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Capacitive ECG Measurement by Conductive Fabric Tape

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***Abstract*—**Capacitive electrocardiogram (ECG) measurement is an attractive approach for long-term health monitoring. However, there is little literature available on its implementation, especially for multichannel system in standard ECG leads. This paper begins from the design criteria for capacitive ECG measurement and presents a multichannel limb-lead capacitive ECG system with conductive fabric tapes pasted on a double layer PCB as the capacitive sensors. The proposed prototype system incorporates a capacitive driven-body (CDB) circuit to reduce the common-mode power-line interference (PLI). The presented prototype system has been verified to be stable by theoretic analysis and practical long-term experiments. The signal quality is competitive to that acquired by commercial ECG machines. The feasible size and distance of capacitive sensor have also been evaluated by a series of tests. From the test results, it is suggested to be greater than 60 cm^2 in sensor size and be smaller than 1.5 mm in distance for capacitive ECG measurement.

***Keywords*—**capacitive driven-body (CDB) circuit, capacitive electrocardiogram (ECG) measurement, capacitive sensor, conductive fabric tape, power-line interference (PLI).

I. INTRODUCTION

BIOPOTENTIAL measurement is crucial for clinical diagnosis and medical research. Conventionally, the Ag/AgCl electrodes with conductive gel pasted on body surface were required for bioelectric events recording. Such approach is widely adopted and reliable, but it is time-consuming for electrodes preparation and is not beneficial in long-term monitoring for the possible irritation and skin allergy.

Non-contact measurement by capacitive sensors provides an alternative approach that may overcome such deficiencies. Capacitive ECG measurement was firstly proposed by Lopez and Richardson, which was originally developed for space use [1]. Thereafter, several studies have been devoted to the researches of capacitive ECG measurement [2-7].

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To date, capacitive ECG has been implemented on chair [4] or on bed [6, 8], and the acquired signals have also been compared with those from Ag/AgCl electrodes [9]. Multichannel capacitive ECG system has also been developed by arranging the capacitive sensors in an array style [3, 7]. However, little information has been published concerning the implementation of multichannel capacitive ECG system in standard leads (i.e. leads I, II, III, aVL, aVR and aVF) and the measurement comparisons with commercially available ECG machine. The long-term stability testing results are rarely available, too. This paper presents a prototype of multichannel ECG system with the conductive fabric tape pasted on a two-layer PCB as the capacitive sensors. The implementation is not too difficult to be tackled in an ordinary laboratory. The derived signals were compared with those acquired by commercial ECG equipments and the signal quality is competitive. The system was also tested in long-term experiments and is shown to be stable. The mean-square errors (MSE) for various capacitive area and distance between the human body and the conductive fabric tape are also demonstrated to depict the feasible capacitive area and distance for such system. The proposed system could be helpful for those who are interested in implementing one capacitive ECG measurement system for their specific researches. The results reported here may also be useful in evaluating the utility of capacitive ECG in clinical medicine.

II. METHODS AND MATERIALS

A. Design Criteria

The schematic diagram for a single-channel capacitive biopotential measurement is depicted in Fig. 1. In this illustration, $C_{1,2}$ (C_1 and C_2) denotes the capacitance resulted from the capacitive sensors. $Z_{b1,2}$ (Z_{b1} and Z_{b2}) represents the equivalent impedance with respect to the circuit common at the input node which is the impedance of bias network (if it is included in the design) in parallel with the common-mode input impedance of the input-stage buffer. $Z_{b1,2}$ can be simplified by a parallel R-C network (denoted as $R_{b1,2} \parallel C_{b1,2}$) which is usually at the order of several to thousands of GΩ ∥ pF.

