ECG Recording on a Bed During Sleep Without Direct Skin-Contact

Yong Gyu Lim, Ko Keun Kim, and Kwang Suk Park*, *Member, IEEE*

***Abstract*—A new indirect contact (IDC) electrocardiogram (ECG) measurement method (IDC-ECG) for monitoring ECG during sleep that is adequate for long-term use is provided. The provided method did not require any direct conductive contact between the instrument and bare skin. This method utilizes an array of high-input-impedance active electrodes fixed on the mattress and an indirect-skin-contact ground made of a large conductive textile sheet. A thin cotton bedcover covered the mattress, electrodes, and conductive textile, and the participants were positioned on the mattress over the bedcover. An ECG was successfully obtained, although the signal quality was lower and the motion artifact was larger than in conventional direct-contact measurements (DC-ECG). The results showed that further studies are required to apply the provided method to an ECG diagnosis of cardiovascular diseases. However, currently the method can be used for HRV assessment with easy discrimination of R-peaks.**

***Index Terms*—Active electrode, ECG monitoring on a bed, high-input impedance, indirect-contact ECG.**

I. Introduction

THIS study dealt with an electrocardiogram (ECG) measurement during sleep that enables long-term daily recording at home by minimizing intrusion on the subject's life.

In comparison with other measurements of the cardiovascular system, ECG measurement has the merits of simplicity, minimal constraint, it is relatively inexpensive, and is adequate for long-term measurement. ECG provides much useful diagnostic information about the cardiovascular system. In addition, heart rate and its variability (HRV) derived from the ECG data reflect autonomic nervous system activity and the mental status [1].

It is not odd to measure useful signals such as ECG even during sleep. ECG recording during sleep is used mainly to diagnose various sleep disorders as a part of polysomnography. Recently, the need for ECG measurement during sleep at home has increased. One purpose is to diagnose sleep disorders by using an instrument set reduced from the full complement of polysomnography equipment for easy home usage [2]. Another is to assess long-term (24-h) HRV. In addition, the increasing interest in daily-life health monitoring has increased the need for ECG measurement during sleep.

One of major constraints of ECG measurement is the requirement for maintaining contact between the electrode and bare skin during the measurement. This constraint is highlighted with long-term daily measurement. There are some other cardiovascular signal measurements for home usage, such as photoplethysmography (PPG), phonocardiogram, impedance-cardiogram, echocardiogram, and ballistocardiogram. Most of these methods are very difficult to be applied to nonintrusive daily measurement during sleep. There have been, nevertheless, many previous studies on nonintrusive measurement using a bed [3]–[6]. Those studies made some progress in eliminating the requirement of direct instrument attachment to the body. However, most of the previous studies dealt with mechanical signals originating from the dynamics of the cardiovascular system.

There are some previous studies of ECG measurement on a bed aimed at nonintrusive daily ECG monitoring [7], [8]. They utilized conductive textile sheets fixed to the pillow and bed cover. However, since the feel of the conductive textile is not pleasant, that method seems inadequate for long-term everyday ECG monitoring, especially since the conductive textile placed on the pillow contacts the very sensitive neck and face. In addition, long-term contact between skin and the conductive textile may cause skin irritation. This paper presents a new ECG measurement method without direct conductive contact between the skin and electrodes, aimed at nonintrusive daily ECG recording during sleep. In the presented method, the electrodes are attached to the mattress and ECG is measured through the clothes, without direct skin-electrode contact.

II. Methodology

ECG measurement through clothes is characterized by very high electrode impedance (the impedance between the electrode and the skin). The study on ECG measurements with electrodes of high impedance dates back to the late 1960s. At that time, many types of insulated electrodes were designed as an alternative to the Ag-AgCl electrode for special uses [9]–[12]. The insulated electrodes have very high electrode impedance in comparison with the Ag-AgCl electrode; therefore, the insulated electrodes show a lower ECG quality. Naturally, most of the studies were aimed at enhancing signal quality. One solution is to reduce the electrode impedance by applying a thinner insulating layer, and another is to improve the electric circuitry. One important achievement of the previous studies was the introduction of active electrodes in which a high-input-impedance preamp is embedded.

The ECG measurement through clothes is similar to the measurements made with insulated electrodes in view of the electrode impedance. However, they are different in their target uses and constraints. Generally, clothes are much thicker than the

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Y. G. Lim is with the Department of Oriental Biomedical Engineering, Sanji University, Wonju 220-702, Korea (e-mail: yglim@sangji.ac.kr).

K. K. Kim is with Interdisciplinary Program in Medical and Biological Engineering, Seoul National University, Seoul 100 799, Korea.

*K. S. Park is with the Department of Biomedical Engineering, Seoul National University, Seoul 100 799, Korea (e-mail: kspark@bmsil.snu.ac.kr).

