu$c/mm$^2$ [4]. The levels of stimulation employed in our electrotactile displays, therefore, could be presumed to be safe on theoretical grounds since the charge transferred per pulse is one thousandth the level otherwise posited to be acceptable.

*8) The Choice of a Stimulating Waveform:* Since the dermis accumulates electrochemical changes from capacitatively coupled monophasic pulses, a biphasic waveform was indicated. In actual practice, one of us (Saunders) has observed that biphasic pulse pairs produced less long-term skin-reddening and a more comfortable sensation than did monophasic capacitatively coupled pulses. The other of us (Szeto) has completed a long-term study [51] in which 5 subjects wore single stimulators 10 h/day over three 2-week test periods. Each test period evaluated one of three waveforms, each capacitatively coupled to a concentric silver electrode on the upper arm. The three waveforms were: 1) monophasic, 150 $\mu$s pulses at 0.5–25 mA, presented at 30 pps; 2) 150 $\mu$s pulses of alternating phase at 0.5–15 mA, presented at 30 pps; and 3) 10 kHz biphasic pulses at 0.5–15 mA, presented in bursts of 15 pulses, at a burst rate of 30/s. Each test subject selected the current amplitude that produced clear and comfortable tactile sensations. The average current level chosen was 1.21 mA for monophasic 150 $\mu$s stimuli, 1.87 mA for alternating phase 150 $\mu$s stimuli, and 5.90 mA for biphasic 20 $\mu$s stimuli. Although the current levels for the three waveforms were quite different, the charges per pulse were within ±4 dB.

Despite the intense protracted stimulation regimen used in Szeto and Mao's study, periodic examinations of the stimulation site by a dermatologist, photographic records, and biopsy reports all indicated that these stimuli produced only mild, reactive, and readily reversible changes in the epidermis [51]. Biphasic stimulation was rated as the most comfortable, but its shorter pulse width required a greater current amplitude to yield an equivalently strong sensation, and therefore produced a greater amount of transient skin reddening and other reversible changes. Monophasic stimulation with longer pulses required the least amount of current for adequate sensation, and produced the least amount of transient skin reddening. In comparing current levels, comfort, clarity, and skin reactions, the alternating phase stimulation ranked in the middle. The choice of the stimulating waveform therefore can be based on the specific application involved and the encoding mechanism to be employed, rather than on safety concerns.

### B. Psychophysics of Electrocutaneous Displays

*1) Absolute Threshold as a Function of Charge:* In earlier studies using capacitatively coupled negative-going monophasic pulses [32], Saunders reported that the average threshold charge for pulses repeated at 60 and 200 Hz was 62 nC, and that this value was constant across a range of current amplitudes (1–20 mA), pulse durations (1–100 $\mu$s), and frequencies (60–200 Hz).

Saunders found that although the polarity of monophasic, capacitatively coupled pulses did not affect skin impedance, charging rate, or the shape of intradermal waveforms, the threshold charge for negative-going pulses as 10–20 percent less than that for positive-going pulses. This finding is consistent with the hypothesis that stimulation occurs by depolarizing free nerve endings in the dermis. However, the characteristics of the dermis indicated that biphasic pulses should be employed to avoid the accumulation of direct-current components and irreversible electrochemical changes.

Further study of biphasic pulses showed that if the second, opposite-phase pulse follows immediately upon the first pulse, the observed threshold charge is elevated. We hypothesized that since some minimal time must be required in order for charge transfer and depolarization to take place, the second pulse was actively removing the charge of the first before its effect was fully realized. A separation of 10 $\mu$s between the first and second pulses, or the use of alternating phase pulses, yielded the minimum threshold, and consequently, the greatest efficiency.

*2) Modulating the Intensity of the Sensation via Charge:* The intensity of monophasic, capacitatively coupled pulses could be controlled by varying the charge per pulse via modulating either pulse width or current amplitude [33], [46]. Under these conditions most experienced observers set the discomfort point at a charge which was 8 times threshold. The sensation at the higher charges, however, was qualitatively different from the threshold sensation, and less comfortable. The just-noticeable difference (JND) for intensity was estimated at 8–10 percent. These characteristics were independent of body locus, excluding palmar and plantar surfaces.

Using pulse width modulation as a means of controlling intensity, Szeto *et al.* [49] found that JND's of pulse width averaged 40 percent for pulse widths from 50 $\mu$s to 1 ms at a pulse rate of 15/s. The discrepancy between Saunders' data and Szeto's data may reflect the different stimulus parameters used.

