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Flexible EMG Sensor Array for Haptic Interface

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Abstract: In our previous studies, we have proposed a new man-machine interface that detects myoelectric signals on a forearm for estimating finger motions. One of the problems for achieving a high density Electromyography (EMG) sensor array is wiring to each node. We adopted "Two-Dimensional Communication (2DC)" technology for reducing wires. The 2DC enables one to connect sensor nodes embedded in the thin sheet without any wires. In the previous papers, we proposed how to connect sensors to the 2DC sheet without electrical contact. In this paper, we show a new connector which achieves efficient power transmission with microwave through the 2DC sheet. We also show the experimental results of the myoelectric signals with three sensor nodes on the sheet.

Keywords: Haptic Interface, Electromyography, Two Dimensional Communication

1. INTRODUCTION

For haptic interfaces, it is substantial to know finger postures and related forces so as to give effective stimuli. The Cyber glove (Immersion Corp.), for example, can be used to measure finger motions based on the angles of each joint. In our previous studies [1], we focused on the Electromyography (EMG) for estimating the force of fingers. The myoelectric signal is the signal that can be detected when muscle contraction occurs. Thus we can estimate finger forces based on those signals.

There are two important advantages using EMG for haptic interface. Firstly, we can know the fingers' conditions without any constraints on them, because the most muscles relating to the finger motion exist under a forearm. The contractions of those muscles are transmitted through tendons to fingers. Therefore we can sense the finger states without interfering finger operations. Secondly, the myoelectric signals are generated just before the actual motion occurs. For fingers, the motion occurs about 0.1 s later than the detection of the myoelectric signals. We can estimate finger motions earlier than real motions.

In order to know precise conditions of each finger, it is considered to be important that the number of the measurement points should be large enough. Conversely, we can say that the finger motions can be easily estimated provided the sufficient number of sensors is arranged on the forearm. In our previous study [2] we confirmed that the observation of the two dimensional EMG signal patterns on a forearm enabled us to know which finger moved based only on their intensities.

One of the critical issues for achieving a high-density EMG sensor array is wiring. A large number of wires are required when the measurement points increase. This makes system complex and restricts human motions. There are several previous studies which measures dense 2D Electromyography [3] [4], however, the studies assumed some special situations that make wearing complex devices allowable.

In our previous studies, we adopted "Two-Dimensional Communication (2DC)" technology as a substitution for the wires. The 2DC is the technology that enables us to achieve both signal transmission and electric power supply through the sheet with microwave. The configuration of the communication sheet is simple. Two conductive layers sandwich a dielectric layer. When high frequency voltages are applied between the conductive layers, microwaves propagate two dimensionally in the sheet. This achieves both signal transmission and electric power supply. Since any conductive materials including conductive rubbers and fabrics can be used for the conductive layers, the sheet can be flexible and stretchable. Thus the EMG sensors can be arranged on the stretchable wristband as shown in Fig. 1. We can comfortably measure two dimensional pattern of the EMG on the forearm without any constraints in our proposed system.

One of the issues of the 2DC is how to connect sensor nodes to the stretchable and flexible sheet. Imagine that the sensors are connected to the sheet with solder, for example, the connection can be easily destroyed due to the stress concentration when it deforms. The connection should be achieved without electrical contact. For this requirement, we have proposed a connector named "Resonant Proximity Connector (RPC) [2]" that realizes strong coupling to the sheet without an electrical contact. When the length of the electrodes facing to the both conductive layers is designed to be $\lambda/4$ of the microwave, the connector can be seen as if it touches directly to the each conductive layer due to its resonance.

The connector was designed so as to lessen the effects of sheet deformations, however, its performance was considerably influenced by them. An efficient connector independent on the sheet deformation is required. In this research, we show a new configuration of RPC that can receive electricity about ten times larger than our previous one experimentally. The difference to the previous one is that the new one couples with the directional mode of the microwave as well as the isotropic mode. The previous one only can connect to the isotropic mode. Since the resonance of the connector for the directional mode is less dependent on the sheet deformation, the connector can receive practically

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enough electricity.

We show experimental evaluations of the new connector. Then we demonstrate 3ch EMG system using the proposed connector.

