Communications

ECG Measurement on a Chair Without Conductive Contact

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Abstract—For the purpose of long-term, everyday electrocardiogram (ECG) monitoring, we present a convenient method of ECG measurement without direct conductive contact with the skin while subjects sat on a chair wearing normal clothes. Measurements were made using electrodes attached to the back of a chair, high-input-impedance amplifiers mounted on the electrodes, and a large ground-plane placed on the chair seat. ECGs were obtained by the presented method for several types of clothing and compared to ECGs obtained from conventional measurement using Ag–AgCl electrodes. Motion artifacts caused by usual desk works were investigated. This study shows the feasibility of the method for long-term, convenient, everyday use.

Index Terms—Active electrode, ECG monitoring, high-input impedance, indirect-contact electrode.

I. INTRODUCTION

HEALTH monitoring in daily life is attracting attention [1] and has led to many studies on routine electrocardiogram (ECG) measurements. These ECG measurements can be divided into two types according to where electrodes are attached or fixed. The first type involves measurements with conventional "fixed-on-body" electrodes such as Ag–AgCl electrodes, and the other involves measurements using electrodes installed on appliances or furniture [2], [3]. Although fixed-on-body electrodes are reliable and give good signal quality, they are inconvenient and inadequate for long-term, everyday measurements. In the latter case, measurements using "fixed-in-the-environment" electrodes are nonintrusive and adequate for long-term monitoring. However, although there are quality shortcomings, the nonintrusive nature of fixed-in-the-environment electrodes makes them an attractive option for daily monitoring.

The requirement of maintaining direct contact with bare skin limits the application of fixed-in-the-environment electrodes to a few cases. A study at the University of Sussex showed that ECG waveforms can be obtained using electrodes fixed at a distance of 5 cm from the skin, and that one can discriminate R-peaks without difficulty [4]. We have studied ECG measurements obtained without direct contact with bare skin and obtained practical results using a chair.

Fig. 1 illustrates the concept of making indirect-contact ECG measurements on a chair with a subject wearing normal clothes. The electrodes were attached to the back of the chair and were coupled capacitively with the skin through clothes, and a conductive plane was laid
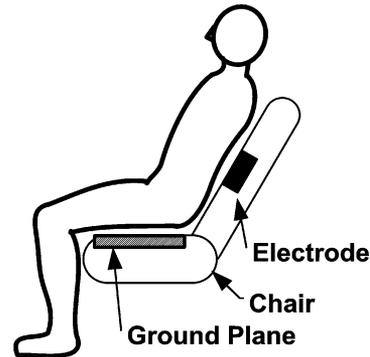
Fig. 1. Experiment setup used for ECG measurements using electrodes located on a chair.
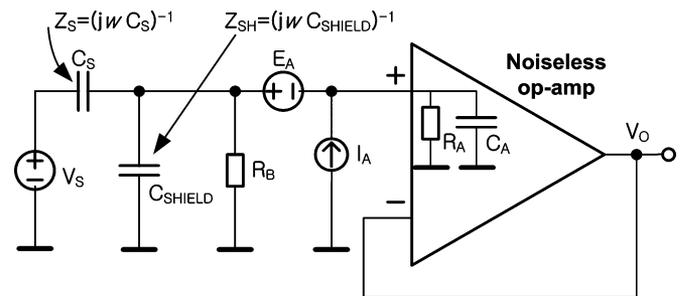

Fig. 2. Diagram of the amplifier used with an indirect-contact electrode (C_s) and signal source (ECG source, V_s).

on the seat of the chair to ground the body. Using this set-up, there is no direct conductive contact between the body and the instrument and ECG can be measured through clothes.

II. HIGH-INPUT-IMPEDANCE AMPLIFIER

Impedance between the skin and indirect-contact electrodes is very high compared with conventional direct-contact electrodes because of the insulating effect of clothes. Therefore, a high-input-impedance amplifier is required to amplify ECG signals through clothes.

