Capacitive Sensing of Electrocardiographic Potential Through Cloth From the Dorsal Surface of the Body in a Supine Position: A Preliminary Study

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***Abstract*—A method for obtaining electrocardiographic potential through thin cloth inserted between the measuring electrodes and the skin of a subject's dorsal surface when lying supine has been proposed. The method is based on capacitive coupling involving the electrode, the cloth, and the skin. Examination of a pilot device which employed the method revealed the following: 1) In spite of the gain attenuation in the high frequency region, the proposed method was considered useful for monitoring electrogardiogram (ECG) for nondiagnostic purpose. 2) The method was able to yield a stable ECG from a subject at rest for at least 7 h, and there was no significant adverse effect of long-term measurement on the quality of the signal obtained. 3) Electrode area was the factor that had most influence on the signal, compared with other factors such as cloth thickness and coupling pressure, but could be reduced to 10 cm^2 for heart rate detection. 4) Input capacitance of the device was assumed to be the dominant factor for the gain attenuation in the high frequency region, and should be reduced with a view to diagnostic use. Although there is still room for improvement in terms of practical use, the proposed method appears promising for application to bedding as a noninvasive and awareness-free method for ECG monitoring.**

***Index Terms*—Home health care, insulator electrode, nonobtrusive monitoring, unconstrained and noninvasive ECG measurement, wearable sensing.**

I. Introduction

The gradual aging of society in Japan, and the resulting increase in healthcare expenditure, has highlighted the need for efficient systems for monitoring of elderly people in their homes over long periods of time. Recording of physiological variables, such as the electrocardiogram (ECG), during everyday life could be useful for management of individuals with chronic health disorders [1]. Furthermore, real-life long-term health monitoring could be helpful for assessing the effects of treatment at home, and would also be potentially beneficial for observing deviations in health status from the norm at an early stage, or for automatically alerting paramedics in emergency cases. Consequently, long-term home health monitoring could be valuable not only for improving the quality of life (QOL) of senior citizens but also for reducing health care expenditure.

ECG has been part of routine cardiovascular evaluation for almost three decades, and is now recognized to have potential for long-term monitoring at home over periods of weeks or even months. Since conventional systems for ECG monitoring are unsuitable for long-term studies, previous researchers have focused on developing dedicated systems for long-term monitoring in daily life [2]–[11]. For example, Kwatra [2], [3], Ishijima [4], and Tamura [5] have developed systems for ECG or heart rate monitoring during bathing. Ishijima [6], [7] developed an automated system for ECG monitoring during sleep using embedded textile electrodes in a bed set. Park [8] proposed a prototype intelligent garment that can collect, process, store, and transmit information about the wearer, such as ECG parameters, and Catrysse [9], Scilingo [10], and Paradiso [11] have implemented this idea by using textile or fabric sensors. Since a noninvasive and awareness-free approach is essential for long-term home monitoring, the idea of sensors embedded in furnishings or daily clothing is quite rational. However, there are still some challenges to be addressed regarding the utility of such systems.

In conventional ECG measurement, an electrolytic paste or a conductive adhesive is almost always required for maintaining reliable ohmic contact with the skin. Therefore, ECG measurement for a long period using conventional methods causes irritation and discomfort, and is a potential cause of skin allergy and inflammation. Despite these disadvantages of conventional ECG detection, sufficient consideration has not been given to skin-to-electrode coupling [8]. Previous authors [6], [7], [9]–[11] have addressed this problem by employing textile electrodes, which do not require any electrolytic paste or conductive adhesive for measurement, to relieve potential irritation and discomfort. Nevertheless, the potential for metal allergy or discomfort upon direct long-term contact of the skin with metal material has not yet been addressed.

With the above background in mind, the authors have focused on the principle of capacitive sensing capable of detecting alternating electrical potential through an inserted thin insulator, and have applied the principle for measurement of electrocardiographic potential through commonly available cloth from the dorsal surface of the body when the subject is supine. In order to explore the feasibility and applicability of this idea, the authors have fabricated a pilot measuring device and conducted some preliminary experiments and model studies using it.

II. Capacitive Sensing

Capacitive sensing is based on the principle of the capacitive (or insulated) electrode [12]–[18], which is an electrode not requiring the use of electrolytic paste or conductive adhesive. The electrode relies on capacitive coupling, conventionally involving a metal electrode, an insulator, and the skin, as shown in Fig. 1(a). The electrode can carry an alternating bioelectric current equivalently through the capacitance of the coupling. In previous studies, various insulators, such as anodized aluminum [12], [13], silicon dioxide [14], [15], pyre varnish [16], anodic insulated tantalum oxide [17], or barium titanate [18], have been tested as these materials exhibit high permittivity. In the present study, commonly available cloth, especially cotton, was substituted for these rigid insulators to relieve the irritation, allergy and discomfort experienced with conventional skin-to-electrode coupling, as shown in Fig. 1(b). Also a sheet of conductive fabric was substituted for the conventional metal electrode so as to achieve a deformable coupling corresponding to the contour of the coupled region.