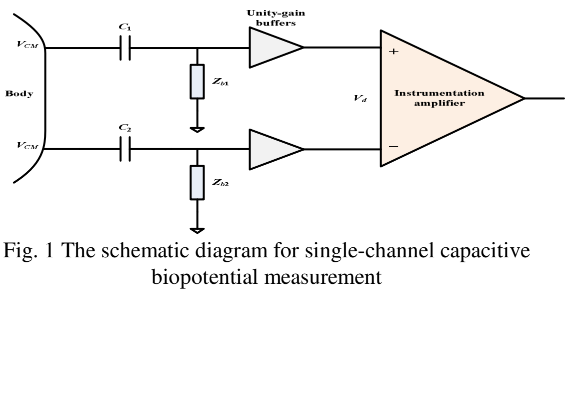


Fig. 1 The schematic diagram for single-channel capacitive biopotential measurement

The network formed by $C_{1,2}$ and $Z_{b1,2}$ is inherently a high-pass filter, and its −3dB cutoff frequency can be approximated as

$$f_{-3dB} = \frac{1}{2\pi \cdot R_{b1,2} \cdot (C_{b1,2} + C_{1,2})}. \tag{1}$$

The values of $C_{1,2}$ and $Z_{b1,2}$ should be selected properly such that the interested signal components could be kept in the pass-band. For ECG measurement, f_{-3dB} should be lower than 0.2 Hz such that the low-frequency components buried in ECG signal can be acquired. By such requirement, equation (1) becomes

$$0.2 \geq \frac{1}{2\pi \cdot R_{b1,2} \cdot (C_{b1,2} + C_{1,2})}. \tag{2}$$

As $C_{b1,2}$ is usually much smaller than $C_{1,2}$, and thus $C_{b1,2}$ can be omitted in (2) without loss of generality. In this study, the relative permittivity of the dielectric material, which is the clothes made from purified cotton, is approximately 1.3. Assuming that the capacitance $C_{1,2}$ is much greater than $C_{b1,2}$, the following constraint can be derived thereby

$$1.44 \times 10^{-11} \geq \frac{d_{1,2}}{R_{b1,2} \cdot A_{1,2}}, \tag{3}$$

where $A_{1,2}$ (unit: m^2) and $d_{1,2}$ (unit: m) represent the area and distance of capacitive sensor ($C_{1,2}$) respectively. This formula is utilized for evaluating the feasibility of input-stage circuits which incorporate capacitive sensors.

As $C_{1,2}$ and $R_{b1,2}$ usually exist unavoidable mismatches, the relationship between the differential-mode input signal (V_d) for the instrumentation amplifier and the common-mode body potential (V_{CM}, essentially the interference coupled from power lines) is derived by the voltage-divider effect as

$$V_d = \frac{j \cdot 2\pi \cdot f \cdot (R_{b1} \cdot C_1 - R_{b2} \cdot C_2)}{(1 + j \cdot 2\pi \cdot f \cdot R_{b1} \cdot C_1)(1 + j \cdot 2\pi \cdot f \cdot R_{b2} \cdot C_2)} V_{CM}. \tag{4}$$

The frequency f in the above equation is 50 or 60 Hz (depending on the power system) in essence. In general condition, $2\pi \cdot f \cdot R_{b1,2} \cdot C_{1,2} >> 1$.

It is assumed that the mismatches of $C_{1,2}$ and $R_{b1,2}$ can be represented as

$$C_1 = C + \Delta C,\ C_2 = C - \Delta C,\ R_{b1} = R + \Delta R,\ R_{b2} = R - \Delta R, \tag{5}$$

with $C >> \Delta C$ and $R >> \Delta R$. Equation (4) can then be simplified to be

$$V_d \approx \frac{1}{2\pi \cdot f \cdot R \cdot C}\left(\frac{2 \cdot \Delta C}{C} + \frac{2 \cdot \Delta R}{R}\right) V_{CM}. \tag{6}$$

This is the interference coupled from power lines that may corrupt the signal quality. For the biopotential measurement with acceptable quality, such interference should be limited to be less than 1% of the biopotential amplitude [10]. That is, the differential-mode biopotential to be measured (denoted as V_{body}) and the interference V_d should satisfy the following inequality

$$V_{body} \geq 100 V_d = \frac{100}{2\pi \cdot f \cdot R \cdot C}\left(\frac{2 \cdot \Delta C}{C} + \frac{2 \cdot \Delta R}{R}\right) V_{CM}. \tag{7}$$

For ECG measurement, the typical value of V_{body} ranges from 0.5 to 4 mV, and V_{CM} may deviate from 50 μV to 10 mV, depending on the measurement environment and whether the driven-right-leg (DRL) circuit is included in the system or not [11-12]. Let's consider the worst case, that is V_{body} = 0.5 mV and V_{CM} = 10 mV. The constraints on R, C and their deviations for capacitive ECG measurement can then be derived directly from (7), which is

$$\frac{\pi \cdot f}{1000} \geq \frac{1}{R \cdot C}\left(\frac{2 \cdot \Delta C}{C} + \frac{2 \cdot \Delta R}{R}\right). \tag{8}$$

Equations (3) and (8) constitute the basic design criteria for the capacitive ECG system that should be satisfied.