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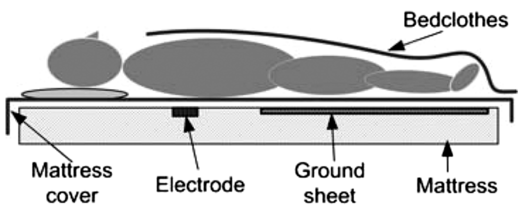

Fig. 1. Setup of the indirect contact ECG measurement system on a bed.

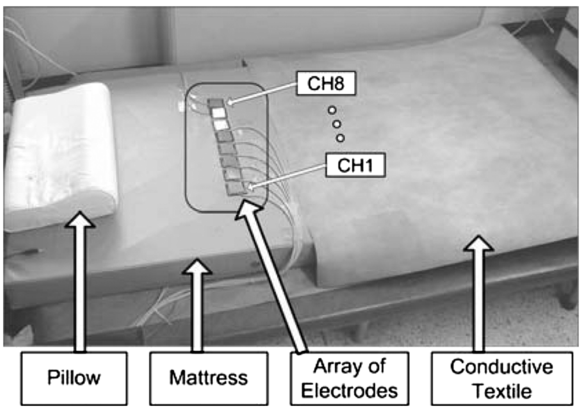

Fig. 2. Mattress with the array of electrodes and the ground plane.

insulating layer of an insulated electrode, which means higher electrode impedance. In addition, the through-clothes measurement aims at removing the need for direct contact between the bare skin and the instrument. The target uses also restrict the common electrode or the reference electrode to indirect contact with bare skin.

Figs. 1 and 2 show the setup for the presented ECG measurement method. An array of active electrodes was attached to a mattress in a row and a large conductive textile was laid on the lower area of the mattress. A mattress cover was placed over the mattress, electrodes, and the ground textile. ECG was measured with a subject lying on the mattress wearing normal pajamas.

In our previous study [13], we developed an ECG measurement method that enabled ECG recording through clothes without direct skin-contact while the subject was sitting on a chair. For ECG measurement during sleep without direct skin contact, we adopted active electrodes similar to those presented in our previous study and utilized a conductive textile as a reference electrode. However, the apparatus for the "ECG-on-bed" was different in some aspects, owing to differences in requirements and constraints.

A. Active Electrodes

Fig. 3 shows the profile of the active electrode. The active electrode was composed of three components; electrode face, preamp, and shield.

The electrode face was a conductive plate that sensed the potential variation on the skin. The electrode face was made on a square PCB (4 cm × 4 cm) clad with copper. For the sake of protecting the electrode face from corrosion, and the improvement of touch feeling for possible contact to bare skin, the electrode face was coated with an electrically resistive paint during PCB fabrication. The thickness of the resistive coat was measured to be about 0.01 mm.

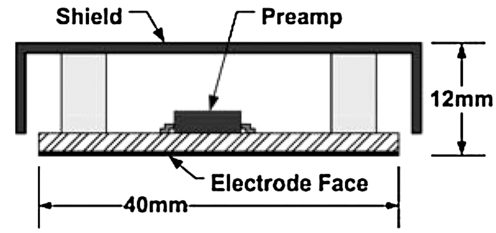

Fig. 3. Active electrode.

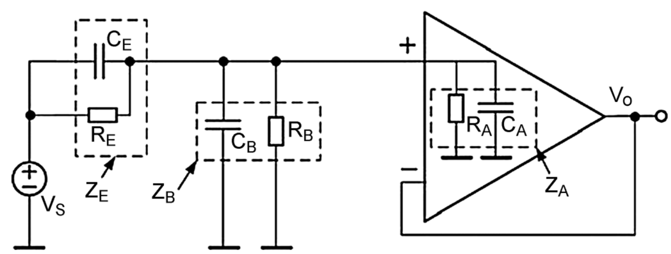

Fig. 4. Diagram of the active electrode including the electrode-skin interface model (Z_E).

ECG measurement through clothes requires a high-input-impedance amplifier because of the high impedance of the clothes. Therefore, a high-input-impedance amplifier was designed and embedded in the electrode. The preamp was mounted on the rear side of the electrode face. Fig. 4 shows the electric diagram of the active electrode including the electrode impedance Z_E and ECG source V_S. The circuitry of the preamp looks rather simple because there are only two actual components (an op-amp and a resistor R_B). However, in order to assess the characteristics of the electrode, we should consider C_B, which is the capacitance between the input (the electrode face and input line pattern) and the ground of the amplifier (including shield). Moreover, the electrode impedance Z_E should also be considered because it is comparable in amplitude to the electrode input impedance.