Capacitive sensing is also based on an impedance matching circuit to mediate the high impedance of the coupling with a low impedance required by the subsequent circuitry. For example, a field-effect transistor

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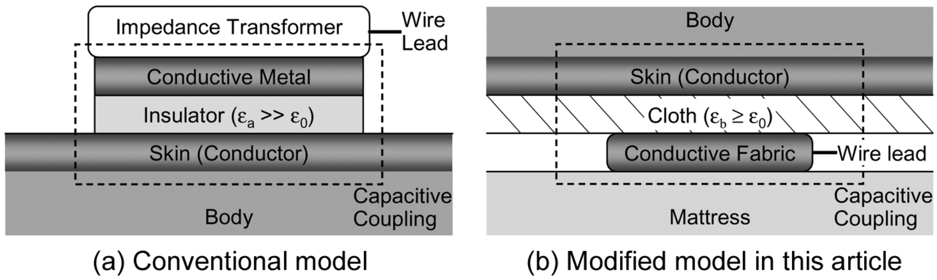

Fig. 1. Schematic models of (a) conventional and (b) modified capacitive electrodes coupled to the skin. ε_a and ε_b indicate dielectric constants of the inserted insulator and the cloth, respectively. ε_0 is the permittivity of free space.

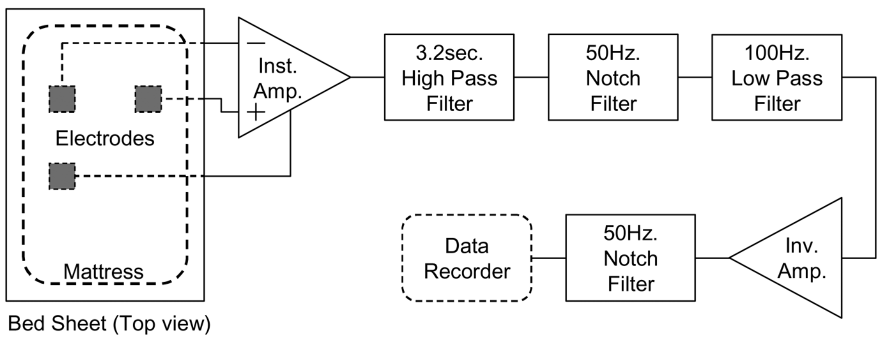

Fig. 2. Configuration of measuring electrodes and block diagram of the present system.

(FET) source follower has been used as an impedance matching circuit and actively mounted in an electrode housing [15], [16] for reducing common mode interference caused by power lines. In the present study, an instrumentation amplifier [19] composed of operational amplifiers with high input impedance (National Semiconductor, LF356, 1000 $\mathrm{G}\Omega$ according to the specification sheet)[1] was employed in order to match the greater impedance of the coupling due to the smaller dielectric constant of the inserted cloth in comparison with conventional insulators. The impedance matching circuit was not mounted in the electrode housing to avoid hard contact with the skin.

III. Materials and Methods

A. Fabric Electrode

The electrocardiographic signal is picked up by electrodes made of a square sheet of conductive fabric with conductive acrylic adhesive (3M, 2191FR). The electrodes are stuck to a mattress with the adhesive and covered by a cotton bed sheet. Two lead electrodes are located under the right and left scapulae of the subject when lying in a supine position, as shown in Fig. 2. A reference electrode is placed beneath the right lumbar region. Skin-cloth-electrode coupling is held by the subject's weight on the cloth and by repulsive force from the mattress. The lead electrodes are connected to a measuring device, as described in the next section, by using shielded wires 2.0 m long. Three sets of measuring electrodes, each with an area of 70, 40, or 10 cm^2, were prepared for examining the influence of the electrode area on an obtained signal.

B. Signal Extraction

The pilot measuring device with filtering and amplification circuitry was fabricated using off-the-shelf components. The device consists of an instrumentation amplifier, a high-pass filter, two notch filters, a low-pass filter and an inverting amplifier. A block diagram of the device is shown in Fig. 2. The instrumentation amplifier is employed not only as a differential amplifier but also as the impedance matching circuit. The circuit elements of the high-pass filter and the low-pass filter are determined in order to obtain a cutoff frequency of 0.05 and 100 Hz, respectively. A Butterworth filter (NF, CF-2BE-HB-50Hz-Q5) is used for the former notch filter in order to reduce 50-Hz interference. The device is powered by four series D batteries to obviate the possibility of electric shock. A dc-dc converter (Datel, BST-12/105-d5)[2] is used to boost the nominal 6.0 V supply to a regulated ± 12 V.

C. Simultaneous Measurement With a Conventional Device

One male subject, aged 22, naked from the waist up, was instructed to lie supine on the mattress bearing the electrodes with an area of 70 cm^2. A cotton bed sheet 395 $\mu\mathrm{m}$ thick was used to cover the electrodes and the mattress. Electrocardiographic potential was recorded using the developed device from the dorsum of the subject through the inserted cotton sheet. The output signal from the device was digitized at 1 kHz by an analog-to-digital converter and stored in a personal computer using a data acquisition system (Biopac Systems, MP-150 system). As a reference signal, the ECG was recorded using a conventional bioamplifier (Teac Instruments, BA1104-CC) and disposable electrodes (Nihon Kohden, Vitrode F-150M).[3] The disposable electrodes were attached directly to the dorsal surface of the subject at positions close to each electrode on the mattress. As another reference, the Lead I ECG was measured using a standard limb lead with the same conventional bioamplifier. Both reference signals were wirelessly transmitted and received with a telemeter unit (Teac Instruments, TU-4), and then digitized simultaneously with the output signal of the developed device.

[1][Online]. Available at http://www.national.com/pf/LF/LF356.html.

