Electrocutaneous Communication in a Guide Dog Robot (MELDOG)

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***Abstract*—Two main problems to be solved in designing truly effective mobility aids for the blind are: 1) to determine what kinds and how many pieces of information are necessary and/or sufficient to mobilize humans, and 2) to establish the optimal coding and display method of the acquired information.**

If a robot is to be designed which can independently travel from one place to another, using a city map with specific landmarks stored in its memory and obstacle information gathered by the sensors on board the robot, these pieces of information acquired and used by the robot provide one solution to the first problem.

In this paper, a guide dog robot (MELDOG) is described which approaches the first problem in this manner. MELDOG intends to enhance mobility aids for the blind by providing them with the functions of guide dogs, i.e., obedience in navigating its blind master, intelligent disobedience in detecting and avoiding obstacles in his/her path, and companionship in communicating between the master and the robot.

For the second problem of displaying the information acquired, the electrocutaneous communication systems being developed, based on the fundamental experiments on electrocutaneous stimulation, are explained. These include constant pulse energy circuits which keep the perceived sensational magnitude constant despite the change of the skin impedance, and two-dimensional phantom sensations which reduce the number of the electrodes used.

INTRODUCTION

INDEPENDENT travel or mobility is one of the strongest desires of the more than 340 000 blind or severely visually impaired individuals in Japan.

Little theoretical research has been done on the processes of mobility, i.e., the necessary and/or sufficient pieces of information about the surroundings that enable normal human mobility. Mann broke down mobility processes into three functions: 1) the blind person's next step, 2) his/her directional orientation, and 3) his/her navigation along reasonably long travel paths on both familiar and unfamiliar terrain [1].

Although ideal mobility aids for the blind should have these three functions, existing mobility devices at the present stage of significant evaluation, namely, the Pathsounder [2], the Sonic Glasses [3], the Laser Cane [4], [5], the Mowat Sensor [6], and the Nottingham Obstacle Detector [7], have only functions 1 and 2. The information processing system employed in functions 1 and 2 is usually very simple and crude so that the blind user must concentrate on the devices, resulting in the fatigue of the user or loss of other information which otherwise might be obtained through the remaining senses.

With the advent of recent electronic technological breakthroughs, it is becoming possible to design more intelligent mobility aids for the blind which combine the above three functions with the enhancement of functions 1 and/or 2, by increasing the information processed by the device or the machine [8], [9]. Some of these devices warn only if the blind persons are in danger, thereby not distracting the attention of the blind traveler from other potential cues through their remaining senses. This design concept of supervisory systems [8] is very similar to traveling with a guide dog (seeing-eye).

A six-year project dubbed MELDOG was started at the Mechanical Engineering Laboratory, in fiscal year 1977, to enhance mobility aids for the blind by providing them with the functions of guide dogs, i.e., obedience in navigating or guiding its blind master, intelligent disobedience in detecting and avoiding obstacles in his/her path, and well-organized man–machine (animal) communication which does not interfere with his/her remaining senses.

In this paper, the design concept of MELDOG and some experimental results with the test hardware (Mark I, II, III, and IV) are discussed. Special emphasis is put on the electrocutaneous communication system which is or will be used in MELDOG. Experimental results of the electrocutaneous communication per se, which are the basis of the system design, are also discussed.

GUIDE DOG ROBOT

There are two main problems to be solved in designing truly effective mobility aids for the blind. The first (I) is theoretical understanding of human mobility (normal and blind) analogous to the elucidation that linguistics reading and speech research provide on human language and communication, e.g., to determine what kinds and how many pieces of information are necessary and/or sufficient to mobilize humans. The second main problem (II) is to establish the optimal coding and display method of the thus acquired information.

In an effort to gain more understanding of the process of mobility, whether of a blind or sighted traveler, a mathematical model of the mobility process was proposed [10]. The analytical model proposed treated a traveler as a processor of information which was gathered with a probe, was used for path traversal, and was lost in the uncertainties of memory, decay, and disorientation. The basis for the model is an equation that balances information gains and losses.

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Another approach may be through analysis by synthesis. If a robot can independently travel from one place to another, using a city map with specified landmarks stored in the memory of the robot and obstacle information gathered by the sensors on board the robot, these pieces of information acquired and used by the robot are one solution to the first problem (I). MELDOG approaches this problem of mobility in this manner.

Functionally, mobility can be broken down into 1) the next step, 2) orientation, and 3) navigation. These functions are generally thought of in that order. When we observe, however, the training process of guide dogs, these functions are in reverse order; i.e., obedience, which corresponds to the guidance or navigation of the blind master, comes first, and then intelligent disobedience, which corresponds to obstacle detection and avoidance, comes next. In the design of the guide dog robot, function 3 was considered first.

The fundamental database of the robot is its navigation map stored in the auxiliary memory, e.g., cassette tapes, and transferred into the main memory of the robot when in use. The navigation map consists of information about intersections, i.e., names and types of intersections; distance between two adjacent intersections; and orientation to the adjacent intersections. In other words, for the first-order approximation we adopted John Kenneth Dupress's travel style of identification of the unobstructed tunnel that would permit the traveler's safe transport through the surrounding space [1]. Information as to the kind of shops which exist along the street is not essential. Only the relations among intersections are essential. The navigation map appears as tunnels that connect intersections. This connection map is represented as an automaton [8].

The next step the robot should take is to identify the real intersection as specified on the map and correct its position and orientation so that it can travel farther. In order to do so, specific landmarks are chosen for each intersection. In the initial phase (from 1977 to 1982), white painted lines on the streets with a length of about 2 m and a width of 0.15 m, were adopted as the landmarks. These marks had to be set at every crossing at this stage of development. The automaton representation map for the robot could be automatically produced by an off-line computer from an ordinary map using picture processing techniques. Landmark laying instructions which would be used to place the landmarks on the streets could be provided at the same time.

At the second stage (from 1983 to the present), preregistered natural landmarks such as poles and walls are being used as markers for the correction of the robot's position and orientation. However, the navigation method is fundamentally the same.