In Fig. 2, C_S is the capacitance between the body and an electrode. The intrinsic noises of the op-amp are represented by a voltage noise source, E_A, and a current noise source, I_A [5]. R_B is a resistor for the bias current of the op-amp. C_{SHIELD} is the total capacitance between the input (electrode face and input circuitry) and the circuit ground. R_A and C_A are the input resistance and the capacitance of the op-amp, respectively.

The gain $G(s)$, derived in (1), has the form of a high-pass filter. The passband gain is determined by the ratio of C_S to the total capacitance including CS, and the cutoff frequency is affected by R_B.

$$G(s) = \frac{C_S R_B s}{1 + (C_A + C_{\text{SHIELD}} + C_S) R_B s}. \tag{1}$$

Equation (2) shows the SNR of the amplifier in Fig. 2 with the assumption that I_A and E_A are uncorrelated with each other [5].
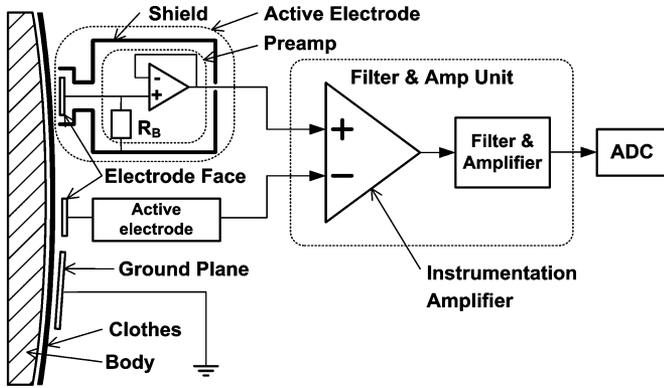
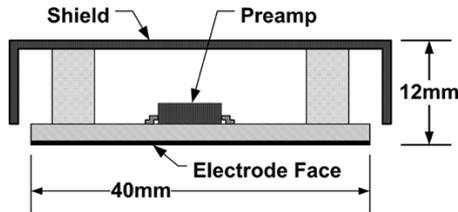Fig. 3. Block diagram of the indirect contact ECG measurement.

Fig. 4. Configuration of the active electrode.

E_{THERMAL} in (2) is the thermal noise voltage caused by the resistor R_B

$$\mathrm{SNR} = \frac{|V_S|}{\sqrt{I_A^2|Z_S|^2 + E_A^2\left|1 + \frac{Z_S}{Z_{\mathrm{SH}}} + \frac{Z_S}{R_B}\right|^2 + E_{\mathrm{THERMAL}}^2\left|\frac{Z_S}{R_B}\right|^2}}. \quad (2)$$

Equations (1) and (2) show that the characteristics of the measurements through clothes depend largely on the source impedance Z_S.

III. EXPERIMENTAL APPARATUS

Fig. 3 shows the whole measurement setup. Two ECG signals as potential variations on skin were sensed through clothes by active electrodes. Then, the difference between the two signals was filtered and amplified. A large seat-mounted conductive sheet grounded the body without contact with skin.

A. Active Electrode

The active electrode is composed of an electrode face, a preamp, and a shield, as shown in Fig. 4.

1) Electrode Face: The electrode face senses potential variations on skin. Although the electrode face is called an electrode in general, we refer to it as an electrode face in this paper, i.e., as a part of the active electrode. It was designed as a square plate (4 cm × 4 cm), made of PCB clad with copper and plated with gold. The thickness of the clothes was regarded to be 1 mm with a relative permittivity of 2. Thus, the capacitance between the body and the electrode face was estimated to be 30 pF, and its impedance at 20 Hz was estimated to be 0.25 GΩ.

2) Preamp: The electrode face was directly connected to a high-input-impedance preamp. The preamp was mounted on the other side of the PCB, as shown in Fig. 4. An op-amp TI OPA124 was used in the preamp and the preamp had unity gain. The input resistance and the input capacitance of the OPA124 were 10^{13} Ω and 1 pF, respectively. A

resistor (R_B) was connected to the input pin of the op-amp to provide a path of bias current; its resistance was 3 GΩ.

3) Shield: The shield was made of aluminum with a height of 12 mm from the electrode face, as shown in Fig. 4. The shield surrounded the rear of the electrode face. The capacitance, C_{SHIELD}, between the shield and the electrode face was estimated to be 15 pF by FEM simulation.