*3) Quantal Biphasic Stimulation: A Better Control of Intensity:* Based upon Saunders' finding that electrochemical changes were retained in the dermis until discharged, he specified a quantal biphasic pulse, with sufficient charge in the first phase to reach threshold, followed by an opposite-phase second pulse at a latency greater than 10 $\mu$s. These quantal pulses were generated at a free-running rate of 10 kHz (100 $\mu$s intervals) and were gated in bursts of 1-32 pulses. The burst repetition rate was varied from 1 to 300 Hz, yielding a vibratory sensation. In this format the modulation of intensity was smooth and reliable; the higher intensities were strong, without being perceived as painful. Over a dynamic range of 1-32 pulses per burst, approximately 16 levels of intensity could be reliably discriminated. Observers rated this form of stimulation as more comfortable than monophasic, capacitatively coupled pulses of either polarity.

*4) Temporal Resolution:* The sensation of vibration fuses to a constant pressure sensation at a burst repetition rate of a few hundred Hz, analogous to mechanical vibration, and the JND's at these frequencies are quite large. However, single pulses of 10 $\mu$s duration and a charge of 100-200 nC can be perceived; the minimum temporal separation required for the perception of nonsimultaneity has been measured as 12 ms on the abdomen; and the order of presentation of two spatially distinct stimuli can be correctly identified at a separation of 40 ms [33].

The commonly accepted range of useful frequency is 1-100 Hz. Within this range, the JND for frequency has been measured as 38 percent at 2 Hz, 23 percent at 10 Hz, 16 percent at 20 Hz, and 23 percent at 100 Hz [49].

*5) Spatial Resolution:* We have taken the two-point threshold to be that minimum distance at which the order of two stimuli delivered 0.5 s apart can be determined. Saunders [33] obtained values on the abdomen of 6-9 mm on center for concentric electrodes. Even smaller separations might be possible, if nonconcentric electrodes with a surrounding ground were employed.

Using two concentric electrodes mounted on a micromanipulator and applying various monophasic waveforms at each electrode, Solomonow *et al.* [42], [43] found comparable values for electrotactile discrimination thresholds. They also found the two-point threshold to vary with body site, stimulus frequency, and time delay between the electrodes. Their most interesting finding was that learning can reduce a person's two-point discrimination threshold [44]. If a function as fundamental as the two-point threshold can be altered by learning, then it is reasonable to expect that the human ability to process electrocutaneous signals could improve significantly with practice.

*6) Temporal Interactions and Habituation:* Habituation, i.e., a change in the perceived magnitude of the sensation, occurs after prolonged electrotactile stimulation, as in other sensory modalities. The rate of habituation (or sensory adaption) is a function of the rate of stimulation; low vibratory rates (less than 10 Hz) show very little habituation, whereas the intensity of stimulation at or above 1000 Hz falls to near-threshold (or below) after a few seconds.

Although no direct studies of electrocutaneous sensory adaptation have been reported, some rough estimates are available. Szeto *et al.* [47], [50] found during their tracking studies that noticeable sensory adaptation occurs after as little as 15 min for biphasic stimulation. Monophasic stimulation appears to be less susceptible to sensory adaptation.

Receptors subserving the tactile sense are known to adapt to constant inputs of temperature and pressure [17], [23]. Since properly applied electrocutaneous stimulation can produce similar sensations, the observed sensory adaptation to both monophasic and biphasic stimulation is not surprising. Further, since electrocutaneous stimulation bypasses primary tactile receptors [5], [6], [30], its temporal characteristics would probably be different from the mechanically induced adaptation that Hahn [18] characterized. An adequate description of this phenomenon awaits further research.

*7) Spatial and Temporal Interactions in Arrays:* If the ratio of the intensities of two adjacent electrodes is varied, a phantom sensation can be perceived whose locus changes as a function of the intensity ratio. We have not, however, been able to elicit this phenomenon as a function of temporal separation or time-of-arrival.

Substantial surround-masking effects can be observed in two-dimensional arrays. These effects limit the high-spatial-frequency information which can be presented in, for example, a tactile display of visual information. The temporal resolution of the skin, however, is superior to that of the eye, and this capability is reflected in the ability of observers to integrate dynamic, moving elements of the display as in, for example, the tactile display of formant frequencies for the identification of vowels in speech.