With this predetermined map on board the robot, the guide dog robot 1) navigates, 2) orients, and 3) solves the next step as follows (Fig. 1). First, in principle, the master takes the initiative. The master orders the robot by control switches through a wired link. The robot precedes the master and stops on each landmark, which is set or previously registered at every crossing, and waits for the master's next order (right, left, straight, or stop) and obeys it. If the master does not know the area and wants full automatic guidance, all he has to do is assign the starting code and the destination code. The robot determines whether there is a route to reach the destination. If plural routes exist, it chooses an optimal route and guides the master accordingly [11] [landmark subsystem of Fig. 1 (a)]. Second, in normal travel, the speed of the robot is controlled so that it coincides with that of the master's walk. Thus, if the master walks slowly or quickly, the robot also moves slowly or quickly, keeping the distance between them almost constant. As long as the master is considered to be safe by the robot, he is not warned so that he may concentrate on his remaining senses and his own decision. Only when he fails to detect an obstacle or is outside the safety zone is he warned by the robot [12] [man–machine communication subsystem of Fig. 1(a)]. Third, when the robot detects a dangerous situation on the road, it no longer obeys the master's command but gives him a warning. If the obstacle is moving toward the master, it stops and alerts the moving object and the master. If the obstacle is moving in the same direction but slower than the master, it asks the master to reduce speed to follow the preceding object, probably a human traveler. If something is crossing in front of the robot, the robot waits until it passes. If it detects obstacles which do not move, it tries to determine if it is possible to find space (or tunnel) that will permit the safe transport of the master around the obstacle. If space exists, it guides the master safely around the obstacle to the next landmark. If no space exists, it tries to find a new route to the destination without using that path [8] [obstacle detection subsystem of Fig. 1(a)].

TACTILE COMMUNICATION SYSTEM IN MELDOG

The second problem in designing optimal mobility aids is to find the best coding and the best display method for the acquired information. Mann proposed the "mobility simulator" to study the systematic design of mobility aids as early as 1965 [13], and Brabyn has reported [14] on a scheme of the realization. This system is now in operation at least at M.I.T. and the Mechanical Engineering Laboratory (MEL) in Japan [15].

The robot MELDOG has the capability to study mobility information systematically if it acquires sufficient information and this information is passed to the master via the man–machine communication subsystem. The choice of the communication methods (coding and display scheme) is versatile. The test MELDOG hardware carries a versatile microprocessor (LSI 11/02) which can be used as a display device emulator. It can also measure the movement of the master using the ultrasonic sensors on board the robot to determine the relative position and orientation of the master. Thus, we can determine the optimal display scheme of the acquired knowledge using the master's behavior as its criterion of optimality.

We have been working on the use of the electrocutaneous communication systems because of their potential

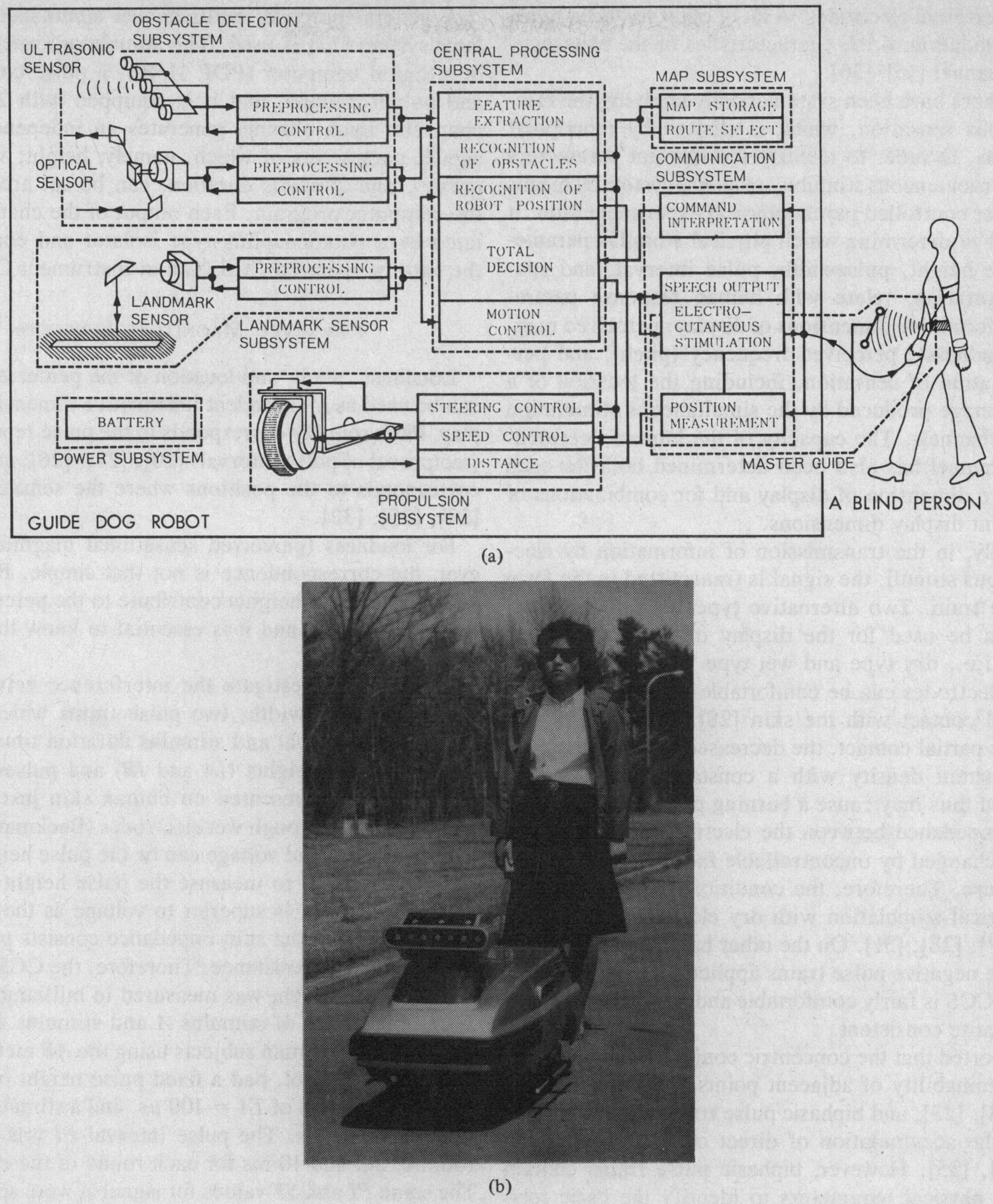

Fig. 1. (a) Schematic diagram of the guide dog robot system (MELDOG). (b) General view of the guide dog robot test MELDOG Mark IV hardware.

advantages of small size, light weight, low power consumption, silence, and fixability to the human anatomy as compared to vibrotactile display systems. As for speech output, we plan to use it to inform both the master and other pedestrians and/or drivers passing by. If the master wants private communication with the robot, electrocutaneous communication would be preferred since it will not interfere with the most reliable remaining sense, i.e., hearing. An electrocutaneous communication system is also preferable from the standpoint of keeping the blind master's privacy since no other person can know what is being communicated from the robot to the master.