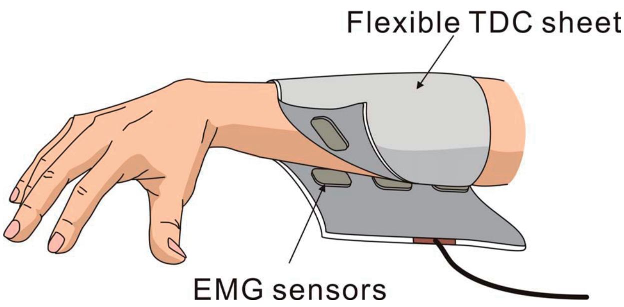

Fig. 1 Schematic diagram of the wristband-shaped electrode array for electromyography

2. RESONANT PROXIMITY CONNECTOR

2.1 Basic principle and previous prototype

In our previous studies, we reported "Resonant Proximity Connector (RPC)" for effective coupling between a sensor and the 2DC sheet [2]. When the total length of the electrode is designed to be $\lambda/4$ of the microwave, the connection to the sheet can be seen as an electrical short owing to its resonance. The resonance condition of the connector has weak dependence on the gap distance between the electrode and the conductive layers of the 2DC sheet. One important feature of the RPC is that it requires no electrical contact between them. Stress concentration is avoided at the connecting point when the sheet deforms. Also the conductive layers can be covered with a thin insulating material so as to prevent them from oxidization and unexpected electrical short.

In our previous prototype the performance of the connector actually depended on the sheet deformation against our concept. The allowed deformation of the thickness for sufficient power transmission was practically smaller than 1 mm. When the deformation was larger than the limit, the reactance component of radiation impedance varied considerably.

2.2 X-shaped RPC

In this paper, we propose a new connector for efficient electricity reception. Figure 2 shows the connector with the four thin electrodes whose lengths are equal to $\lambda/4$ of the microwave respectively. The connector is used as shown in Fig.3 though the other two electrodes which are set perpendicular to the paper are not shown in the figure.

In order to illustrate the principle simply, we use the two-electrodes-based (I-shaped) model as shown in Fig.4. Here, we assume that only the two thin electrodes are set along x-direction. In our previous studies, the electrodes were curved so as to lessen the size of the connector, however, for our new connector the thin electrodes are straightened and arranged in opposite direction. In this case, the connector can couple with two different types of the resonant modes.

The one mode is shown in Fig.5 (a) that is an isotropic mode which is the same one as the former connector used. Since the connector is seen as if it connects directly to the sheet near the base of the electrode, it can couple with the two-dimensionally isotropic mode of the microwave due to its symmetric property. We call this symmetric mode "mode A".

The other mode is a directional mode (Fig 5 (b)). When a microwave propagates along x-direction, the antisymmetric resonance is induced as shown in the figure provided the total length of the two electrodes is equal to the half wavelength. The microwave propagating along the y-direction can not cause resonances. This means the connector has the directivity. We call this directional mode "mode B". One remarkable aspect is that the resonant condition of the mode B has weak dependence on the sheet thickness since the total length of the electrode along x-direction does not change when the sheet deforms. Even when the receivable energy relating to the mode A is lessened due to the sheet deformation, the coupling condition to the mode B suffers less change. The total directivity of the new connector is given as the superposition of the two modes. In our previous RPC, the energy relating to the mode B was not considered.

When the two connectors are arranged so as to be orthogonal to each other and their rectified electric power is connected serially, the electricity can be received omnidirectionally. We fabricated four-electrodes type (X-shaped) of the connector as shown in Fig.2.

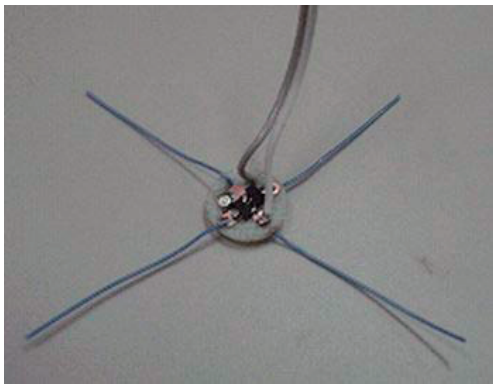

Fig. 2. The new connector having four thin electrodes.

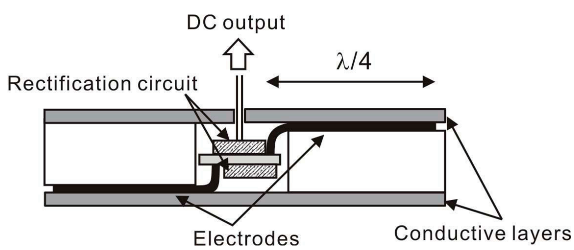

Fig. 3 . Cross section of the new connector.