B. Filter and Amp Unit

Two output signals from each active electrode entered the instrumentation amplifier in the "filter and amp unit" (Fig. 3). Then, the difference signal from the instrumentation amplifier output was filtered and amplified. A high-pass filter, a notch filter, and a low-pass filter were designed in the filter and amp unit. The final passband of the filter and amp unit was from 0.5 Hz to 35 Hz and the total gain was 5000.

C. Chair and Grounding

An office chair with a back and armrests that could rock back and forth and swivel was used in this experiment. The seat and back of the chair were cushioned and covered with artificial leather.

A sheet of conductive textile (45 cm × 30 cm) was laid on the seat of the chair as a ground plane. The impedance between the body and this ground plane was estimated to be in the range of 500 k$\Omega \sim 10$ MΩ, as this was dependent on the clothes worn, the contact area, and the cushion condition.

D. Measurement Setup Used to Determine Frequency Response of the Active Electrode

From (1) and (2), the frequency response of the electrode with regard to clothing is needed to evaluate system performance. Accordingly, a means of measuring the frequency response of the electrode was designed. A copper plate (20 cm × 20 cm) was laid on a table, connected to a function generator, and then covered with sample cloth. An active electrode was placed on the cloth, and the cloth was pressed with a pressure of about 800 N/m^2 by a weight placed on the electrode. The output of the active electrode was measured with varying the output frequency of the function generator from 0.1 Hz to 500 Hz.

IV. RESULTS

Electrodes were attached to the back of the chair 12 cm apart (between centers) so as to be adjacent to the subject's twelfth rib. Fig. 5 shows the ECG obtained with the presented method. For comparison purposes, a conventional ECG measurement was also carried out on the same subject using two Ag–AgCl electrodes attached to the skin in the same positions on the back and using another Ag–AgCl electrode attached to the left calf as a common electrode (Fig. 6). Fig. 5 shows that the ECG waveform and the noise level depended on the clothing worn.

Fig. 7 shows a waveform obtained when the subject was swinging the upper part of his body back and forth. The subject was wearing a regular business suit and leaning back in the chair. The electrodes maintained contact with the body, and the motion of the body varied the pressure applied to the electrodes and clothes. The waveform in (a) is in the 0~35 Hz band and that in (b) is in the 8 Hz~35 Hz band.

Fig. 8 shows the frequency responses of the electrodes for different types of clothing. The frequency response obtained shows the features of a high-pass filter, though it depended on the clothing. The PTFE plate shows the characteristics typical of a 20 pF capacitor, whereas the others show complex characteristics.

To investigate the degree to which the common mode rejection ratio (CMRR) of the measurement system was responsible for the low signal
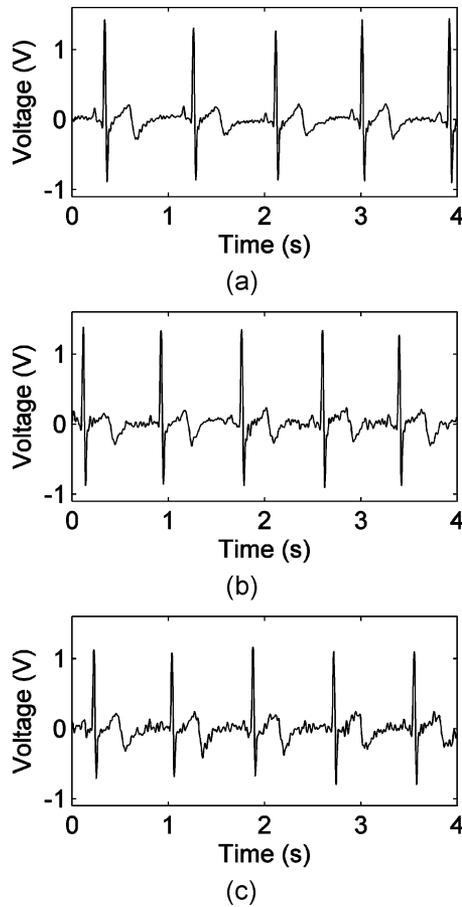
Fig. 5. ECG waveforms obtained by the presented method; (a) through a cotton shirt (twill weave, thickness 0.33 mm), (b) through a woolen business suit (thickness 0.37 mm) over a cotton dress shirt (thickness 0.22 mm), and (c) through a acrylic shirt (twill weave, thickness 0.65 mm). In all cases, the subject wore cotton underwear. Thickness gauge: Mitutoyo 7301.
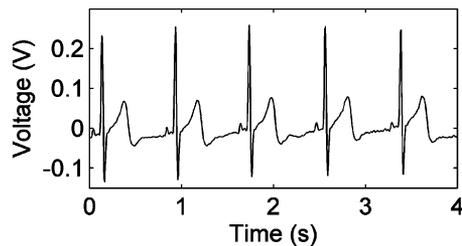