FUNDAMENTAL EXPERIMENTS ON ELECTROCUTANEOUS STIMULATION

Various investigations of electrocutaneous communication systems, and studies on human characteristics to electrocutaneous stimulation, have been conducted for application to the various fields, including the augmentation, substitution, and replacement of human sensory functions using the cutaneous sense as an auxiliary or alternative sensory communication channel from devices/machines to humans. Typical systems are, quite simply, informative displays that utilize the skin's sensitivity as an input channel to the human by passing small currents through the skin from external electrodes. A lot of effort has been made for the elucidation of the characteristics of the electrocutaneous channel [16]–[36].

The authors have been systematically studying the electrocutaneous sensation, which is elicited by monopolar pulse trains, in order to identify the relevant parameters of the electrocutaneous stimulus–response system by means of computer controlled psychophysical experimentation. It is an effort to determine which physical stimulus parameters: pulse height, pulsewidth, pulse interval, and spatial configuration, relate with human response parameters or informative dimensions of display: perceived magnitude (loudness), perceived frequency (pitch), and perceived location of sensation, including the location of a phantom image produced by the simultaneous stimulation of plural channels. The capacity of the human electrocutaneous channel has also been determined both for each informative dimension of display and for combinations of independent display dimensions.

Generally, in the transmission of information by electrocutaneous stimuli, the signal is transmitted in the form of a pulse train. Two alternative types of external electrodes can be used for the display of electrocutaneous stimulus, i.e., dry type and wet type. It has been found that dry electrodes can be comfortable if they make complete areal contact with the skin [28]. However, if they make only partial contact, the decreased area can result in greater current density with a constant current source (CCS), and thus may cause a burning pain sensation [31], and the impedance between the electrodes and skin can easily be changed by uncontrollable factors like sweating and pressure. Therefore, the conditions for the comfortable electrical stimulation with dry electrodes have been sought [17], [28], [31]. On the other hand, sensation elicited by the negative pulse trains applied through wet electrodes by CCS is fairly comfortable and the cutaneous sensation is quite consistent.

It is reported that the concentric configuration increases the discriminability of adjacent points and limits current spread [18], [23], and biphasic pulse trains should be used to avoid the accumulation of direct current components [28], [29], [35]. However, biphasic pulse trains contain too many physical parameters to identify the basic relationship between stimulus and response. Preliminary experiments have revealed that an arrangement of three wet electrodes—the outer two of which are connected and used as a common, and a negative pulse is applied to the center electrode—has an equivalent effect as the concentric configuration. The sensation elicited by the negative-going monophasic pulse trains applied by this arrangement, with pulsewidth around 100 μs and pulse interval longer than 10 ms, is fairly comfortable and the cutaneous sensation is quite distinct.

Based on the above consideration, we have selected wet electrodes, e.g., Beckman type (ϕ 8 mm), and negative-going monophasic pulse trains driven by CCS with a three electrode arrangement suitable to study the relation between the physical parameters and informative dimensions.

A general purpose, multichannel simultaneous stimulator system [37] is used. The stimulator system consists of a digital computer (PDP 11/40), a pulse control unit, and output circuits, and it is equipped with 256 output channels. Each channel generates an independent pulse signal, parameters of which, namely, height, width, frequency, and stimulus duration, can be set arbitrarily by the computer program. Each output of the channel is isolated by a photocoupling-type isolator and connected to the battery-operated CCS (San-ei Instrument Co., 5361).

PERCEIVED MAGNITUDE SENSATION

Loudness, pitch, and location of the perceived stimulus can be used as independent informative dimensions of display. Pitch roughly corresponds to the pulse repetition rate (reciprocal of pulse interval) [20], [24], [26], and location corresponds to the positions where the sensation occurs [23], [25], [32].

For loudness (perceived sensational magnitude), however, the correspondence is not that simple. Both pulsewidths and pulse heights contribute to the perceived magnitude sensation, and it is essential to know their mutual effects.

In order to investigate the interference between pulse height and pulsewidth, two pulse trains which have the same pulse interval and stimulus duration time, but have different pulse heights (IA and IB) and pulsewidths (TA and TB), were presented on human skin just above the triceps brachii through wet electrodes (Beckman ϕ 8 mm). Both current and voltage can be the pulse height parameter, but in order to measure the pulse height and width precisely, current is superior to voltage as the height parameter because the skin impedance consists primarily of capacitance and resistance. Therefore, the CCS was used, and the pulse height was measured in milliamperes.