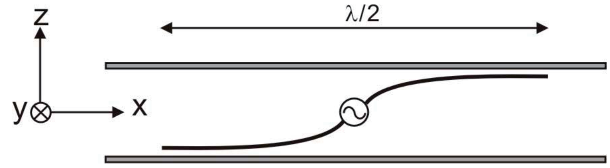

Fig. 4 Two-electrodes-based (I-shaped) model of the new connector

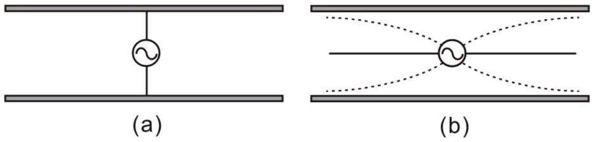

Fig.5 Two modes of the resonance on the connector.
(a): Isotropic mode, (b): Directional mode.

2.3 Experimental evaluations

We experimentally evaluated the principle. Figure 6 shows the textile 2DC sheet made of a conductive fabric. The conductive layers were covered with a non-conductive cloth. There were 18 connection apertures on the surface of the sheet. We measured the receivable electricity at 9 positions shown in the figure with the new connector (X-shaped connector), the one directional connector (I-shaped connector) and our previously proposed connector (curved λ/4 electrodes).

Received electric power is shown in Fig. 7. The horizontal axis shows the position number of the sheet (given in Fig.6) and the vertical axis represents the electricity on a logarithmic scale. We supplied 10 W of a microwave at 2.4 GHz to the sheet. The received electricity with the new X-shaped connector is apparently larger than the other two except the position 6. The averaged receivable electric power was 577 mW with the X-shaped connector, while it was 243 mW with the I-shaped connector and it was 8 mW with our previous one. The coupling to the directional mode (mode B) is effective for efficient power reception.

Figure 8 shows the variation of the received electricity depending on the sheet deformation. We pressed the sheet above the connector with a flat surface at a single position. The horizontal axis shows the compressed depth from the stationary position and the vertical axis indicates the received electricity on a logarithmic scale. The averaged received electricity with the new connector was 256 mW, while it was 23.6 mW with the previous one. Regarding the ratio *R* of the variance to the received power, *R* of the X-shaped connector is smaller than that of the previous one.

Fig.6 Textile 2DC sheet. Each number denotes the position.

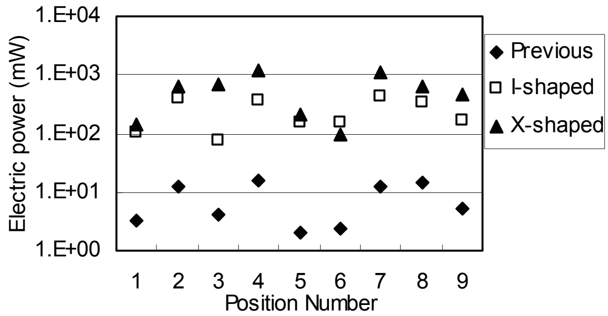

Fig.7 Received electric power with three types of connectors.

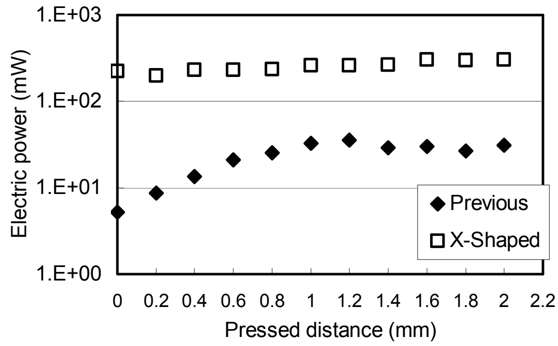

Fig.8 Received electric power depending on the sheet deformation.

3. EMG MEASUREMENT

3.1 Time division multiplexing circuit

For multiple sensor integration, we adopt time division multiplexing (TDM) method. Multi channel data transmission and power supply are achieved alternately in the same sheet. You can use many other protocols that assure multi channel communication such as the frequency division multiplexing (FDM), the code division multiple access (CDMA) and etc., we adopted TDM from the easiness of implementation.