Screening external interferences was important to obtain good signal quality because the circuit impedance in the active electrode was very high. Therefore, the active electrode was shielded by a shield that surrounded the rear side of the electrode and was connected to the ground of the preamp.

B. Frequency Response of the Active Electrode

In the electrode model shown in Fig. 4, the parallel connection of C_E and R_E represents the electrode impedance between the electrode face and the skin. C_E is determined by the dielectric property and the thickness of the clothes, as well as the effective electrode area. Most clothes are generally regarded as insulators. However, in the presented method, clothes cannot be regarded as pure capacitors. We have observed many results that can be explained by the introduction of conductance. From our observations and common knowledge, we can be fairly certain that conductance depends heavily on moisture.

Based on the model, the gain G_S for the V_S (ECG source) is obtained by (1) and (2). The input impedance Z_A of the op-amp

OPA124 used in the presented active electrode is great enough to be disregarded in comparison with Z_B

$$G_S(s) = \frac{V_O}{V_S} = \frac{Z_B//Z_A}{Z_E + Z_B//Z_A} \quad (1)$$

$$G_S(s) = \frac{R_B + C_E R_B R_E s}{(R_B + R_E) + (C_B + C_E) R_B R_E s}. \quad (2)$$

C. ECG Measurement by Electrode Array

During sleep, people can move to any location on a bed and lay in various positions. Therefore, in order to measure ECG with electrodes fixed on the bed, we have the choice of using large or multiple electrodes. Capacitive electrodes produce a motion artifact due to variation in the capacitance [14], [15]. Therefore, the size of the active electrode is limited to keep good contact and to minimize the capacitance variation. Given these considerations, we used an electrode array for ECG measurement on a bed.

D. Indirect-Contact Ground

The electric potential measurement requires a reference. Standard ECG measurements use the potential of an electrode or the arithmetic mean potential of several electrodes as a reference [16]. Because no electrode keeps good contact with the body throughout the measurement time, and some electrodes may produce large motion artifacts at a given time, the arithmetic mean potential cannot be applied to the presented method. In addition, the high electrode impedance and the high electrode input impedance increase the influence of external interferences and power line interference. One of the effective ways to reduce the interference effect is to reduce the impedance between the body and the ground of the instrument [17], [18].

From the above considerations, an additional reference electrode was necessary for the presented method and its impedance to the body should be as low as possible. Moreover, the reference electrode should contact the body indirectly owing to the aim of this study. To meet these requirements, we designed an indirect contact ground in the form of a large conductive textile laid on the lower area of the bed to indirectly contact the lower body. The large effective contact area compensated for the high impedance per unit area to some degree. Because the indirect contact ground was laid under the lower body, the measured ECG signal would be similar to lead II or lead III.

III. Experiment Setup

The experiment setup for the indirect-contact ECG (IDC-ECG) measurement on a bed is described with three components: active electrodes, the mattress assembly, and electronics.

A. Active Electrode

As shown in Fig. 3, the size of the electrode face was 4 cm $\times$ 4 cm and the height of the electrode was 12 mm. The shield was made of aluminum plate. The capacitance (C_B in Fig. 4) between the electrode face and the shield was estimated to be about 18 pF by FEM simulation. Op-amp OPA124 (TI) was used in the preamp. The value of the resistor R_B, a path for bias current of the op-amp, was 1.6 GΩ. In (2), the resistance R_B affects the low-frequency gain and cutoff frequency. The higher resistance R_B results in a higher gain in low-frequency and a lower cutoff frequency. In the indirect contact measurement on a bed, a very large motion artifact caused by triboelectricity was expected. So, a 1.6-GΩ resistor, which is lower than the resistor in [13], was selected as R_B in order to prevent output saturation caused by large motion artifacts. However, the R_B value was selected rather intuitively with the observations in [13] and we did not closely evaluate the propriety of the selection. As a result of the lower resistance, ECG waveform distortion in low-frequency range was inevitable. The electrode capacitance of an insulator with a thickness of 1 mm and relative permittivity of 2 was estimated to be about 30 pF. For the pure capacitive electrode impedance of 30 pF, the cutoff frequency of (2) was expected to be about 2 Hz.

To assess the actual frequency response of the active electrode, the following experiment was carried out. A copper plate (20 cm $\times$ 20 cm) was laid on a table, connected to the output of a function generator, and then covered with the mattress cover and pajama used in the experiment. An active electrode was placed on the clothes. The output of the active electrode was measured

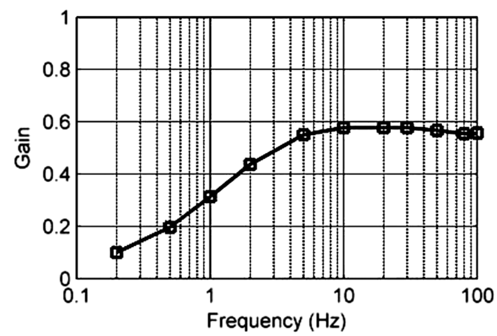

Fig. 5. Measured gain of the active electrode through a mattress cover (thickness 0.5 mm) and pajama (thickness 0.3 mm).