[2][Online]. Available at http://www.datel.com/data/power/bst-3w.pdf.

[3][Online]. Available at http://www.nihonkohden.com/products/type/supplies/dispo_electrodes.html.

D. Experimental Long-Term Monitoring Using the Device

In order to examine a stability of the capacitive coupling of our device, an experiment to examine long-term monitoring was conducted using the same volunteer. The subject, naked from the waist up, was instructed to lie supine at rest for 7 h on the mattress, which was covered by the same cloth, in the same manner as that described in Section III-C. An initial measurement was conducted for 15 min, and then every hour thereafter.

E. Influence of Coupling Condition on the Obtained Signal

If the coupling between an electrode and the skin can be modeled as in Fig. 1(b), the capacitance C of a coupling having a distance of d meters, a coupling area of S square meters, and a coupling dielectric constant ε is given by

$$C = \varepsilon \frac{S}{d}. \tag{1}$$

Since any change in the capacitance of the coupling is related to a change in impedance (i.e. voltage loss) at the coupling, the output signal of the developed device is considered to be affected by the area of the coupling (i.e. electrode area), the distance between the electrode and the skin, and the dielectric constant of the coupling. Then, the influence of coupling condition on the output signal can be examined by changing the electrode area, the thickness of the inserted cloth, and the pressure on the coupling.

1) Influence of Electrode Area: Measuring electrodes having an identical area of 70, 40, or 10 cm^2 were placed on the mattress according to the configuration shown in Fig. 2. The electrodes were covered with the cotton sheet 395 $\mu\mathrm{m}$ thick. Then, three synthetic human dorsal surfaces made of square pieces of conductive fabric having the same area of the measuring electrode were attached to the cloth at locations just above each of the measuring electrodes, so that the whole area of each measuring electrode was involved in capacitive coupling with each synthetic dorsal surface. The electrodes, the sheet and the synthetic dorsal surfaces were subjected to 1751 Pa pressure using a weight to mimic the pressure resulting from a supine adult subject 1.65 m in height and weighing 60 kg (dorsal area 0.336 m^2). The simulated ECG signal was input from an ECG generator (Nihon Kohden, AX-201D) to the synthetic dorsal surfaces. The output signal from the device, and also the frequency-gain response from 0.01 to 500 Hz, was measured for each of the different electrode areas.

2) Influence of Thickness of the Inserted Cloth: Cotton sheets with different thicknesses of 395, 463, or 1020 $\mu\mathrm{m}$ at atmospheric pressure were inserted between the electrode and the skin, and the output signal from the device was measured for each thickness. Three 70-cm^2 electrodes, three synthetic dorsal surfaces, the ECG generator and a pressure of 1751 Pa were used in a similar way to that in Section III-E1). Also, the frequency responses were measured respectively for each sheet thickness.

3) Influence of Pressure on the Coupling: The simulated ECG and the frequency response for each thickness of cotton sheet were measured at a pressure of 1751, 1083, or 578 Pa. The 70-cm^2 electrodes, the synthetic dorsal surfaces and the ECG generator were used in a similar way to that described in Section III-E1). The pressure of 578 Pa mimics experimentally the conditions created by a 2.36-kg neonate with a contact area of 0.040 m^2. Additionally, the least pressure capable of detecting the R-wave was investigated for each cloth thickness. The room temperature was $19\,^\circ\mathrm{C} \sim 22\,^\circ\mathrm{C}$ and the relative humidity was $46\% \sim 56\%$.

4) Influence of Electrode Conductivity: In order to investigate whether the conductivity of the electrode influenced the results, the simulated ECG and the frequency response without any inserted cloths were measured at each pressure. The 70-cm^2 electrodes, the synthetic dorsal surfaces and the ECG generator were used in a similar way to that described in Section III-E1).

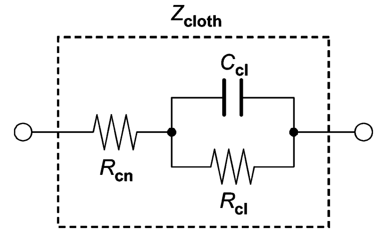

Fig. 3. Equivalent circuit of the capacitively coupled cloth. R_{cn} is the contact resistance between the cloth and the conductive surfaces such as electrode or skin. R_{cl} and C_{cl} are dc resistance and capacitance of the cloth, respectively.

F. Frequency-Impedance Response of the Inserted Cloth

In order to investigate the interaction between the electrical properties of the inserted cloth and the obtained signal, the frequency-impedance response was measured for each cloth with a precision LCR meter (Agilent Technologies, 4284A). The cloth was fixed at a constant pressure by a dielectric test fixture (Agilent Technologies, 16451B). The frequency was varied from 20 Hz, which is the lower limit of the instrument, to 300 Hz. To complement the deficit of the raw data below 20 Hz, we used the model of capacitively coupled cloth shown in Fig. 3, and then estimated the model parameters for each cloth by fitting the following equation to the measured impedance of the cloth, Z_{cloth}:

$$|Z_{\mathrm{cloth}}(f)| = R_{cn} + \frac{R_{cl}}{\sqrt{1 + (2\pi f C_{cl} R_{cl})^2}} \tag{2}$$

where C_{cl} and R_{cl} are the capacitance and dc resistance of the cloth, R_{cn} is the contact resistance between the cloth and conductive surfaces, and f is the frequency.