We fabricated low-power consuming circuit for achieving TDM on the 2DC sheet. In our prototype, the whole size of the circuit (EMG sensing unit, PWM modulator, electrical power regulator etc.) is 30x20x8 mm whose power consumption is as small as 10 mW in average. Details are shown in Appendix.

3.2 Experimental settings

Based on our TDM method, it is possible to achieve both power supply and signal transmission with a single 2DC sheet. In this research, however, we fabricated double-layered sheet for avoiding interferences. Fig. 9 shows the schematic illustrations of our prototype. The bottom layer was used for power supply with 2.3 GHz microwave and the top layer was used for signal transmission with 2.6 GHz microwave.

There were totally 18 apertures on the sheet for connection. We used 3 of them for power supply and the other 3 for signal transmission. In our prototype, there was a standing wave in the sheet due to reflections at the side of the sheet. As a result, receivable electricity was depended on the location of the aperture. We intentionally chose the apertures for power supply and for signal transmission.

We adopted commercially available electrodes with pastes for the EMG measurement. Measured data were sampled at 1MHz through A/D converter to PC. The 10 W of power signals were supplied whose modulation frequency was 400 Hz.

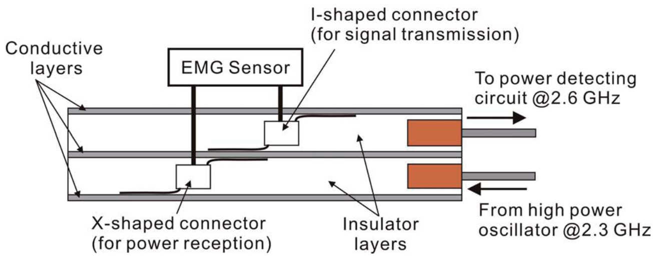

Fig. 9 Experimental settings with double-layered 2DC sheet.

3.3 Three channel data on a forearm

In the experiment, two channels were set along the “flexor carpi radialis muscle” which relates to the bending motion of a wrist (Channel A and B). The channel A was closer to an elbow than the channel B whose interval was about 6 cm. While the other one was set on an “extensor digitorum muscle” which relates to the extension of the wrist (Channel C). The data was measured when one subject bended and extended his wrist alternately in 3 seconds.

Figure 10 shows the observed signal at the power detecting circuit. In this graph two sensors were active. Due to the coupling condition difference, the amplitudes of the PWM signals were different. We confirmed that the width of the signal changed depending on the myoelectric signal intensity.

The demodulated data of three-channel EMG signals are illustrated in Fig. 11. The horizontal axis shows time and the vertical axis shows the integrated values of the voltage in each period. The value is logarithmically proportional to the myoelectric signal intensity. We used the integrated value here for the easiness for implementation.

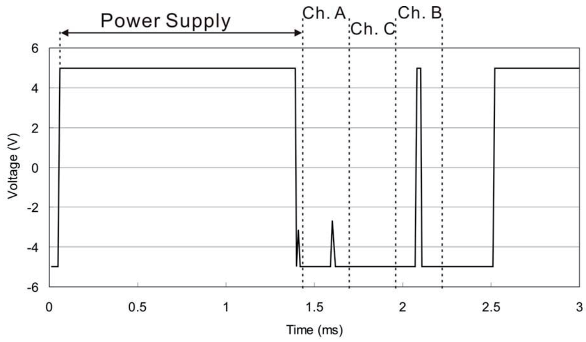

Fig. 10 Observed signal at the power detecting circuit.

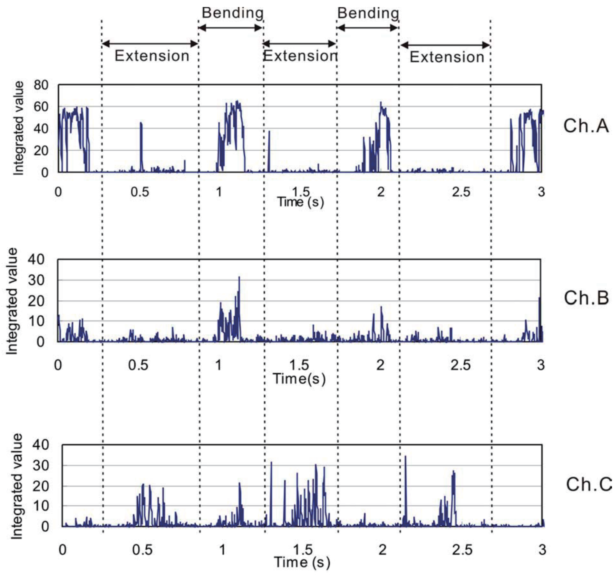

Fig. 11 Demodulated signals. The channel A and B were set upon the muscle related to the bending motion of the wrist. The channel C was set upon the muscle related to the extension of the wrist.