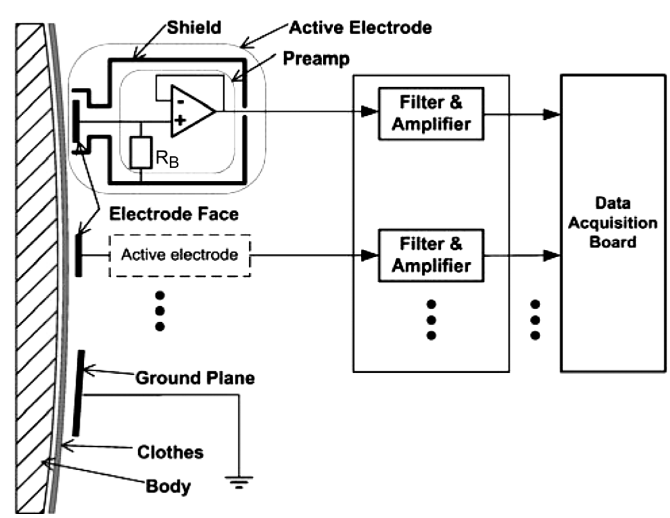

Fig. 6. Structure of the instrumentation system for the IDC-ECG on a bed.

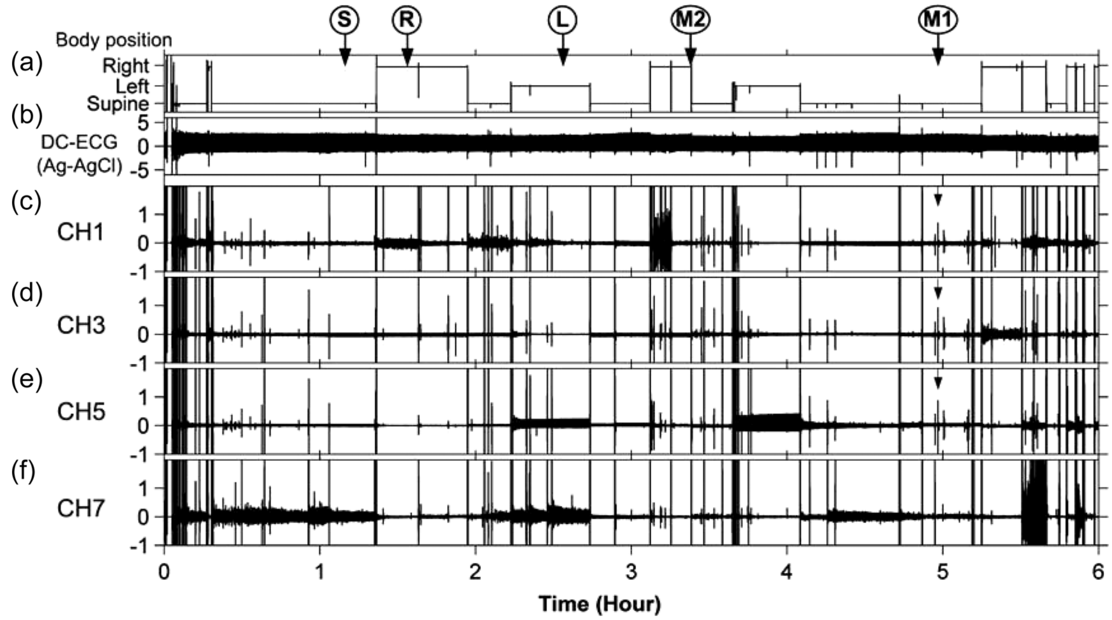

Fig. 7. Outputs obtained from two of the eight electrodes over a 6-h sleep period: (a) output of position sensor; (b) output of the DC-ECG using an Ag-AgCl electrode; (c)–(f) outputs of the electrodes CH1, CH3, CH5, and CH7, respectively. The arrows labeled with circled characters indicate the waveform times in Figs. 8 and 9.

by varying the output frequency of the function generator. The actual gain is shown in Fig. 5.