The difference of stimulus A and stimulus B in Fig. 2 was judged by human subjects using the AB method. Stimulus A, the control, had a fixed pulse height of $IA = 4.7$ mA, a pulsewidth of $TA = 100$ μs, and a stimulus duration time of $ST = 2$ s. The pulse interval PI was selected at 100, 50, 20, and 10 ms for each round of the experiment. The same PI and ST values for signal A were set for signal B. Each of the seven values of signal B's pulsewidths (i.e., $TB = 150$, 200, 250, 300, 400, 800, or 1600 μs) was assigned to signal B. For each of the TB values, one of several arbitrary current values was selected for IB to complete signal B. All possible combinations of pulse height IB and pulsewidth TB were selected, and each stimulus B thus formed was compared to the control stimulus A. Either stimulus A or B was presented 50 times to human subjects, and the subjects judged which of the stimuli was presented. The information transmitted per stimulus was measured by means of the AB method, and was used as the measure of the distinctness of the difference of the two stimuli. Fig. 3 shows an example of the experimental results. Each mark in the figure is the result calculated from 50 judgments, and the marks ⊙, ⟁, ⊡, ◈, ✳, ⧖, and · indicate the results for TB of 150, 200, 250, 300, 400, 800,

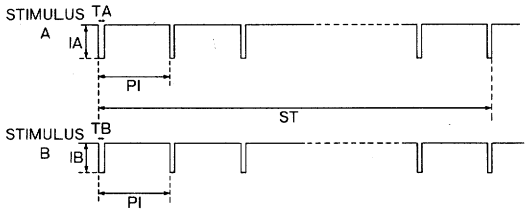

Fig. 2. Experimental condition of the perceived magnitude sensation. $ST = 2$ s, $TA0 = T1 = 100$ μs, $IA = 4.7$ mA, and $PI = 100$, 50, 20, and 10 ms.

and 1600 μs, respectively. The abscissa represents *IB*, and the ordinate represents the information of the stimulus difference transmitted per stimulus in bits.

The discrimination curve for each pulsewidth is v-shaped, which means that there is a certain condition that makes the discrimination of the two stimuli impossible. Specifically, *IB* values at the bottom of the v-shaped discrimination curves for several *TB* values represent the condition under which stimulus *B* has the same perceived magnitude as that of stimulus *A*.

In order to quantitatively define this condition, the *IB* values that give minimum discrimination for various *TB* values are plotted using a log–log scale as a function of *TB* (Fig. 4). The marks ⊙, ×, ◬, and ⊡ are the results for $PI = 100$, 50, 20, and 10 ms, respectively, and each mark is the average value for three subjects. The solid line is the least-mean square approximation of the results when the results with a *TB* of less than 1 ms are considered. The gradient of the line is -0.5.

Thus, when $\log IB = -0.5 \log TB + k'$ holds, i.e., $IB = kTB^{-0.5}$, or $IB^2TB = c = IA^2TA$, the perceived magnitude of stimulus *B* is the same as that of stimulus *A*. As impedance of the skin and tissue is thought to remain constant during such a relatively short period as one round of the experiment, the condition of the equal perceived sensational magnitude, i.e., $IA^2TA = IB^2TB$, can be interpreted as follows:

$$ZIA^2TA = ZIB^2TB \tag{1}$$

where Z is the impedance of the tissue and the electrode.

The quantity ZI^2T is an energy of the pulse. The condition can be expressed as follows:

$$\int_0^T Z\,IA(t)^2dt = \int_0^T Z\,IB\,(t)^2dt, \qquad T < 1 \text{ ms}. \tag{2}$$

The threshold current of minimum sensation for various pulsewidths is measured by using the *AB* method.[1] Fig. 5 shows the result, where threshold current is measured as a function of pulsewidth T. The shape of the curve can be represented by the equation $I = a + b/T$, as has traditionally been the case. These data are replotted by using a log–log scale as in Fig. 6. When we consider data with pulsewidths of less than 1 ms, the relation can be approximated by a straight line with the gradient of -0.6. This also suggests that the threshold of minimum sensation is obtained when the quantity I^2T has the same constant value, or constant energy.

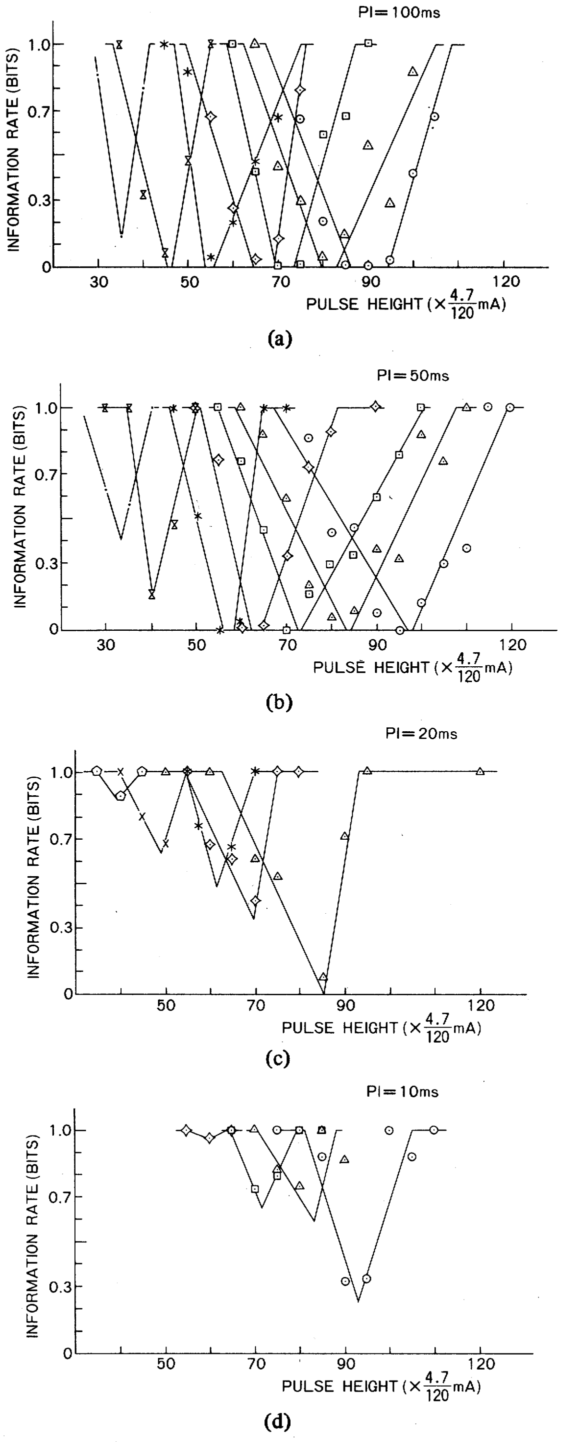

Fig. 3. (a) Detectable difference between stimuli *A* and *B* of Fig. 2 for $PI = 100$ ms. The marks ⊙, ◬, ⊡, ◈, ✳, ⧖, and · indicate the results for *TB* of 150, 200, 250, 300, 400, 800, and 1600 μs, respectively. (b) Detectable difference between stimuli *A* and *B* of Fig. 2 for $PI = 50$ ms. (c) Detectable difference between stimuli *A* and *B* of Fig. 2 for $PI = 20$ ms. The symbols × and ⬠ indicate the results for *TB* of 600 and 900 μs, respectively. (d) Detectable difference between stimuli *A* and *B* of Fig. 2 for $PI = 10$ ms.