G. Input Capacitance of the Developed Device

We expected that the input capacitance of the device would eventually degrade the performance in the high frequency region. Since the input capacitance was too low to be measured with conventional equipment, we devised a method for assessing it. This procedure was performed as follows. First, two off-the-shelf capacitors with the same capacitance were connected at the front ends of the differential input of the device instead of the two lead electrodes, and the frequency response while the capacitors were inserted was measured. Then, the gain in decibels at each frequency with the capacitors inserted was subtracted from the corresponding original gain measured without the capacitors. Second, the developed device connected to the two capacitors was modeled by an equivalent circuit, as shown in Fig. 4. In the figure, the gain of the device with the capacitors, $G_{\mathrm{cap}}(f) = v_{\mathrm{out}}/v_{\mathrm{in}}$, can be expressed using the original gain of the device, $G_{\mathrm{org}}(f) = v_{\mathrm{out}}/v'_{\mathrm{in}}$, and circuit components in Fig. 4, as follows:

$$G_{\mathrm{cap}}(f) = \frac{j2\pi f R_{\mathrm{in}} C_0}{2 + j2\pi f R_{\mathrm{in}}(2C_{\mathrm{in}} + C_0)} G_{\mathrm{org}}(f) \tag{3}$$

where C_0 is the capacitance of the externally inserted capacitors, and C_{in} and R_{in} are the input capacitance and input resistance of the device, respectively. According to (3), the gain difference in decibels between $20\log|G_{\mathrm{cap}}|$ and $20\log|G_{\mathrm{org}}|$, i.e. $20\log(|G_{\mathrm{cap}}|/|G_{\mathrm{org}}|)$, can be described as follows:

$$20\log\frac{|G_{\mathrm{cap}}(f)|}{|G_{\mathrm{org}}(f)|} = 20\log\frac{2\pi f R_{\mathrm{in}} C_0}{\sqrt{4 + 4\pi^2 f^2 R_{\mathrm{in}}^2 (2C_{\mathrm{in}} + C_0)^2}}. \tag{4}$$

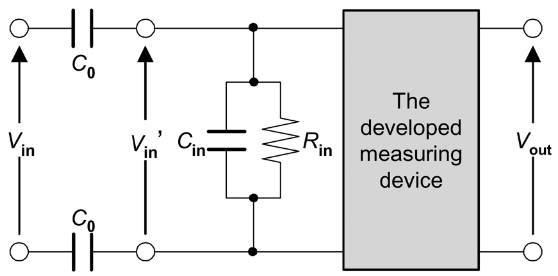

Fig. 4. Equivalent circuit of the input part of the developed device in which capacitors are inserted at the front ends. C_0 is the capacitance of the inserted capacitor. $C_{\rm in}$ and $R_{\rm in}$ are the input capacitance and input resistance of the device, respectively.

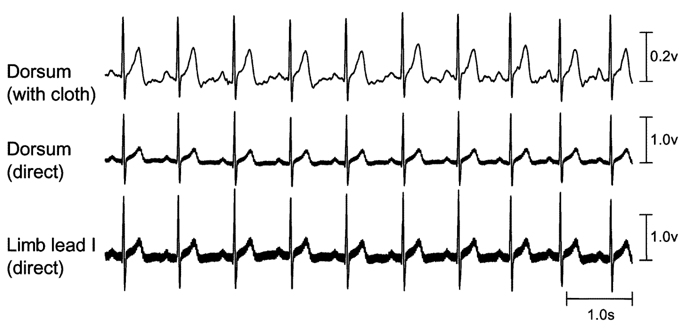

Fig. 5. Recordings obtained simultaneously using the present device with a cotton sheet 395 μm thick from the dorsum (top), using a conventional bioamplifier without a cloth from the dorsum (middle), and using a conventional bioamplifier and a standard limb lead without a cloth (bottom). The top recording was subjected to a 20-point moving average to eliminate power line noise.

Since the value of C_0 is known and $R_{\rm in}$ can be obtained from the specification sheet of the IC used in the device, we can consider (4) as a function of f and $C_{\rm in}$. Therefore, we can estimate $C_{\rm in}$ by fitting (4) to the experimentally obtained gain difference between $20\log|G_{\rm cap}|$ and $20\log|G_{\rm org}|$.

In the actual experiment, ceramic 102-pF or 478-pF capacitors were connected individually via the lead wires, and the two frequency responses were measured. Moreover, frequency responses not including the wires were also measured. The capacitance of the capacitors was measured with the precision LCR meter.

IV. Results

A. Simultaneous Measurement With a Conventional Device

Fig. 5 shows recordings typical of those obtained. The output signal of the developed device showed periodic waveforms synchronized to, and similar to, the reference ECGs. Therefore, the proposed method was considered useful at least for detecting heart rate and/or other periodic parameters of the ECG even with the cloth inserted. However, the ratios of the peak amplitude of the P-wave and T-wave to the R-wave in the top recording appeared to be slightly greater than those in the reference ECGs. This distortion was probably due to inconstant gain through the pass band of our device when the cloth was inserted.

B. Experimental Long-Term Monitoring Using the Device

As can be seen in the top recording in Fig. 6, depressed R-waves and skewed T-waves were observed at the beginning of the experiment.

Fig. 6. Long-term recordings using the present device from the dorsum with an inserted cotton sheet 395 μm thick. S/N ratio of the recordings after 4 h was as good or better than that at the 4-h point. The room temperature was 21 °C and the relative humidity was 24%.