B. Mattress Assembly

To record the ECG regardless of body position on bed, we adopted a multiple-electrode instrumentation method. Fig. 2 shows the setup of presented method. Eight active electrodes were fixed on a mattress in a row with a center-to-center distance of 58 mm. The row of the electrodes was positioned at the height of the participant's ninth thoracic vertebra. The electrodes were named "CH1" to "CH8" starting from the participant's right. The mattress was filled with lumped urethane sponge chips and covered with artificial leather; its dimensions were $800 \times 1900 \times 150$ (W $\times$ L $\times$ T, mm). The electrodes were inserted in the mattress so that the electrode faces protruded about 3 mm from the mattress top. The lower area of the mattress under the participant's lower body was covered with a large conductive textile. A thin single-layer cotton mattress cover (thickness 0.5 mm) covered the mattress, the electrodes, and the ground plane. The participant wore cotton underwear (thickness 0.3 mm), cotton pajamas (thickness 0.3 mm), and lay on the mattress under bedclothes padded with cotton (150 cm $\times$ 180 cm, 3 kg).

C. Electronics and Data Acquisition

We prepared eight filter-and-amp modules for each electrode (Fig. 6). Potential variations on the participant's skin were sensed by the active electrodes through the clothes and mattress cover. Then, the signals from the electrodes entered their respective filter-and-amp modules. The filter-and-amp module was composed of filters and an amplifier: a second-order high-pass filter ($f_c = 2$ Hz), a notch filter (60 Hz), and a fourth-order low-pass filter ($f_c = 100$ Hz). The total gain of the filter-and-amp module was 300. The cutoff frequency of the high-pass filter was higher than the usual ECG amps. To prevent the possible output saturation caused by large motion artifacts, the cutoff frequency of the HPF was raised and the total gain was lowered. The outputs of the filter-and-amp modules were digitized by a commercial data acquisition board (NI DAQPad-6015, 16 bit) with a 500-Hz sampling rate. Finally, the digitized signals were filtered by a digital low-pass filter ($f_c = 40$ Hz).

The conductive textile and the instrument ground were connected to a metal water pipe of the radiator in the room for electrical safety [19].

D. ECG With Ag-AgCl Electrodes for Comparison

We carried out direct-contact ECG (DC-ECG) recording utilizing Ag-AgCl electrodes simultaneously with the IDC-ECG, as a reference during experiments. During simultaneous measurement of the IDC-ECG and the DC-ECG, unless the ground of the DC-ECG is isolated from the ground of the IDC-ECG, the signal quality of the IDC-ECG may be affected by the DC-ECG measurement. The reason for this effect is that the impedance between the body and the ground of the IDC-ECG is decreased by the low impedance between the body and the ground of the DC-ECG. Therefore, we prepared a custom-made isolated ECG amp for the DC-ECG measurement. A commercial isolation amplifier (TI 3656, isolation capacitance 6 pF) was applied to the isolation. The signal was filtered by analog and digital filters, so that its final bandwidth was 0.5 Hz–40 Hz and its gain was 3000. The common electrode (RL) of the DC-ECG was not driven-right-leg, but passive.

IV. Results

Experiments were carried out for six hours using the described IDC-ECG equipment. A position sensor (CMP, Compumedics Ltd., Victoria, Australia) was attached to the participant's chest to monitor his position. The output voltage