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[1] Preliminary threshold experiments revealed that the minimum sensation threshold current increass with time (number of hours) when a constant current source (CCS) is used, and that threshold voltage of minimum sensation decreases with time when a constant voltage source (CVS) is used. In other words, if we apply the same pulse train with the same current height after a 2 or 3 h intermission, the perceived magnitude decreases, while if we apply the same pulse with the same voltage height, the perceived magnitude increases. These phenomena suggested the possibility of the impedance change of sensory tissue and the effectiveness of power or energy control of the pulse.

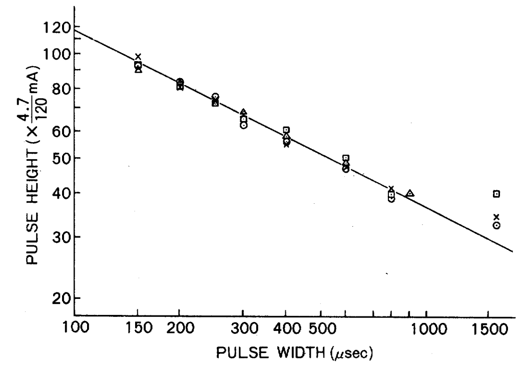


Fig. 4. Plot of the average *IB* values that indicate the minimum discrimination using a log-log scale as a function of *TB*. The marks ⊙, ×, ◬, and ⊡ are the results for *PI* = 100, 50, 20, and 10 ms, respectively.

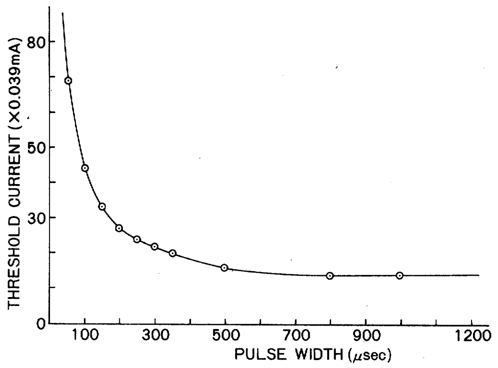


Fig. 5. Threshold current for minimum sensation as a function of pulsewidth.

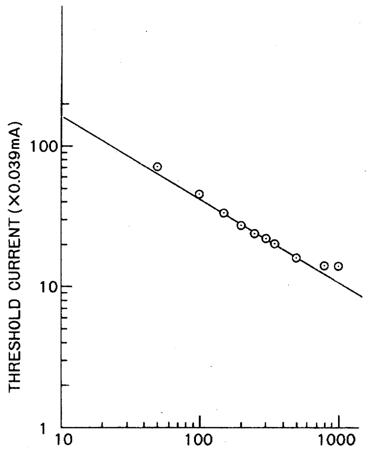


Fig. 6. Replot of the result of Fig. 5 on a log-log scale.

These findings [37] support the view Green [16] had that energy is the most relevant parameter for the absolute threshold. That pulse energy is the most relevant parameter for the perceived magnitude sensation, when the pulsewidth is less than 1 ms, strongly suggests that the fine structure of the pulse within 1 ms would not greatly affect the perceived magnitude sensation, but its total energy within 1 ms would.

CHANNEL CAPACITY

The channel capacity of a magnitude information transmission system was measured [38] using the constant energy stimulator designed. The channel capacity was calculated using the following formula:

$$R = \log_2 \int_{\min E}^{\max E} \frac{1}{\Delta E(E)}\, dE \qquad (3)$$

where $\Delta E(E)$ is the just-noticeable difference (jnd) measured as a function of energy E, and min$E$ and max$E$ are the energies that give the minimum and maximum thresholds of perceived magnitude, respectively. The channel capacity is 3.0–4.0 bits per symbol.

The channel capacity of a pitch information transmission system was calculated using jnd's measured as a function of frequency, and the maximum information transmission rate was estimated from the results of forced choice tests. The channel capacity is 2.5–3.0 bits per symbol in the frequency range of 10–100 pps. The maximum information transmission rate is about 2.1 bits per symbol, which is a little larger than the corresponding value for the magnitude dimension of 1.7–1.9.

Two-variable display systems where two aspects of the stimulation, i.e., perceived magnitude and pitch sensation, were used simultaneously were studied [39]. Fig.7 shows a typical two-variable electrocutaneous information transmission system which transmits two types of information. The problem with this type of display is that these two variables are not independent, i.e., magnitude sensation is affected by the change of frequency and vice versa. The solution to this problem is to change pulse energies along the equal magnitude sensation curves and pulse repetition rates along the equal pitch sensation curves. The latter effect, however, can be neglected. Fig. 8 shows the equal magnitude sensation curves for electrocutaneous stimulation. Each dotted line in the figure indicates the difference limen from the corresponding equal magnitude curve. The channel capacity of a two-variable electrocutaneous information transmission system is estimated from the number of crosspoints of equal magnitude curves and equal pitch sensation lines. Its value ranges from 5.6 to 6.0 bits per symbol. The maximum information transmission rate estimated from the forced choice test is 2.7–3.2 bits per symbol [39].

It is found that the display that utilizes the cutaneous phantom sensation yields interesting results [20], [30]. Two equally loud electrocutaneous stimuli, simultaneously presented to adjacent locations on the skin, are not felt separately, but rather, combine to form a sensation midway between the two electrodes. This is also the case of binaural sound localization in hearing and the vibration phantom sensation on the skin [18]. This phantom location can be controlled by relative magnitudes of the two stimuli and by the time delay between them [40]. Thus, the number of electrodes can be reduced by using this phantom

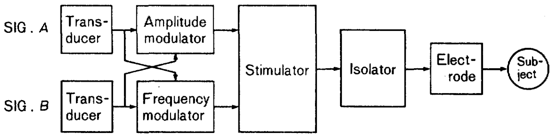


Fig. 7. Two-variable information transmission system.