*8) Theoretical Bandwidth of the Skin:* If each stimulator is capable of presenting 4 bits of intensity information, at rates up to 300 Hz, its maximum bandwidth is 1200 bits/s. However, extrapolation to a linear array of 32 stimulators spaced 12 mm apart is not linear because the observer cannot track 32 independent events; yet observers can accurately track and identify dynamic tactile forms constructed from 3 separately moving bands of stimulation on such an array. Comparable tracking studies for two-dimensional arrays, to our knowledge, have not been reported.

## III. Information Encoding for Sensory Communications

### *A. Information Bandwidth Limitations*

With the main bioelectric phenomena that govern electrocutaneous stimulation generally established, major efforts are now being made to elucidate how best to present important sensory information to the human receiver using electrocutaneous stimulation. Tactile perception of sensory information requires that the information be converted (or encoded) into stimulating waveforms which are clear, comfortable, and easily understandable. The efficacy of the information transfer depends on the type of codes used, the number of channels, the extent of sensory adaptation versus learning, the skin's inherent information channel capacity (or bandwidth), and the tactile display format. Some of these factors have been studied; others await further research.

A number of investigators [2], [19], [36], [52], [53] have used skin surface stimulation as well as direct stimulation of the afferent nerve to examine the channel capacity of the skin sense. Despite the various methods used, there is general agreement that the information rate possible using one electrode is around 2-5 bits/s. In the electrocutaneous tracking studies of Szeto *et al.* [46], [47], wherein various coding schemes (i.e., ways of modulating the stimulation signal) were compared, similar bandwidths were obtained. Despite the limited channel capacity when using a single electrode, some useful applications have been found for sensory feedback in upper limb prostheses [20], [28], [29], [41].

### *B. Single Electrode Codes*

Using the recommended stimulation procedures discussed earlier, a number of pulse modulation coding schemes are possible with a single electrode tactile display. For example, changes in the informational signal could be encoded as changes in pulse rate, analagous to frequency-modulated radio signals, or as changes in sensation magnitude, analagous to amplitude modulation. The sensation magnitude could be varied as a function either of the pulse amplitude (i.e., current) or the pulse width of the stimulus. Frequency and intensity variations, while not totally independent of each other, can be kept sufficiently separate as to allow them to carry different informational signals [29], by putting the pulse rate (PR) under the control of one signal and the pulse width (PW) under the control of another.

When using a single electrode to transmit one channel of information, the tactile sensations might be enhanced and the bit rate increased if PR and PW were linked so that they increased or decreased together. Alternatively, PR and PW might change in opposite directions in response to the same sensory command signal so that the perceived overall intensity level would be more constant. The best approach would of course depend on the application involved and the information rate required.

### *C. Multiple Electrode Codes*

Since only 2-5 bits of information per second per electrode can be realized, higher rates of information transfer would require multiple electrodes. Many of the pulse modulation schemes described earlier can be directly applied to multiple channel sensory communication systems, provided that some time allowance is made for the human receiver to timeshare his attention between electrodes. With multiple electrodes, the additional variables that can affect the efficacy of sensory communications include the spatial geometry, density and size of the electrode matrix, and the time relationships among the various electrodes. The physical characteristics of the electrode matrix are important because intimate and consistent electrode contact with the skin must be maintained at all times to avoid uncomfortable sensations to the human receiver.

It has also been found that if some underlying spatial relationship between the electrodes' locations and the informational signal can be incorporated into the artificial communication system, the information input rate could be enhanced and the mental processing load could be reduced. Such strategies have been used in the tactile television system [9], [12], [13] and the electrotactile vocoder [32], [35] developed at the Smith-Kettlewell Institute of Visual Sciences in San Francisco.

### *D. Evaluation of Codes*

With a little imagination one can readily derive a number of coding schemes for communicating information to a handicapped person using single or multiple electrodes as the tactile display. For single electrode systems the frequency, intensity, and stimulation site could be varied for optimal effect. For multiple electrode systems the spatial and temporal interelectrode relationships plus loci of stimulation are additional factors that could be varied. Under certain conditions, the electrocutaneous counterpart of the mechanical tactile phi phenomenon [1], [16] or somesthetic sensory saltation [15] may appear which could obscure or improve information transmission.