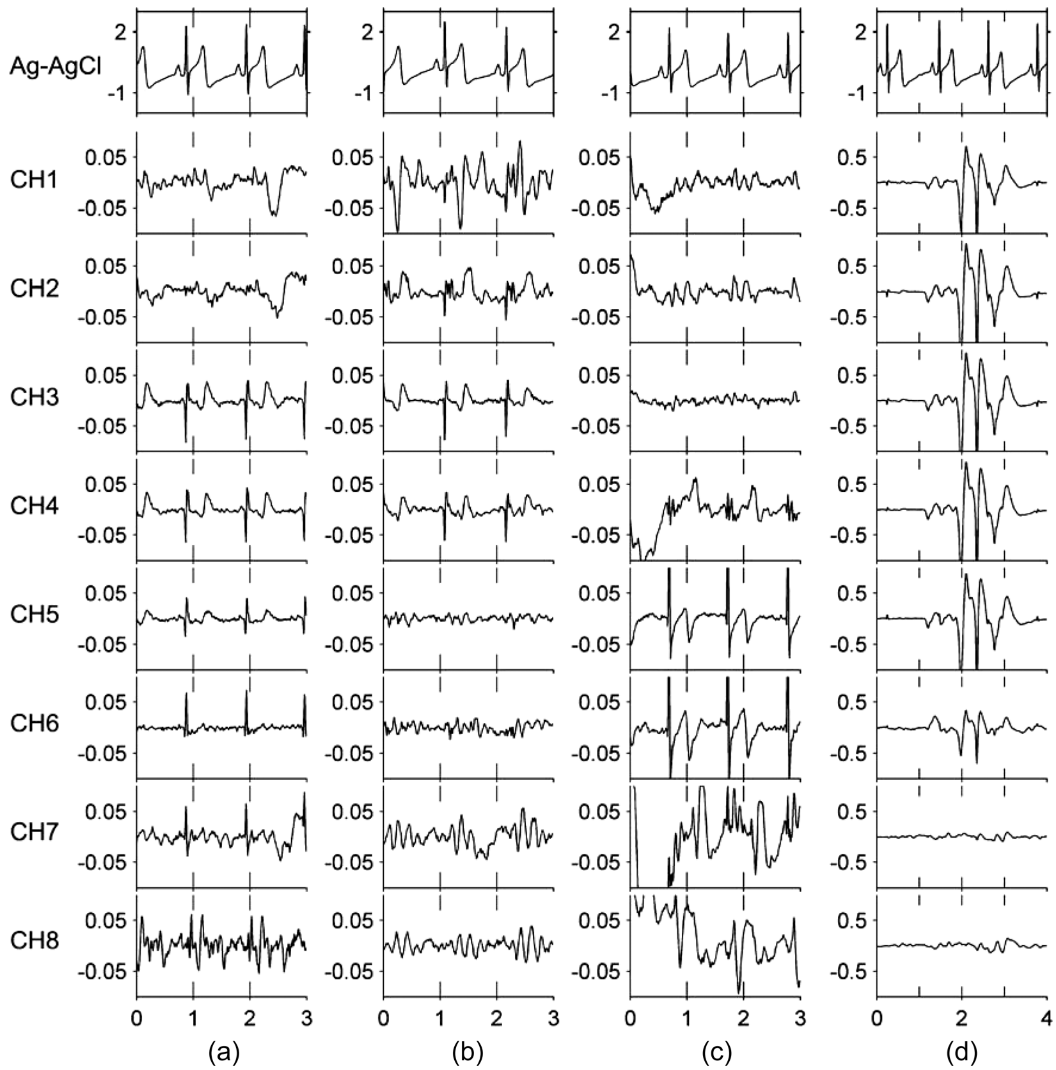

Fig. 8. ECG waveforms according to different body positions: (a) supine position; (b) on right side; (c) on left side; (d) supine position with a small limb movement. The times of each of the above images are indicated in Fig. 7 by circled characters; (a) "S", (b) "R," (c) "L," and (d) "M1."

of the position sensor varies according to body position. Moreover, conventional DC-ECG measurement using Ag-AgCl electrodes was carried out simultaneously with the IDC-ECG for comparison. An isolated ECG amplifier described above was used for DC-ECG. The Ag-AgCl electrodes were placed on the right clavicle $(-)$, on the left abdomen $(+)$, and on the right abdomen (common). Therefore, the DC-ECG waveform would be similar to standard Lead II. We verified that the isolated DC-ECG measurement and the position sensor did not make any visible effect on the IDC-ECG results.

Fig. 7 shows the outputs of the selected four electrodes for 6 h, with the outputs of the position sensor and the DC-ECG as references. Fig. 7 does not show the details of the waveforms because the waveforms are condensed with respect to time. The peaks with large amplitudes were caused by motion artifacts. The figure shows much more peaks in the IDC-ECG recording than in the DC-ECG recording. The peaks occurring without any change in body position may have been caused by movements of the limbs.

Fig. 8 shows the details of the ECG waveforms at the marked times in Fig. 7. Fig. 8(a) shows that the ECG waveforms are different from each other according to the contact positions on the participant's back. The outer electrodes (CH1, CH2, CH7, and CH8) show large artifacts and small ECGs because of their imperfect contact with the participant's body. Fig. 8(b) shows the waveforms obtained with the participant lying on his right side. The electrodes on the right gave better ECG results than those on the left. Fig. 8(c) shows the results of an opposite situation to (b). Fig. 8(d) shows an instance of artifacts caused by small limb movements without a body position change. Similar motion artifacts occurred at several electrodes that may have been attached well to the body. However, we can not see such a motion artifact in the DC-ECG (labeled "Ag-AgCl" in Fig. 8).

Fig. 9 shows examples of ECG waveforms during a position change. The participant changed his body position from on his right side to supine. As shown, while the R-peaks appeared in the output of CH3 and not of CH5 before the change, the R-peaks appeared in the output of CH5 after the position change. The large motion artifact lasting over 5 s at the time of position change resulted from the body movement in a large scale. In addition, relatively small artifacts are also shown lasting over ten seconds after the large artifact. We suppose that the small artifacts were caused by minute movements of body or bedding during settlement following the large movement.