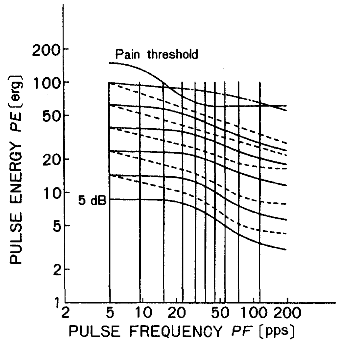


Fig. 8. Equal perceived magnitude curves as a function of frequency.

sensation display in comparison to a display requiring a discrete electrode for each position desired. The channel capacity of a location information transmission system using the phantom sensation depends on the distance *d* between the two electrodes, and ranges from 2.0 to 3.0 bits for *d* between 50 and 150 mm [40].

Fig. 9 shows the two-dimensional extension of this phantom display. The phantom image produced by three sets of electrodes [Fig. 9(a)] is the basis of the two-dimensional display because phantom display areas can easily be extended as in Fig. 9(b). Fig. 10 shows the experimental arrangement used. The channel capacity of a two-dimensional phantom location information transmission channel is estimated to be about 4 bits per symbol from the jnd's of the perceived location. The maximum information transmission rate is about 2.8 bits per symbol [41].

Master Guide

In the MELDOG system, the location of the master is measured by the robot, in real time, by the triangulation among the ultrasonic oscillator put on the belt of the master and two receivers on board the robot. The speed of the robot is controlled by the walking speed of the master. The safety zone is set behind the robot, in which the master is permitted to walk. When he is outside the zone he is warned by the robot,while he receives nothing when he is safe. When the orientation of the master is not appropriate, the master guide detects the condition and informs the master. These signals are transmitted through a wired link and presented to the master in the form of electrocutaneous stimulation on the skin. Two sets of Ag–AgCl wet electrodes are located on the skin of both brachia. The signals used are pulse trains with a pulsewidth of about 100 μs, the energy of which is controlled by the circuit shown in Fig. 11. The voltage across the electrodes and the pulse current are measured, multiplied, and integrated to be compared to the desired pulse energy. The monostable multivibrator is reset immediately to control the energy of the pulse by controlling the pulsewidth. The pulse current can be set automatically to a value proportional to the desired energy so that the controlled pulsewidth is always about 100 μs.

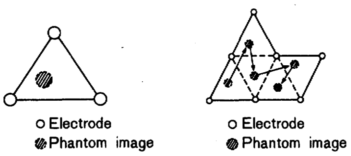


Fig. 9. (a) Electrode arrangement of the fundamental two-dimensional phantom sensation. (b) Extension of stimulation area.

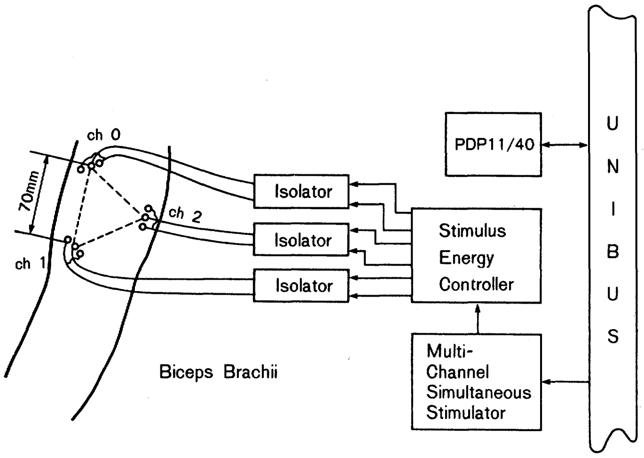


Fig. 10. Stimulation system of the two-dimensional phantom sensation.

In the test MELDOG Mark I hardware, the repetition rate of the pulse train was set at 100 pps for the usual warning that the master was outside the safety zone, and 10 pps for the warning that the master's orientation was inappropriate. For example, the signal presented to the right arm with 100 pps means the master should step to the right to come back to the safety zone, and with 10 pps means (s)he should turn his/her body counterclockwise to correct his/her orientation [12].

Conclusions

A method was proposed to guide a blind individual using a robot which processes both the information stored in the memory of the robot and the obstacle information acquired by the sensors on board the robot, and by passing the acquired information from the robot to the blind master.

Test hardwares (MELDOG Mark I, II, III, and IV) were constructed and the feasibility of the method was demonstrated.

Electrocutaneous communication systems were proposed, and fundamental experiments of information transmission were reported, and their applicability to the communication system of MELDOG was demonstrated.