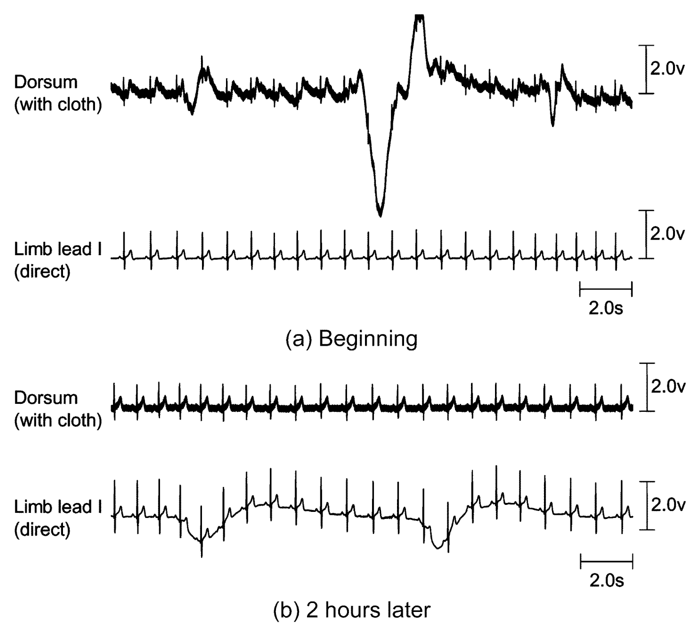

Fig. 7. Typical recordings contaminated by motion artifacts during long-term measurement. (a) Recording made at the beginning, and (b) 2 h later. The top recording in each figure is that obtained using the present device from the dorsum with a cotton sheet 395 μm thick inserted, and the bottom recording is the wirelessly measured lead I ECG.

However, they had disappeared before measurement was conducted at the 1-h point. In addition, there was a marked decrease in hum noise in recordings after 4 h, approaching a level as low as that in the directly measured ECG.

Fig. 7 shows typical recordings contaminated by motion artifacts at the beginning, and 2 h later. Since the impedance of the coupling is assumed to have been high at the beginning, as can be seen from the low signal-to-noise (S/N) waveform in Fig. 6, the output signal from the device at the beginning was susceptible to motion artifacts. As a result, it is evident at the beginning of the experiment that the motion of the subject contaminated the output from the developed device, but not the lead I ECG from the conventional electrocardiograph [see Fig. 7(a)]. However, once the initial impedance of the coupling had decreased after a while, the output of the device seemed to be more stable than a conventional electrocardiograph. As seen in Fig. 7(b), limb movements contaminated the lead I ECG, but not the output signal from the developed device, contrary to the situation observed at the beginning.