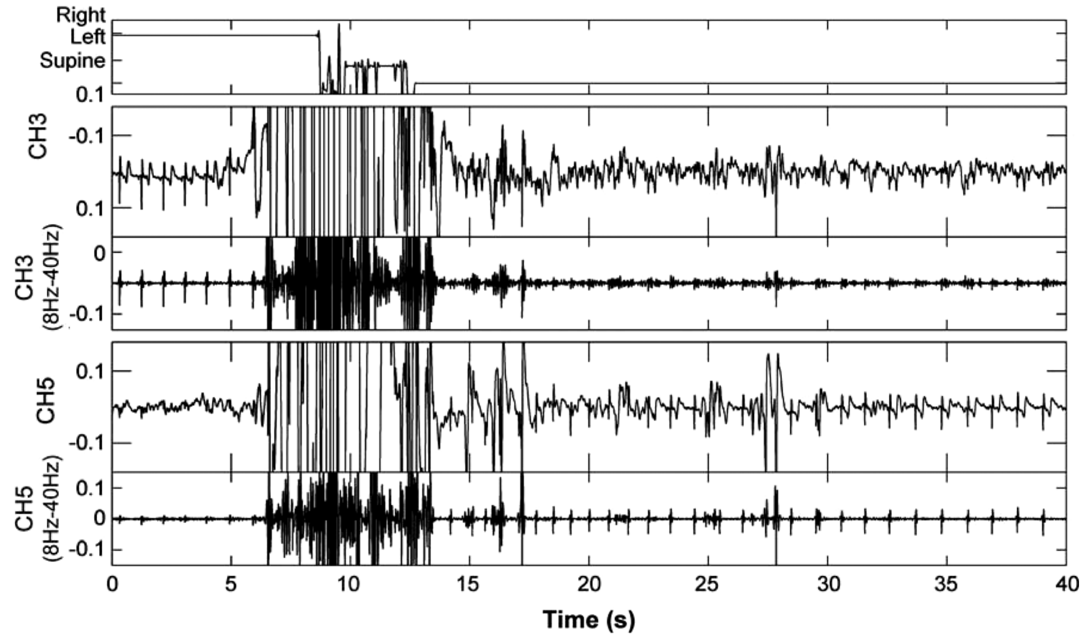

Fig. 9. Waveforms before and after a change in body position. Upper, body position; middle, output of CH3 and its filtered waveform; lower, output of CH5 and its filtered waveform. The time of the position change is indicated in Fig. 7 by circled characters "M2."

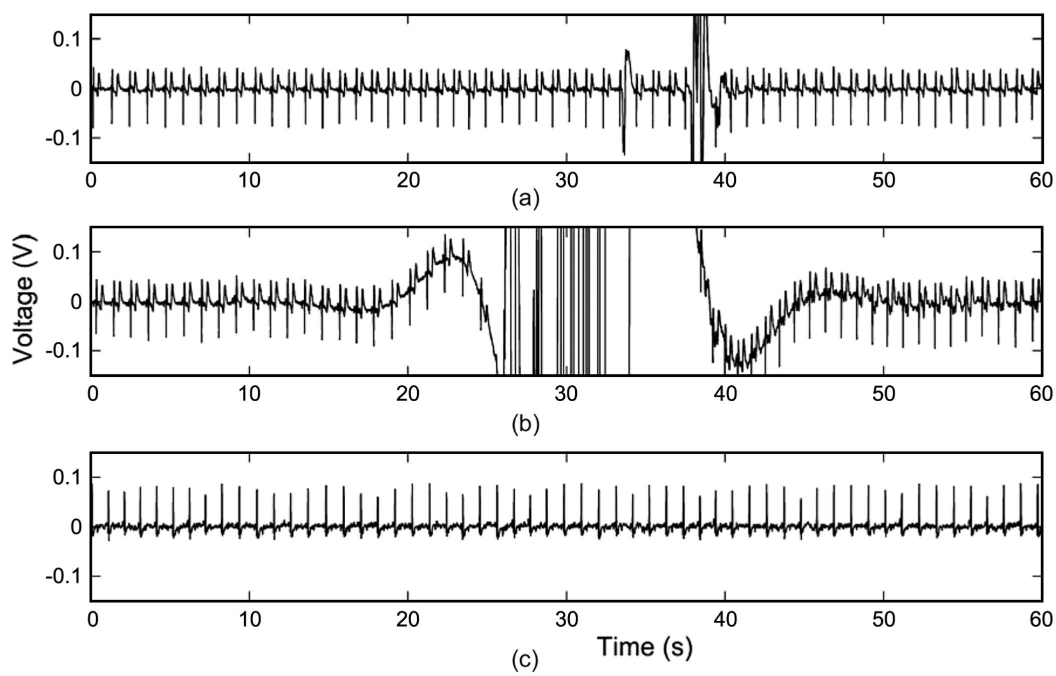

Fig. 10. Example waveforms extracted from the different electrode outputs shown in Fig. 7 at various times.

Fig. 10 shows several example waveforms of the IDC-ECG. Fig. 10(a) shows short duration motion artifacts that may have been caused by small limb movements. Fig. 10(b) shows a large motion artifact caused by a body position change. Fig. 10(c) shows an example of steady waveform lasting over 1 min.