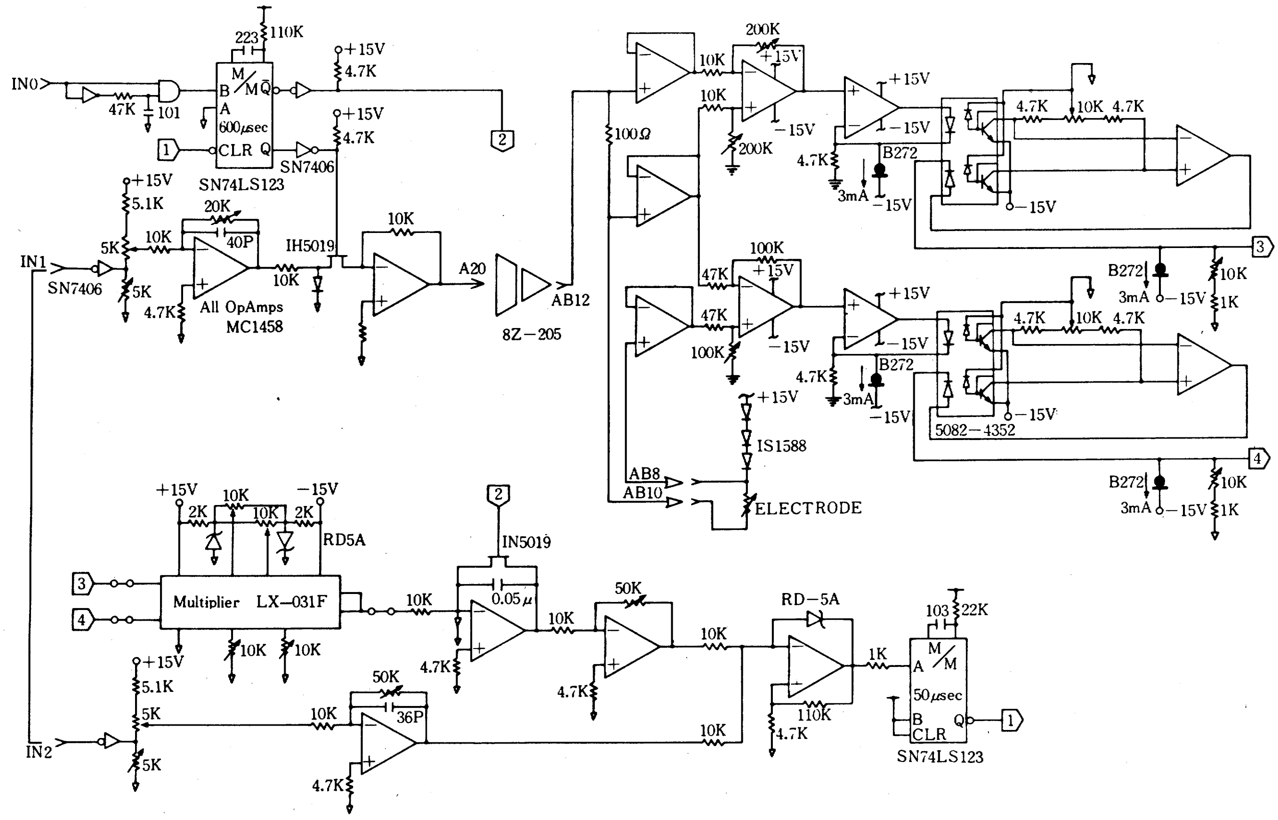


Fig. 11. Constant energy source for electrocutaneous stimulation.

The remaining problem includes the choice of sensory display of the navigation information acquired by the robot appropriate for presentation to the remaining exterior receptive senses of the blind individual.

Quantitative comparison of display schemes is inevitable for optimal choice of the appropriate display [15]. The guide dog robot system can also be used as an instrument for this quantitative comparison of display schemes.

REFERENCES

[1] R. W. Mann, "Technology and human rehabilitation: Prostheses for sensory rehabilitation and/or sensory substitution," *Adv. Biomed. Eng.* vol. 4, pp. 209-353, 1974.

[2] L. Russell, "Evaluation of mobility aids for the blind, Pathsounder Travel Aid evaluation," Nat. Acad. Eng., Washington, DC, 1971.

[3] L. Kay, "Sonic glasses for the blind—Presentation of evaluation data," Amer. Found. Blind, Res. Bull. 26, 1973.

[4] L. W. Farmer, J. M. Benjamin, Jr., D. C. Cooper, W. R. Ekstrom, and J. J. Whitehead, "A teaching guide for the C-5 Laser Cane: An electronic mobility aid for the blind," Kalamazoo College Educ., Western Michigan University, Kalamazoo, 1975, p. 78.

[5] G. Jansson and B. Schenkman, "The effect of the range of a laser cane on the detection of objects by the blind," Univ. Uppsala, Sweden, Rep. 211, 1977.

[6] D. L. Morrissette, G. L. Goodrich, and J. J. Hennessey, "A follow-up study of the Mowat Sensor's applications, frequency of use, and maintenance reliability," *J. Vis. Impair. Blindness*, vol. 75, pp. 244-247, 1981.

[7] A. G. Dodds, J. D. Armstrong, and C. A. Shingledecker, "The Nottingham Obstacle Detector: development and evaluation," *J. Vis. Impair. Blindness*, vol. 75, pp. 203-209, 1982.

[8] S. Tachi, K. Komoriya, K. Tanie, T. Ohno, and M. Abe, "Guide dog robot—Feasibility experiments with MELDOG MARK III," in *Proc. 11th Int. Symp. Ind. Robots*, Tokyo, Japan, Oct. 1981, pp. 95-102.

[9] C. C. Collins and M. E. Deering, "A microcomputer based blind mobility aid," in *Proc. 6th Annu. Conf. IEEE Eng. Med. Biol. Soc.*, Los Angeles, CA, Sept. 1985, pp. 52-55.

[10] A. Krigman, "A mathematical model of blind mobility," Eng. Proj. Lab., Dep. Mech. Eng., Mass. Inst. Technol., Cambridge, MA, DSR Rep. 70249-2, 1967.

[11] S. Tachi and K. Komoriya, "Guide dog robot," in *Proc. 2nd Int. Conf. Robot. Res.*, Kyoto, Japan, Aug. 1984.

[12] S. Tachi, K. Tanie, K. Komoriya, Y. Hosoda, and M. Abe, "Guide dog robot—Its basic plan and some experiments with MELDOG MARK I," *Bull. Mech. Eng. Lab.* Japan, no. 32, 1978; or *Mech. Mach. Theory*, vol. 16, pp. 21-29, 1981.

[13] R. W. Mann, "The evaluation and simulation of mobility aids for the blind," Amer. Found. Blind, Res. Bull. 11, 1965.

[14] J. A. Brabyn, H. R. Sirisena, and R. S. Clark, "Instrumentation system for blind mobility aids simulation and evaluation," *IEEE Trans. Biomed. Eng.*, vol. BME-25, pp. 556-559, June 1978.

[15] S. Tachi, R. W. Mann, and D. Rowell, "Quantitative comparison of alternative sensory displays for mobility aids for the blind," *IEEE Trans. Biomed. Eng.*, vol. BME-30, pp. 571-577, Sept. 1983.

[16] R. T. Green, "Absolute threshold for electrical shock," *Nature*, vol. 194, pp. 1303-1304, 1962.

[17] R. H. Gibson, "Electrical stimulation of pain and touch systems," *Nature*, vol. 199, pp. 307-308, 1963.