Because there are many ways of encoding information, careful evaluations of their effectiveness are crucial to the usefulness of the intended sensory aid. Depending on the coding scheme and the intended application, different strategies have been employed to assess the efficacies of codes. Word and speech recognition rates have generally been used to indicate the effectiveness of a particular electrocutaneous communication system [31]-[33]. Travel rates and avoidance of obstacles have been used to show the feasibility of mobility aids based on electrocutaneous stimulation [13].

Two factors, the significant amount of time and energy required plus the urgent need for some workable aid to help handicapped people, have perhaps oriented the evaluation strategies heavily toward demonstrating that some particular device is helpful and away from elucidating which coding scheme is intrinsically better than another. One methodology that appears suitable for comparing sensory feedback codes for single and dual channel communication systems has been developed by Szeto *et al.* [48], [50]. The major premise underlying their electrocutaneous tracking approach was that superior tracking performances by many subjects using a particular code (or code pair) indicated a superior code (or code pair). An effective code (or code pair) would be clear and easily interpreted by subjects, thereby allowing them to achieve superior pursuit tracking scores. The basic apparatus used to test pairs of codes for dual channel electrocutaneous communications is shown in Fig. 1.

In an electrocutaneous tracking study involving 21 subjects testing 7 codes for single channel communication, Szeto *et al.* [47] showed clearly that significant differences between codes existed. For presenting slowly varying information such as elbow joint angles or grasp forces in upper limb prostheses, the frequency modulation codes were significantly better than intensity modulation codes, whether the electrocutaneous stimulation used was monophasic or biphasic. Furthermore, low frequency modulation codes (1-15 pps) were significantly better than high frequency modulation codes (20-100 pps) because accurate frequency discrimination above 60 pps was much worse than below 40 pps. Monophasic frequency and

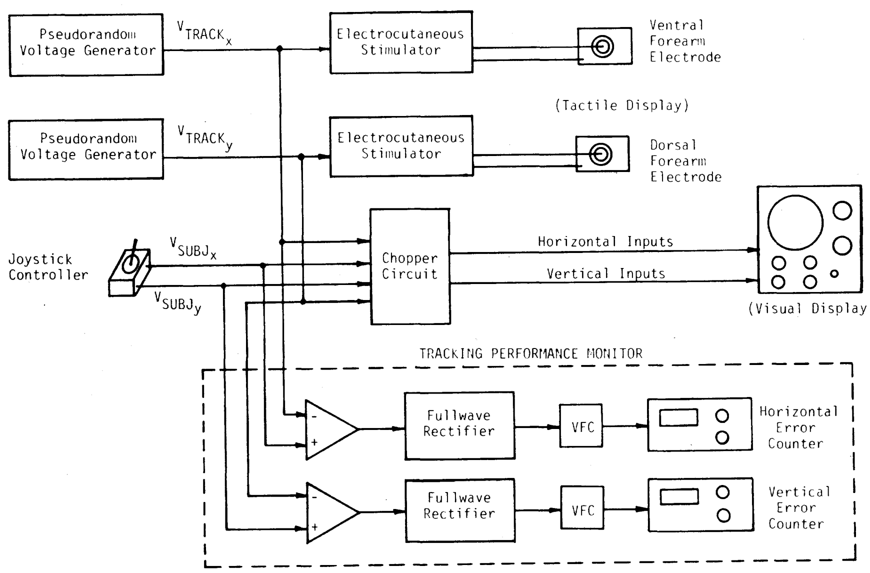


Fig. 1. System block diagram of apparatus for comparing pairs of electrocutaneous codes using a two-dimensional tracking task.

intensity modulation codes were consistently better tracked, respectively, than biphasic frequency and intensity modulation codes because the latter type of stimulation produced significant amounts of sensory adaptation.

In a dual channel tracking study of electrocutaneous code pairs, Szeto [50] found that human tracking of two independent electrocutaneously encoded random signals was a difficult task. When given up to 2 h to learn the code pairs, the subject-to-subject variance in their learning rates was large enough to obscure expected differences in effectiveness between the various pairs of codes. His results reflect clearly the information bandwidth limitations of the human being, as well as his learning ability. Although the tracking scores were not significantly different, the ease with which certain code pairs could be learned was quite different. The easiest code pair to learn was two low pulse rate modulation codes matched together. Whether or not the two electrodes were placed on the same dermatome also influenced the ability of subjects to interpret the tactile sensations. This finding is consonant with Geldard's work on somesthetic saltation [15].