Fig. 6. ECG waveform obtained with conventional Ag–AgCl electrodes attached to the back skin of the same subject as in Fig. 5 (BIOPAC MP150, gain 1000, 0.5 Hz~35 Hz).

quality, a simple measurement of the CMRR was carried out. Both of the electrodes were connected to a function generator and the output voltage of the measurement system was measured for frequencies from 0.1 Hz to 40 Hz. The CMRR was found to be larger than 80 000:1 in this frequency range. Though the measurement setup was somewhat different from that described in [6], the CMRR of the experimental ECG measurement device used here was higher than required in [6]. These results show that differential noise transformed from common-mode noise by the measurement system may not be the dominant factor of the low signal quality.
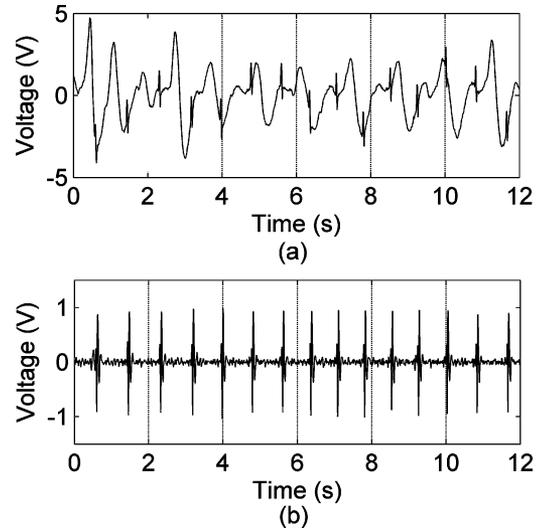

Fig. 7. ECG recording obtained with the subject swinging the upper body back and forth, wearing a regular woolen business suit, (a) 0.5 Hz~35 Hz, (b) 8 Hz~35 Hz.
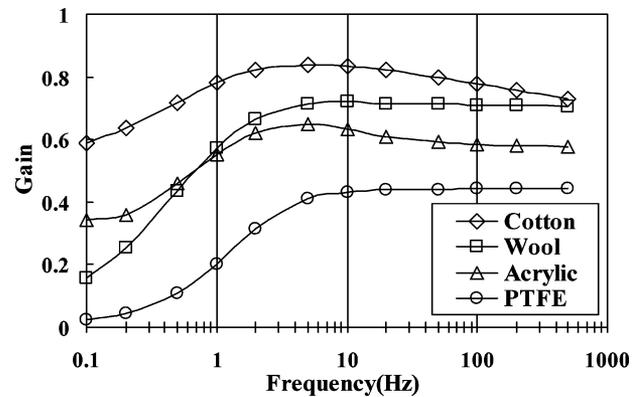

Fig 8. Frequency responses of the active electrode for different types of clothing and for a PTFE plate: cotton (twill weave, thickness 0.6 mm), wool (twill weave, thickness 0.55 mm), acrylic (twill weave, thickness 0.65 mm), and PTFE plate (thickness 1.5 mm, relative permittivity 2).