[18] G. von Békésy, *Sensory Inhibition*. Princeton, NJ: Princeton University Press, 1967.

[19] T. W. Beeker, J. During, and A. den Hertog, "Artificial touch in a hand prosthesis," *Med. Biol. Eng.*, vol. 5, pp. 47-49, 1967.

[20] R. H. Gibson, "Electrical stimulation of pain and touch," in *The Skin Senses*, D. R. Kenshalo, Ed. Springfield, IL: Thomas, 1968, pp. 223-261.

[21] P. Bach-y-Rita, C. C. Collins, F. A. Saunders, B. White, and L. Scadden, "Vision substitution by tactile image projection," *Nature*, vol. 221, pp. 963-964, 1969.

[22] G. B. Rollman, "Electrocutaneous stimulation: Psychometric functions and temporal integration," *Perception Psychophys.*, vol. 5, pp. 289-293, 1969.

[23] C. C. Collins, "Tactile television: Mechanical and electrical image projection," *IEEE Trans. Man-Machine Syst.*, vol. MMS-11, no. 1, pp. 65-71, 1970.

[24] I. Kato, M. Kumamoto, S. Tamura, and Y. Tsunekawa, "Human's cognition ability for electric stimulation signals," *Adv. Extern. Contr. Human Extremities* in *Proc. 3rd Int. Symp. Extern. Contr.*, Dubrovnik, Yugoslavia, Aug. 1970, pp. 69–84.

[25] C. C. Collins and F. A. Saunders, "Pictorial display by direct electrical stimulation of the skin," *J. Biomed. Syst.*, vol. 1, pp. 3–16, 1970.

[26] R. E. Prior, "Study of electrocutaneous parameters relevant to dynamic tactual communication systems," Ph.D. dissertation, Dep. Elec. Eng., Univ. Calif., Los Angeles, 1972.

[27] F. A. Saunders, "An electrotactile sound detector for the deaf," *IEEE Trans. Audio Electroacoust.*, vol. AU-21, no. 3, pp. 285–287, 1973.

[28] —, "Electrocutaneous displays," in *Proc. Monterey Conf. Cutaneous Commun. Syst. Dev.*, Geldard, Ed., Monterey, CA, 1973.

[29] D. E. Hardt, "Parameters of electrocutaneous stimulation," M. S. thesis, Dep. Mech. Eng., Mass. Inst. Technol., Cambridge, MA, 1974.

[30] K. Sundstrom, "An investigation of cutaneous phantom sensations produced via electrical stimulation," M.S. thesis, Dep. Mech. Eng., Mass. Inst. Technol., Cambridge, MA, 1974.

[31] J. L. Mason and N. A. M. Mackay, "Pain sensation associated with electrocutaneous stimulation," *IEEE Trans. Biomed. Eng.*, vol. BME-23, pp. 405–409, 1976.

[32] M. Solomonow, L. Raplee, and J. Lyman, "Electrotactile two point discrimination as a function of frequency, pulse width and pulse time delay," *Ann. Biomed. Eng.*, vol. 6, pp. 117–125, 1978.

[33] R. Butikofer and P. D. Lawrence, "Electrocutaneous nerve stimulation—1: Model and experiment," *IEEE Trans. Biomed. Eng.*, vol. BME-25, pp. 526–531, 1978. "Electrocutaneous nerve stimulation—II: Stimulus waveform selection," *IEEE Trans. Biomed. Eng.*, vol. BME-26, pp. 69–75, 1979.

[34] A. Y. J. Szeto, R. E. Prior, and J. Lyman, "Electrocutaneous tracking: A methodology for evaluating sensory feedback codes," *IEEE Trans. Biomed. Eng.*, vol. BME-26, pp. 47–49, Jan. 1979.

[35] A. Y. J. Szeto and F. A. Saunders, "Electrocutaneous stimulation for sensory communication in rehabilitative engineering," *IEEE Trans. Biomed. Eng.*, vol. BME-29, pp. 300–308, Apr. 1982.

[36] M. Solomonow, "Electrotactile sensation and pain thresholds and ranges as a function of body site, laterality and sex," in *Proc. 4th Annu. Conf. IEEE Eng. Med. Biol. Soc.*, Philadelphia, PA, Sept. 1982, pp. 329–332.

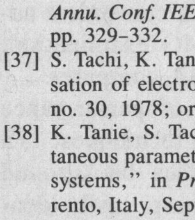
[37] S. Tachi, K. Tanie, and M. Abe, "Experiments on the magnitude sensation of electrocutaneous stimuli," Bull. Mech. Eng. Lab., Japan, no. 30, 1978; or *Japan. J. Biomed. Eng.*, vol. 15, pp. 315–320, 1977.

[38] K. Tanie, S. Tachi, K. Komoriya, and M. Abe, "Study of electrocutaneous parameters for application to dynamic tactual communication systems," in *Proc. 1st Mediterranean Conf. Med. Biol. Eng.*, Sorrento, Italy, Sept. 1977.

[39] —, "Study on two-variable electrocutaneous communication," in *Proc. 7th Int. Symp. Extern. Contr. Human Extremities*, Dubrovnik, Yugoslavia, 1981.

[40] S. Tachi, K. Tanie, K. Komoriya, and M. Abe, "Information transmission by electrocutaneous phantom sensation," summary of papers on general fuzzy problems, *Working Group Fuzzy Syst.*, Tokyo Inst. Tech., Tokyo, Japan, vol. 4, 1978, pp. 10–15.

[41] S. Tachi, K. Tanie, K. Komoriya, K. Asaba, Y. Tomita, and M. Abe, "Information transmission by two-dimensional electrocutaneous phantom sensation," in *Proc. 8th Annu. Northeast Bioeng. Conf.*, Cambridge, MA, Mar. 1980, pp. 258–262.



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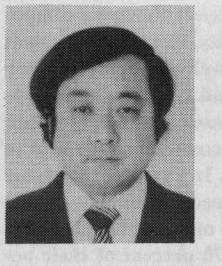

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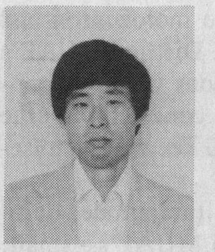

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