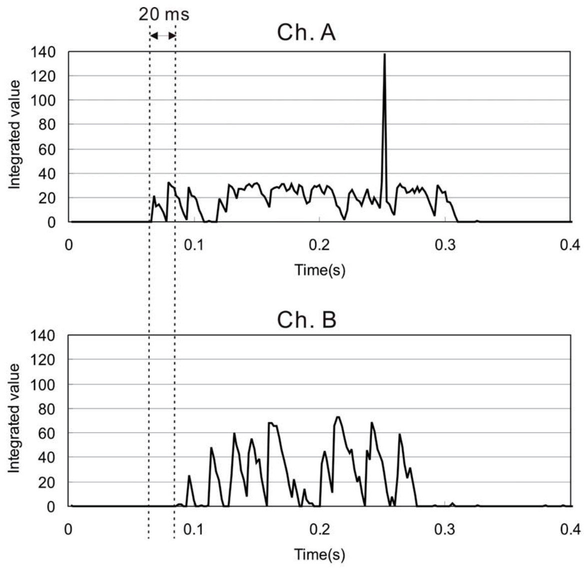

Fig. 12 Two channels data upon the same muscle.

The graphs show that the EMG signals occur alternately corresponding to the wrist motion. The signals from the channel A and B can be observed almost the same timing. However, when we focus on the onset of the signals (Fig. 12), we found the delay between the two signals. Since the conduction velocity of the myoelectric signal is known about 3~6 m/s and the interval of the two measurement points was 6cm, the 20 ms delay in our experiment represent the signal propagation.

4. CONCLUSIONS

In our previous studies, we proposed a new haptic interface that can predict finger motions based on myoelectric signals. Since the myoelectric signals relating to the finger motions can be obtained on a forearm, the finger motions are not restricted. In order to improve the accuracy of the motion estimation, it is considered that a high-density electrode array is required. So as to satisfy the requirement, "Two-Dimensional Communication (2DC)" technology we had proposed before is useful. Wristband-shaped EMG sensor array can be achieved without complicated and annoying wires.

In this paper, we show how to connect sensors to the sheet efficiently without electrical contact. We proposed a resonance-based connector for this requirement. Based on our experimental results, the connector can receive around 500 mW of electric power in average when 10 W of a microwave is supplied. It is sufficient for activating our EMG sensor unit.

We adopted a time division multiplexing (TDM) method for realizing both signal transmission and power supply with the same frequency in a single 2DC sheet. In this paper we used double-layered 2DC sheet for easiness of the implementation. We experimentally demonstrated that the 3 EMG sensors could be activated with microwave provided through the sheet. The demodulated data corresponded to the wrist motions. The results also indicated that the arrayed sensors could be used for tracing the myoelectric signal propagation.

APPENDIX

There are two stages in our TDM method. One is for power supply to each sensor unit and the other is for data transmission. Figure A-1 shows the time chart of our proposed method. After the power supply stage, each sensor sends data sequentially with PWM. The graph shows the envelope of the burst wave whose carrier frequency is supposed to be 2.4 GHz.

Figure A-2 shows the one configuration for achieving the TDM. The circuit is composed of the passive components, the logic circuits and the analog switch. Therefore the circuit can be driven with low power consumption.

The behavior of the circuit is understood by Fig. A-3 which shows the voltages at the points a~e in Fig. A-2. (We use description V_i (i = a~e) as the voltage to the common potential at the point i.) When the power signal ends, the V_a also becomes low with a time delay t_1. The delay t_1 is determined by the product of the resistance R_1 and capacitance C_1. Thus the rising edge of the V_b occurs t_1 second later than the falling edge of the power signal. The V_b is used to switch the state of the analog switch S. The switch S is connected to the EMG circuit during the V_b is low. On the other hand when the V_b becomes high, the point c is connected to the common potential through the resistance R_2. Then the V_c decreases with its time constant R_2C_2. The decay time t_2 is determined by the momentary voltage of the V_c at the exact moment when the V_b returns to high level. As a result, we obtain the PWM signal V_e. This signal is used as an "enable signal" of the 2.4 GHz oscillator. It is important that the time delay t_1 can readily be designed with different pairs of R_1 and C_1 so that the delays are different among multiple sensor chips.