B. Material for Conductive Fabric Tape

The conductive fabric tape utilized in this research is a kind of metal fiber of low resistance (code 85773, Taiwan Golden Enterprise®). This material features with low resistance (< 0.1 Ω/m), high elongation (30±5%), softness, thinness (0.13 mm) and light in weight (72±7 g/m^2). It is coated by conductive adhesive base (< 0.1 Ω/m) on one side and is persistent for wide operating conditions. The material is easy to be cut by scissors and can be connected tightly to the copper wire on PCB by solder.

C. Experiments

The following experiments were conducted to verify the performance of the proposed system in this research.

Experiment 1: The ECG patterns were measured by standard equipments with conventional Ag/AgCl electrodes and the implemented prototype system simultaneously for a healthy male subject aged 25. In the tests, leads I, II and III were compared with those acquired by BioPac MP 150™, whereas leads aVR, aVL and aVF with those by Fukuda FCP-145U™.

Experiment 2: To evaluate whether the proposed system is feasible for long-term measurement, the ECG monitoring for a healthy male subject aged 30 was performed during sleep. In this experiment, the system was inlaid in a mattress. And, the printed side of the PCB (with conductive fabric tapes) was put upwards such that the supine subject in sleep could contact the sensors directly on his back.

Experiment 3: As indicated in (3) and (8), the capacitances resulted from capacitive sensors may influence system characteristics. It is necessary to evaluate the feasible area and distance of capacitive sensors for ECG monitoring in practice. The ECG simulator (PS400, Fluke®) is utilized as signal source at various sensor areas and distances for this experiment. Before further performance evaluation, the measured signals were normalized according to the following process.

Let $x[n]$ (n=1, 2, ⋯, N) represent the acquired signal. Firstly, find the maximum and minimum value of $x[\cdot]$, and denote them by $x_{\max}$ and $x_{\min}$ respectively. Then each $x[\cdot]$ is normalized by the following formula

$$\bar{x}[n] = \frac{x[n] - x_{\min}}{x_{\max} - x_{\min}} \quad \text{for } n = 1, 2, \cdots N. \tag{9}$$

The normalized data for signal source and the measured pattern are denoted as $\bar{x}_S[\cdot]$ and $\bar{x}_D[\cdot]$ respectively. The mean-square errors (MSE) between $\bar{x}_S[\cdot]$ and $\bar{x}_D[\cdot]$ at various parameters (area and distance) are adopted to evaluate the distortion. The computation of MSE is as below

$$\mathrm{MSE} = \frac{1}{N}\sum_{n=1}^{N}\left(\bar{x}_S[n] - \bar{x}_D[n]\right)^2. \tag{10}$$

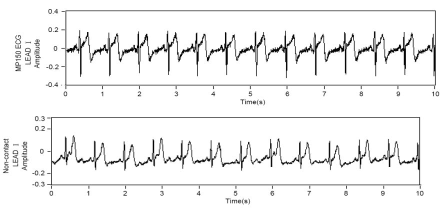
For all of the experiments, the temperature is kept at 20°C and the relative humidity is 55%. The clothes for the subjects are made from purified cotton. The capacitive sensors constructed by conductive fabric tapes are well-controlled to be of the same size in each experiment. Thus, the criterion (8) can be assumed to be satisfied in this research. Each clothes utilized in the experiments has a thickness of 0.5 mm. For the experiments on the effect of different capacitive distance, the thickness was increased by putting more clothes on the subject.

DC battery is utilized as the voltage source in this research. In addition, the capacitive driven-body (CDB) circuit similar to that proposed in [5] is also included in this prototype system.