### E, Sensory Adaption in Electrocutaneous Sensory Aid Applications

When electrocutaneous stimulation of constant intensity is applied, the resulting sensation quickly increases in magnitude, reaching a plateau after only a few seconds. The duration of the plateau lasts from minutes to hours depending on the stimulus parameters. Onset of sensory adaptation has been found to occur more quickly for high stimulus frequencies.

Following the plateau stage, the sensation diminishes in magnitude over time, either reaching a final asymptotic value, or perhaps disappearing entirely. When using sensory aids employing electrocutaneous stimulation, the sensory adaptation phase becomes quite important when transmitting steady-state or slowly varying signals (below 2 Hz) for 30 min or more. The problem of sensory adaptation is not so crucial for tactile vocoders because they encode transient auditory sounds, but it is more critical in visual substitution aids and prosthetic sensory feedback where the information is often slowly varying.

## IV. Applications of Electrocutaneous Stimulation in Rehabilitation Engineering

### A. Sensory Feedback in Arm Prostheses/Orthoses

The relative success of myoelectrically controlled prostheses during the 1960's has motivated many efforts in the 1970's to incorporate supplemental sensory feedback into these arms. It is ironic that such artificial feedback is more necessary in state-of-the-art powered artificial arms than in cable-operated manual arms [38]. Sensory cues formerly available in a manual arm prosthesis through cable tension and excursion and shoulder movements are missing in myoelectric controlled electrically powered arms.

Artificial sensory feedback can be introduced to the user through mechanical vibrators [22], direct afferent neural stimulation [2], [8], [24], [55] and electrocutaneous stimulation [20], [28], [29], [41]. Mechanical vibrators, however, have proven to be rather bulky and inefficient for most prosthetic applications. Direct neural stimulation techniques carry the disadvantages of being invasive; they encounter difficulties in patient acceptance [24] without being able to offer a higher information bandwidth [2], [8]. Many of the early difficulties associated with electrocutaneous stimulation have been largely resolved, thus making very attractive its potential advantages of small size, modest power demands, good reliability, and wide range of stimulus parameters.

In view of these advantages, supplemental sensory feedback using electrocutaneous stimulation has been employed by a number of investigators. Prior *et al.* have added feedback to both body-powered and externally powered artificial arms

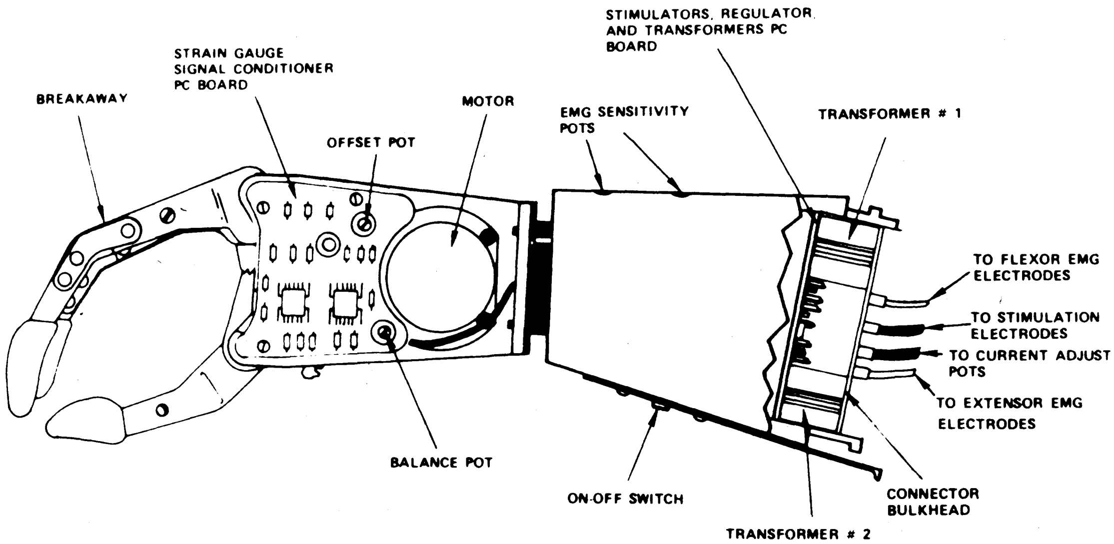


Fig. 2. Supplemental sensory feedback of grasp force and hand opening using two electrodes. Reprinted with permission from *Proc. San Diego Biomedical Symposium*, vol. 15. New York: Academic, 1976.