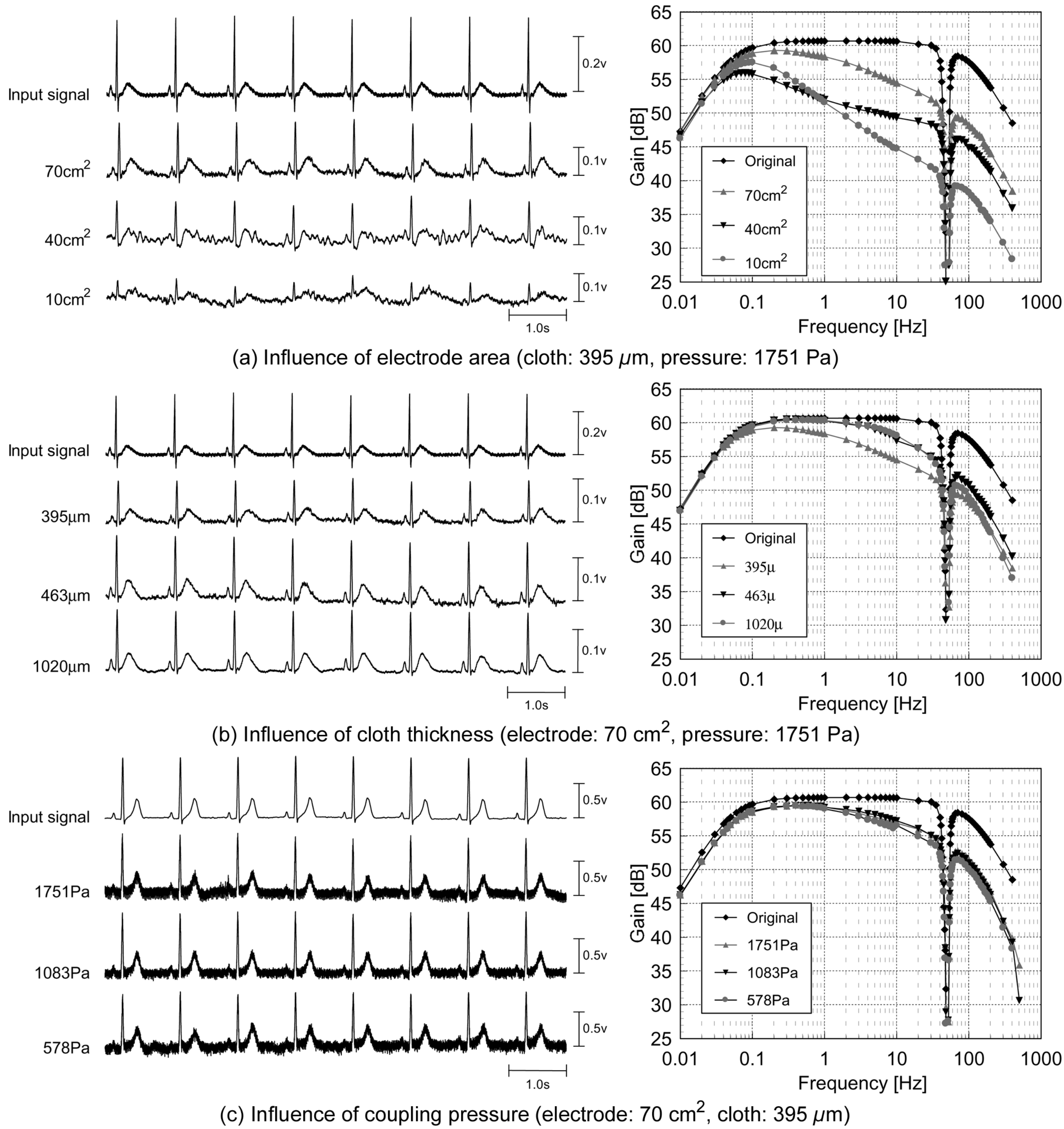

Fig. 8. Influence of (a) electrode area, (b) cloth thickness, and (c) coupling pressure on ECG recording (left column) and on frequency response (right column). The frequency response labeled "original" corresponds to that obtained directly from the measuring electrode. The gain at 50 Hz was sharply attenuated because 50-Hz band elimination filters were used in the system. The original frequency response satisfied the Japanese Industrial Standard for an electrocardiograph (JIS T1201), except around the 50-Hz frequency band.

This phenomenon was often observed in recordings after 2 h. Therefore, there was no significant adverse effect of long-term measurement on the quality of the output signal from the developed device.

C. Influence of Coupling Condition on the Output Signal

1) Influence of Electrode Area: As shown in Fig. 8(a), the periodic R-wave was evident in all recordings, even for the smallest electrode area of 10 cm^2. However, the waveform of the PQRST complex became distorted as the electrode area decreased. Frequency responses in Fig. 8(a) suggests that this distortion was attributable to a tilted frequency-gain response in the 0.1–100 Hz region. Since the gain in the high-frequency domain decreased as the area of the electrode decreased, the R-wave was depressed as the area decreased and the S-wave was absent in the recording obtained with the 10-cm^2 electrodes. Therefore, the area can be reduced to 10 cm^2 if only heart rate is being detected.

TABLE I
ESTIMATED VALUES OF CONTACT RESISTANCE (R_{cn}), dc RESISTANCE (R_{cl}), CAPACITANCE (C_{cl}), AND RELATIVE PERMITTIVITY OF THE INSERTED CLOTHS. RELATIVE PERMITTIVITY WAS ESTIMATED BY USING (1)

Thickness [μm]	R_{cn} [MΩ]	R_{cl} [MΩ]	C_{cl} [pF]	Relative Permittivity
395	1.93	209	59	2.3
463	3.96	152	71	3.3
1020	8.23	121	39	4.0

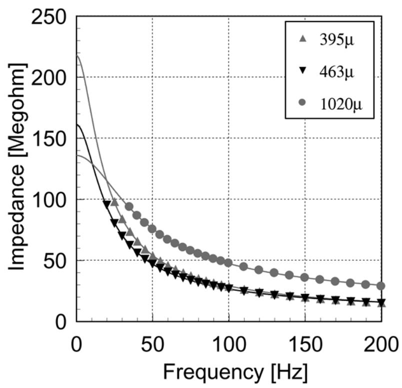

Fig. 9. Frequency-impedance responses of the inserted cloths. Solid lines in the figure are the best fitted curves obtained by applying (2). Estimated parameters are shown in Table I. The room temperature during the measurement was 24 °C and the relative humidity was 40%∼46%.

2) Influence of Thickness of the Inserted Cloth: As can be seen in Fig. 8(b), the PQRST complex was sufficiently recognizable even with a cloth thickness of 1020 μm, and the quality was similar to that of other recordings obtained with thinner cloth. Although the gain had a tendency to decrease as the frequency increased, all the frequency responses obtained with the cloth indicated a close gain for each frequency. In fact the responses obtained with the 463 μm cloth and the 1020 μm cloth were almost the same. This may be because the coupling impedance of these cloths is almost the same at a pressure of 1751 Pa. On the other hand, the least gain was obtained for the 395-μm cloth, despite the fact that this was the thinnest. This minimal gain for the 395 μm cloth was presumably due to the low dielectric constant and the high resistivity of the cloth. Thus, it appears that the thickness of the inserted cloth at atmospheric pressure can be increased to at least 1000 μm as long as the electrode is coupled firmly with sufficient pressure.

3) Influence of Pressure on the Coupling: As indicated in Fig. 8(c), reduction of the pressure from 1751 to 578 Pa had little influence on both the S/N and the waveform of the recordings for the 395 μm cloth. The frequency responses, shown in Fig. 8(c), agreed with the results. With regard to the other two cloths, little effect was also confirmed. Moreover, the R-wave could be recognized for each cloth thickness even when no weight was placed on the electrodes. Thus, the developed device shows promise for application not only to adult subjects, but also to neonates for heart rate monitoring.