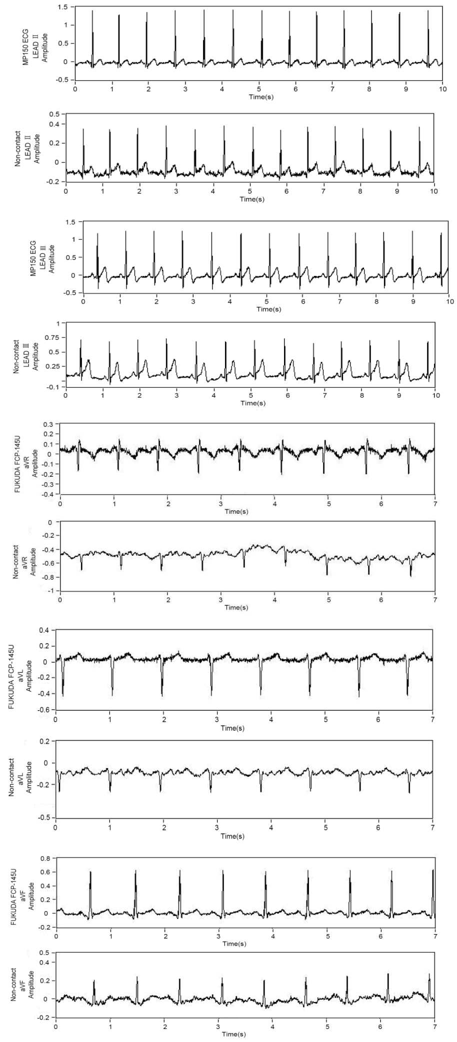
III. Results and Discussion

For *Experiment* 1, the results compared with those derived by BioPac MP 150™ are demonstrated in Fig. 2 (a)-(c) for lead I-III, and those by Fukuda FCP-145U™ are shown in Fig. 2 (d)-(f) for augmented limb leads respectively. From the results, it can be appreciated there exists tiny deviation in morphology and usually more interfered in capacitive measurement. However, the specific features in ECG such as QRS complex and T wave can still be observed. The tiny differences might be resulted from the characteristic deviation on frequency response among the equipments. The lead vectors formed by contact positions on the back of subject might deviate slightly from the standard ones and this may also lead to results with tiny difference. As the area of capacitive sensor is much larger than that of contact Ag/AgCl electrode, it is reasonable that the capacitive measurement system may pick up more displacement current from power mains and this will lead to more evident 50/60-Hz interference.

For *Experiment* 2, the result of three-hour duration is shown in Fig. 3 (a). In this figure, the burst patterns with abrupt change in polarity and enhancement in amplitude (such as segment A marked by dashed box) denote the conditions of subject movement, and the remaining portions represent the case in quiet status (such as segment B marked by dashed box). Fig. 3 (b) and (c) demonstrate portions of ECG pattern in segments A and B, respectively. One major drawback of capacitive ECG system is its susceptibility to motion artifact and this can be found in Fig. 3 (a) and (b). To our knowledge, suppressing the motion artifact remains to be a challenging task in capacitive biopotential measurement. In spite of this deficiency, it can be appreciated that the ECG features can be clearly observed for the signals derived in quiet condition, as fig. 3 show.



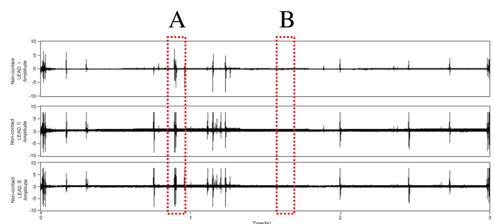
(a)



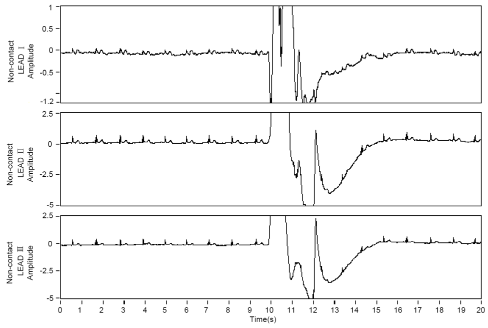
(b) (c) (d) (e) (f)

Fig. 2 Comparisons of ECG patterns acquired by commercial equipments and the presented system. (a)-(c) For lead I, II and III compared with BioPAC MP150™. (d)-(f) For leads aVR, aVL and aVF, compared with Fukuda FCP-145U™