The relationship between the voltage of the measurement circuit V_c and the pulse width t_2 can be calculated by this eqation.

$$t_2 = R_2 C_2 (\log V_{\mathrm{init}} - \log V_{\mathrm{th}}) \qquad \text{(A-1)}$$

Here V_{init} is the momentary voltage of the V_c at the exact moment when the switch S was connected. V_{th} is the low level threshold of the logic circuit.

Figure A-4 plots the experimental results of the relationship between the voltage V_{init} and the pulse width t_2. Here, R_2=300Ω and C_2=0.1μF. A designed differential voltage was supplied to the electrodes with stabilized voltage source. The dashed line indicates the theoretical curve. Though there is constant difference (about 10 μs) between the experimental results and the theoretical values, a tendency of the curve is similar to each other. The difference is considered to be caused by the variation in V_{th} of the used IC. This result indicates that the PWM is apparently possible by our proposed circuit.

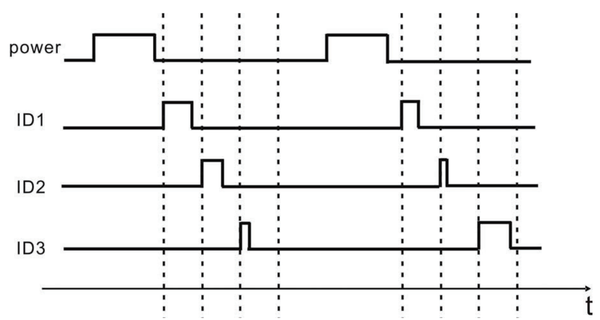

Fig. A-1 Time diagram of the time division multiplexing method. The lateral axis indicates the time. Each sensor has a different delay to the falling edge of the power signal.

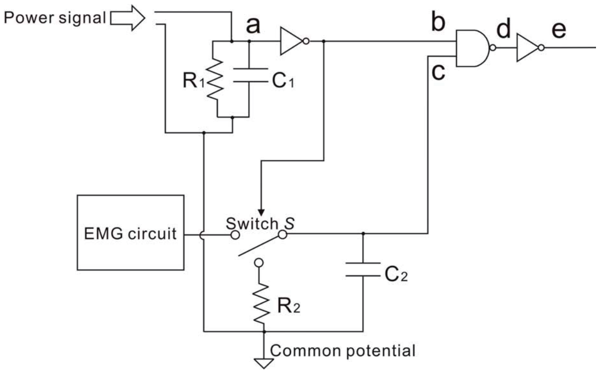

Fig. A-2 Electric circuit that achieves the TDM method with low power consumption.

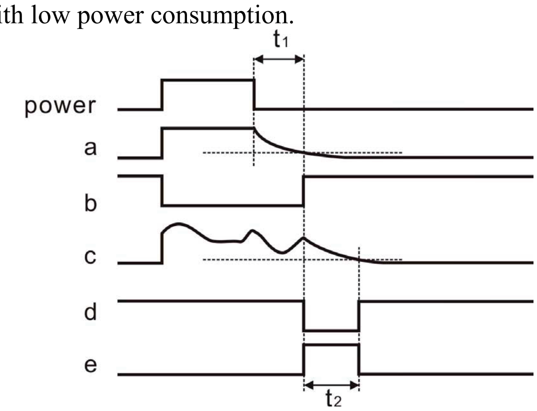

Fig. A-3 Time chart of the voltages at the points a~e shown in Fig. 9.

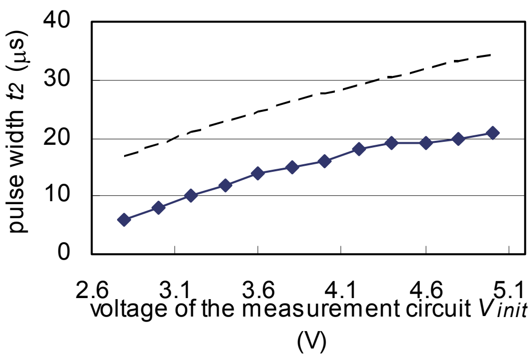

Fig. A-4 The relationship between the voltage of the EMG measurement circuit and the pulse width.

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