4) Influence of Electrode Conductivity: No gain attenuation of the electrodes was confirmed from 0.01 to 500 Hz. Therefore, the gain attenuation observed in Fig. 8 was not due to gain loss at the electrode but to that at the coupling involving the cloth.

D. Frequency-Impedance Response of the Inserted Cloth

In accordance with the results shown in Fig. 9, impedance of the 395 μm cloth was estimated to be the highest in a frequency region below 10 Hz. Table I indicates this was due to the largest dc resistance and the least relative permittivity of the cloth. Thus, the least gain for the 395 μm cloth in Fig. 8(b) can be attributed to these electrical properties of the cloth.

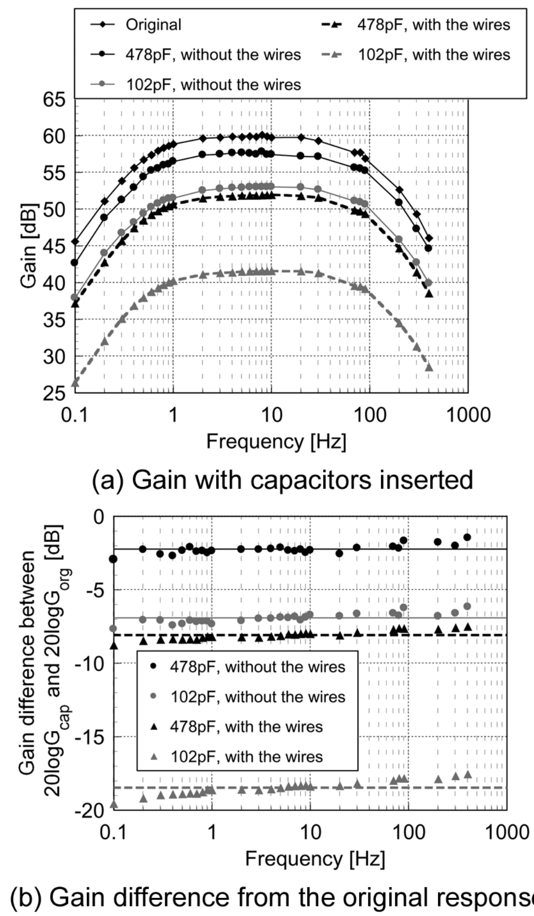

Fig. 10. Frequency response of (a) the gain with capacitors inserted at the front ends of the system and (b) the difference in gain between (a) and the original response. The original frequency response is that measured directly from the electrodes. The solid and dotted lines in (b) are the best fitted curves obtained by applying (4).

E. Input Capacitance of the Developed Device

As shown in Fig. 10, the plots obtained when the lead wires were included were apparently smaller than those excluding the wires. As a consequence, $C_{\rm in}$ of 370 and 65 pF indicated the least root mean square error for the results including and excluding the wires, respectively. Thus, the lead wires had an input capacitance of 305 pF.

V. DISCUSSION

A. Perspiration of the Subject

In a previous article, Geddes [20] reported that "dry" electrodes are really only dry when first applied because of subsequent perspiration, and that impedance at 20 min was between one-fourth and one-fifth of the initial impedance. Therefore, the improvement of the S/N with time in Fig. 6 was probably due to a decrease in both the impedance of the cloth and the contact resistance between the cloth and the skin, originating from the perspiration of the subject. In long-term measurement,

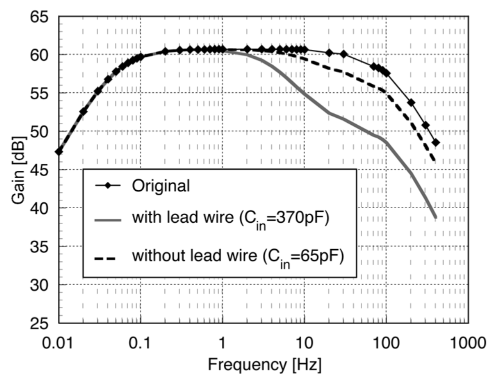

Fig. 11. Frequency responses of the simulated gain using (5) with the 395-$\mu\mathrm{m}$ cloth inserted. The original frequency response was that measured directly from the electrodes. Parameter values for estimating the response are as follows: $R_{cn} = 312\ \mathrm{k}\Omega$, $C_{cl} = 365\ \mathrm{pF}$, $R_{cl} = 34\ \mathrm{M}\Omega$, $C_{\mathrm{in}} = 65$ or 370 pF, $R_{\mathrm{in}} = 1\ \mathrm{T}\Omega$. R_{cn}, R_{cl}, and C_{cl} were compensated from the values in Table I by considering the electrode area of the precision LCR meter (11.3 cm^2) and of the fabric lead electrodes (70 cm^2).

once a perspiration layer has formed, there will be a physical contact between the electrodes and the skin. Therefore, for the next step, the current nickel-plated conductive fabric should be substituted by another fabric containing nonallergenic conductive materials.