(Fig. 2) [26]-[29]. Interference between the myoelectric control signals and the higher voltage stimulating feedback signals, once a major concern, has been overcome either by isolating their grounds [26] or by using a sample-hold element in the control circuits [39].

Electrocutaneous sensory feedback in artificial arms typically attempts to close the control loop so that user dependence on visual feedback can be reduced and his high control burden thus lessened. Information important to the artificial arm user includes the spatial location and orientation of the hand (or terminal device), the grasp force on an object, and the position of the fingers (i.e., hand opening) [38]. One or more of these key parameters has been used in sensory feedback systems [20], [26], [39], [40].

Evaluations of prostheses having artificial sensory feedback have generally been positive as deduced by improved performance on a specific task. However, these evaluations have often been contaminated by annoying mechanical failures of the arm and/or other extraneous "noise," including improper fit and inadequate prosthetic training [21]. Until the problems of reliability and servicing and shortage of prosthetists trained to fit power prostheses are solved, the true value of supplemental sensory feedback cannot be conclusively shown. More research and development work is needed to improve not only powered arms, but also the sensory feedback system for them.

The advantages of closed-loop control are not limited to upper limb prostheses; they also apply to upper limb orthoses using exoskeletal braces and/or functional electrical stimulation (FES) of paralyzed arms. For example, Peckham *et al.* [25] found that acceptance and greater use of their FES orthosis for C5 quadriplegics would be more likely if the orthosis did not require constant visual attention (or feedback) during its operation. Without supplemental sensory feedback, they found movements under FES control to be sometimes imprecise and cumbersome. They are considering the use of auditory or electrotactile feedback to alleviate these problems.

### B. *Visual Prostheses Utilizing Electrotactile Stimulation*

A visual prosthesis for the blind, displaying visual information via the electrocutaneous techniques described herein, has been reported by Collins [10]. A two-dimensional matrix comprising 1024 stimulators in a 32 × 32 array was mounted around the abdomen, chest and lateral surfaces of the trunk. Pulse width modulation was employed to control the intensity of the sensation, corresponding to the luminance of a video image. The prosthesis showed promise as a mobility aid, and a limited amount of visual pattern recognition was possible. It was difficult, however, to maintain such a large array in stable contact with the skin. Stimulators at the edges of the array were often tipped on edge and uncomfortable, and pulse width modulation was found to be a less reliable way of controlling intensity.

### C. *Auditory Prostheses*

Saunders has reported ongoing experiments [35] in which profoundly deaf children are taught to monitor their own voices, recognize environmental sounds, and identify a small number of spoken words and phrases, by means of an electrotactile display of acoustic information via a belt of stimulators worn around the abdomen. (See Fig. 3.)

An audio signal, derived from a microphone and equalizing preamplifier, is sent to 32, active IC, low-power bandpass filters, covering the frequency range from 80 to 8000 Hz at approximate one-third octave intervals. The filter outputs are led to 32 stimulating electrodes mounted on a one-inch wide elastic belt. Interelectrode spacing is 12 mm o.c.; the electrodes are constructed from gold-plated printed circuit boards and gold-plated rivets; and the 4 × 7 mm active center is surrounded by a 1 mm insulating annulus and a 12 × 25 mm ground plane. Prior to application, the skin is moistened with warm water, and no further attention is required.

The individual quantal stimulus consists of a biphasic constant current pulse of 10 $\mu$s duration, with a compliance of

Fig. 3. A classroom model of a tactile sensory aid for profoundly deaf children. Sounds are displayed as touch patterns on a belt worn around the child's abdomen. The patterns corresponding to environmental sounds, the speech of others, and the child's own voice must be learned, much like acquiring a second language.

50 V sufficient to generate 10 mA pulses at a skin impedance of 5K Ω. The overall current amplitude is controlled by a single user-operated potentiometer, regulating the amount of charge per pulse and therefore the overall intensity of the display. The number of quantal pulses per unit time modulates the perceived intensity at a given stimulator, which in turn varies according to the energy present in a given bandwidth.

The electrotactile sensation produced by this aid has been found to be acceptable to deaf children in a classroom setting, and the results of continued training and practice with the aid are encouraging. The incorporation of electrotactile techniques has made possible for the first time a high-resolution tactile aid which is power-conserving, lightweight, and efficient, and which will ultimately be implemented as a miniaturized, completely wearable sensory aid.

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