V. Discussion

The ECG measurements through clothes show large motion artifacts because of the capacitive characteristics between the body and the electrode, and the large ground impedance of the body [13]. The IDC-ECG measurement on the bed also showed large motion artifacts. It is thought that there are two major origins of these artifacts. One is the variation in impedance between the electrodes and the body [14], [15], and the other is the variation in the whole body potential due to triboelectricity [20]. Unlike IDC-ECG on a chair, the nondifferential measurement method used in IDC-ECG on a bed increased the effect of body potential variation, resulting in it being the dominant factor. Fig. 8(d) shows an instance of the body potential variation; similar artifacts occurred in the multiple outputs. To reduce the triboelectricity, we investigated various clothes and we found that cotton produces the least motion artifacts. Thus, we used a mattress cover, bedclothes, and pajamas that were all made of cotton.

Though the frequency response presented in Fig. 5 is real, it is only a reference for the estimation of the real gain during the actual measurement situation. From our observations, it is almost certain that the actual gain with the body is higher than the gain with the copper plate, especially in the low-frequency range. This may be owing to the moisture content increase in the clothes due to perspiration.

At the early design stage, we expected that large motion artifacts would occur and their power would lie mainly in the low-frequency range ($<$10 Hz). For continuous measurement without saturation, the cutoff frequencies of the electrode and the HPF in the filter-and-amp module were raised, and the total gain of the instrument was lowered. However, it was observed that most of the motion artifacts caused by position changes or large movements of the limbs were so large as to saturate the amp even at a low gain. On the other hand, it was hard to discriminate ECG from most of the large artifacts even when they did not saturate the output. Therefore, it may be possible to raise the gain or to lower the cutoff frequency more than shown in the current design.

The ECG waveform acquired by the described IDC-ECG varied according to the contact position on the body, the contact condition, and the type and thickness of the clothes. This variation may be an obstruction to the diagnostic application of the waveform itself. However, further study may enable this application to be used for diagnosis in a restricted area or as an auxiliary method.

This method seems to be easily applied to applications using R-peaks, such as an assessment of HRV or simple heart rate. In addition, this method can provide R-peak positions as a reference time to other nonintrusive measurements.

Unlike the other conventional R-peak detection method, this method imposes the task of selecting good ECG signals from the eight outputs for R-peak detection. Therefore, it may require a 2-D approach, with one axis being time and the other axis the array of the electrodes.

VI. Conclusion

ECG measurement during sleep without direct conductive contact between the skin and electrodes was carried out with an array of high-input-impedance active electrodes and indirect-contact grounding. An ECG was recorded with distinct R-peaks during sleep, regardless of body position and location on the bed. The waveforms varied according to the contact condition and position. Therefore, further study on analyzing the waveform is needed for diagnostic applications that require ECG waveform information. Because of the nondifferential measurement and the triboelectricity in bedding, the ECG obtained by the presented method showed large motion artifacts. Therefore, careful preparation was required to reduce triboelectricity. This study shows the feasibility of using IDC-ECG for long-term daily ECG monitoring during sleep with minimal intrusion.

Acknowledgment

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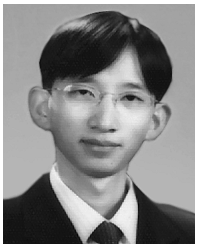
Yong Gyu Lim received the M.Eng. and Ph.D. degrees in biomedical engineering from Seoul National University, Seoul, Korea, in 1990 and 2006.

From 1995 to 2001, he was a Senior Researcher with the Samsung Advanced Institute of Technology and involved in the development of MRI (magnetic resonance imaging) system. He is currently a Professor with the Department of Oriental Biomedical Engineering, Sanji University, Wonju, Korea. His research interests include biomedical signal processing, nonintrusive biomedical instrumentation, and MRI system.



Ko Keun Kim received the B.Sci. degree in electrical engineering from Sun Moon University, Asan, Korea, in 2003. He is currently working towards the Ph.D. degree in interdisciplinary program in medical and biological engineering at Seoul National University, Seoul, Korea, since 2003.

His research interests include missing data problem on unconstrainedly measured biosignal for ubiquitous healthcare.

Kwang Suk Park (M'78) was born in Seoul, Korea, in 1957. He received the Ph.D. degree from the Department of Electronics Engineering, Seoul National University, Seoul, in 1985.

Since 1985, he worked in the field of biomedical engineering, especially for biological signal measurements and processing. Currently, he is running projects on ubiquitous healthcare mainly focusing on nonintrusive monitoring of biological signals. He is currently a Professor at Department of Biomedical Engineering, College of Medicine, Seoul National University. He is also the Director of Advanced Biometric Research Center in Seoul National University.