B. Motion Artifact During Measurement

As shown in Fig. 7(b), limb movements contaminated the lead I ECG, but not the output signal from the developed device, contrary to the situation observed at the beginning. We consider that three possible factors can explain this phenomenon. The first factor is the greater input impedance of the present device. The input impedance was estimated to be greater than 430 $\mathrm{M}\Omega$ in a frequency region below 1 Hz, even with an input capacitance of 370-pF, by using the equivalent circuit in Fig. 4. Thus, once resistive coupling had been accomplished at the electrode-to-skin coupling by perspiration, the slow sub-1-Hz component of the change in contact impedance caused by limb motion would have had less impact on the obtained signal than if a conventional device had been used. The second factor is that the lead wires of the developed device were connected not to the electrodes on the peripheral limbs but to those on the mattress under the trunk. Accordingly, subtle limb motion would hardly disturb the lead wires of the developed device. The third factor is detachment of the sides of the disposable electrode pad due to perspiration, perhaps reducing the adhesion of the electrode.

Thus, future issues that need to be addressed are shifting of the trunk off the electrodes by larger motions such as rolling over, and the motion artifact at the beginning of measurement. One way of coping with these issues would be introduction of multiple arrays of horizontal strip electrodes, and of filters and software to search for stable electrode sets, similar to the approach reported by Niizeki [21]. Another method would be to narrow the filter band width for heart rate monitoring.

C. Gain Attenuation in the High Frequency Region

As shown in Fig. 8, the frequency responses with the cloth inserted indicated that the gain of the device decreased as the frequency increased. This attenuation is perhaps due to the input capacitance of the developed device. Since an equivalent circuit of the input part of the device with cloth inserted at the front ends can be modeled by substituting C_0 in Fig. 4 for Z_{cloth} in Fig. 3, the gain with the cloth inserted, $G_{\mathrm{cloth}}(f)$, can be described as follows:

$$G_{\mathrm{cloth}}(f) = \frac{(j4\pi f C_{\mathrm{in}} + 2/R_{\mathrm{in}})^{-1} \cdot G_{\mathrm{org}}(f)}{R_{cn} + (j2\pi f C_{cl} + 1/R_{cl})^{-1} + (j4\pi f C_{\mathrm{in}} + 2/R_{\mathrm{in}})^{-1}}. \tag{5}$$

We can estimate the frequency-gain responses with the cloth inserted by substituting the parameter values in (5) and by using the experimentally obtained $G_{\mathrm{org}}(f)$. As shown in Fig. 11, the simulated response when C_{in} equals 370 pF exhibited a characteristic similar to the experimentally obtained responses shown in Fig. 8. Moreover, the simulation when C_{in} equals 65 pF predicted that gain attenuation in the high-frequency regions could be suppressed by decreasing the input capacitance. Therefore, it is necessary to reduce the input capacitance, for example by shortening the wire length between the electrodes and the impedance matching circuits, in order to acquire a less distorted waveform and to make the device applicable for diagnostic use.

VI. Conclusion

We have proposed an approach for obtaining electrocardiographic potential from the dorsal surface of a subject lying supine under conditions whereby a thin cloth is inserted between the electrodes and the skin. We fabricated a pilot measuring device and explored the feasibility and applicable scope of the proposed method. Examination of the device yielded the following results.

1) In spite of the gain attenuation in the high frequency region, the proposed method was considered useful for monitoring ECG for nondiagnostic purposes.
2) The method was able to yield a stable ECG from a subject at rest for at least 7 h, and there was no significant adverse effect of long-term measurement on the quality of the signal obtained.
3) Electrode area had a greater influence on the signal than cloth thickness and coupling pressure, but could be reduced to 10 cm^2 for heart rate detection.
4) Input capacitance of the device was assumed to be the dominant factor for the gain attenuation in the high frequency region, and this should be reduced to make the device applicable for diagnostic use.

Although there is still room for improvement in terms of its practical use, the proposed method appears promising for application to bedding as a noninvasive and awareness-free method for ECG monitoring.

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Wavelet Packets Feasibility Study for the Design of an ECG Compressor

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***Abstract*—Most of the recent electrocardiogram (ECG) compression approaches developed with the wavelet transform are implemented using the discrete wavelet transform. Conversely, wavelet packets (WP) are not extensively used, although they are an adaptive decomposition for representing signals. In this paper, we present a thresholding-based method to encode ECG signals using WP. The design of the compressor has been carried out according to two main goals: 1) The scheme should be simple to allow real-time implementation; 2) quality, i.e., the reconstructed signal should be as similar as possible to the original signal. The proposed scheme is versatile as far as neither QRS detection nor *a priori* signal information is required. As such, it can thus be applied to any ECG. Results show that WP perform efficiently and can now be considered as an alternative in ECG compression applications.**

***Index Terms*—ECG compression, electrocardiogram, filter bank, subband coding, thresholding, wavelet coding, wavelet packets (WP), wavelet transform.**

I. Introduction

The electrocardiogram (ECG) is widely used because it is a noninvasive way to establish clinical diagnosis of heart diseases. Therefore, ECG processing has been a topic of great interest and is most commonly used in applications such as monitoring. Long-term records have also become commonly used to extract or detect important information from the heart signals. In these cases, the quantity of data grows significantly and compression is required for reducing the storage required and transmission times.

A large amount of lossy ECG compression methods have been developed using many different techniques. In [1], a classification of compression schemes into three categories is proposed: *Direct methods, transform methods, and other compression methods*. Most of the methods classified in the second group are based on the wavelet transform (WT) and multiresolution analysis, enabling them to obtain good compression ratios (CRs) and real-time implementation. The set partitioning in hierarchical trees (SPIHT) algorithm [2] is the most widely known algorithm that fulfills these features. Subsequently, it has been surpassed by other approaches [3], [4] that give high CR, with the tradeoff being higher computational complexity.

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