V. DISCUSSION

Differences in noise level due to clothing type means that the noise level largely depends on clothing properties. Equation (2) shows that one of the main factors is the impedance of the cloth. The ECG for the cotton shirt, Fig. 5(a) showed higher signal quality than the acrylic shirt, Fig. 5(c). We can infer from Fig. 8 that the impedance of the cotton shirt should be less than that of the acrylic shirt. These facts are consistent with (2): higher SNR with lower source impedance. The T-waves of the ECGs in Fig. 5 and Fig. 6 appeared different from each other. This difference reflects frequency response differences.

Fig. 7 shows an example of motion artifacts due to pressure variation by body motion. As shown, R-peaks can be easily discriminated from the motion artifacts because most of the motion artifact power lies under 10 Hz. Keyboard typing, PC mouse operation, and other moderate hand motions on a desk lead to this sort of motion artifact. However, other types of motion artifact with high-frequency components were also observed, e.g., those due to chair vibrations or lateral movements of skin.

A gradual decrease in motion artifact was observed over a few minutes after the subject sat on the chair, which may have been caused by the increase in moisture in clothes due to sweating. Furthermore, a decrease in noise level was observed when the moisture in clothes increased. These observations show the notable influence of moisture on the measurement necessitating further study on the effects of humidity.

Though the signal quality obtained was poorer than that obtained using conventional methods, the presented ECG measurement method has the substantial advantage of being easily used on a daily basis. The method can be used for daily ECG monitoring as an auxiliary diagnostic device, or for a long-term HRV measurement.

VI. CONCLUSION

We described a method of taking ECG measurements that does not rely on direct skin contact. The method utilizes high-input-impedance active electrodes and indirect-contact grounding. The signal quality of the presented method was lower than those of conventional methods and was dependent on clothing properties. However, our results demonstrate the potential of this technique for long-term, convenient, everyday use.

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Linear Minimum Mean-Square Error Filtering for Evoked Responses: Application to Fetal MEG

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Abstract—This paper describes a linear minimum mean-squared error (LMMSE) approach for designing spatial filters that improve the signal-to-noise ratio (SNR) of multiepoch evoked response data. This approach does not rely on availability of a forward solution and thus is applicable to problems in which a forward solution is not readily available, such as fetal magnetoencephalography (fMEG). The LMMSE criterion leads to a spatial filter that is a function of the autocorrelation matrix of the data and the autocorrelation matrix of the signal. The signal statistics are unknown, so we approximate the signal autocorrelation matrix using the average of the data across epochs. This approximation is reasonable provided the mean of the noise is zero across epochs and the signal mean is significant. An analysis of the error incurred using this approximation is presented. Calculations of SNR for the exact and approximate LMMSE filters and simple averaging for the rank-1 signal case are shown. The effectiveness of the method is demonstrated with simulated evoked response data and fetal MEG data.

Index Terms—Linear minimum mean square error (LMMSE), magnetoencephalography (MEG), spatial filter.

I. INTRODUCTION

The most common method of increasing the signal-to-noise ratio (SNR) of evoked response data is averaging; however, this simplistic approach is often inadequate when the number of trials and/or the SNR is low, which is often the case for magnetoencephalography (MEG) and electroencephalography (EEG) signals. An extreme example of a low SNR signal is the fetal magnetoencephalogram (fMEG) [1]–[3]. In the last few years, MEG has been increasingly utilized to study the development of brain activity in the fetus, as well as the neonate. Fetal recordings are arguably the most difficult of all MEG signals to record due to their very low amplitude and the presence of strong cardiac interference from both the fetus and mother.

Recent attempts to improve the SNR of the fMEG have centered on spatial filtering techniques [4], [5]. MEG recording systems typically allow for acquisition of many channels, and the sensor covariance exhibits considerable spatial structure, which can be exploited with suitable signal processing techniques. Many spatial filtering methods, such as linearly constrained minimum variance spatial filtering [6], require knowledge of the forward solution; however, this is problematic for fMEG due to the lack of a simple, accurate source model. In the absence of a forward solution, more general methods such as principle component analysis (PCA) and maximum-likelihood estimation (MLE) [7] can still be employed. PCA exploits the low-rank spatial structure of the signal, but requires rank determination and whitening to be effective [8]. MLE exploits low-rank spatio-temporal structure and inherently accommodates spatially colored noise, but also requires rank determination.