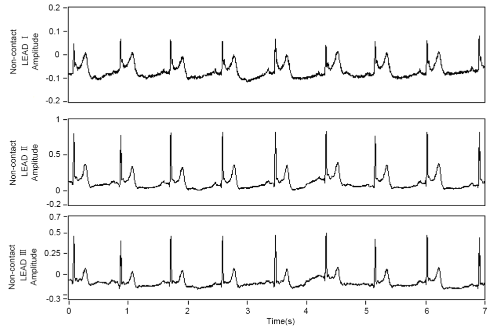
For *Experiment* 3, the MSE versus different tape areas is shown in Fig. 4 (a), and the MSE versus various capacitive distances is illustrated in Fig. 4 (b). From (3), the sensor area should be large and moreover the distance should be small such



(a)

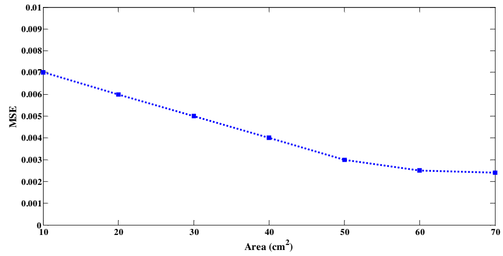


(b)

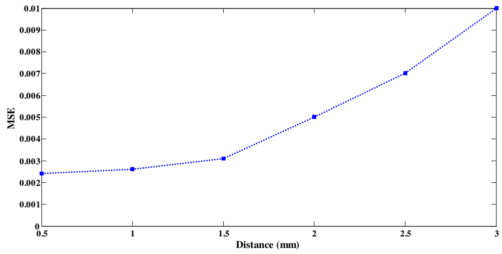


(c)

Fig. 3 ECG monitoring during sleep. (a) ECG of three-hour period. (b) ECG of twenty-second period in segment A indicated in (a). (c) ECG of seven-second period in segment B indicated in (a)



a



b

Fig. 4 The measurement mean-square error (MSE) at different (a) area and (b) distance of the capacitive sensor

that this design criterion could be satisfied. In our prototype, the low-noise high-precision Difet® operational amplifier, OPA124 (Burr-Brown®) is adopted for the input buffer. It has a very high common-mode input resistance (typically 10^{14} Ω). Equation (3) can thus be reduced to be

$$1.44\times10^{3} \geq \frac{d_{1,2}}{A_{1,2}}. \tag{11}$$

Theoretically, this criterion can be satisfied even at $d_{1,2}$=3 mm and $A_{1,2}$=10 cm^2. As illustrated in Fig. 4 (a) and (b), the MSE approaches the minimum asymptote as the area is larger than 60 cm^2 and the distance is smaller than 1.5 mm. It can also be appreciated that the value of MSE decreases with increased area, and the trend is reversed for increasing distance.

The capacitive sensor is simulated by pure capacitance model in this research. In reality, there is also a resistive component as well (which is usually greater than 100 MΩ) for the insulated material [6, 13]. That is, the capacitive sensor is a R||C network in its essence. The resistance in R||C model may be affected by many factors, such as the humidity, material types, material thickness, sweating, etc. It is not easy to evaluate the system characteristics precisely by the R||C model of sensor.

As the adopted input buffers (OPA124) possess ultra-high input impedance (common-mode, 10^{14} Ω||3 pF), which would make the transfer functions nearly identical for both models. Therefore, the pure capacitance model can be assumed to be feasible for capacitive sensors as the suitable input-stage circuit is adopted in the design.

This study utilized rigid double layer PCB for system implementation. The subject should lean tightly against the rigid board and the subject may feel somewhat uncomfortable on the back. In such case, the flexible PCB may be a better choice. In order to prevent the interference from the power lines more effectively, four layer PCB can also be utilized with the inner layer being grounded to shield the conductive fabric tapes from the power-line coupling.

IV. Conclusion

This paper presents a prototype of multichannel capacitive ECG measurement system. Conductive fabric tapes pasted on a double layer PCB are adopted as the capacitive sensors. Two design criteria are proposed for its implementation. This system also incorporates the so-called capacitive driven-body (CDB) circuit to reduce the common-mode interference coupled from power mains. The limb-lead ECG patterns acquired by the presented system have been compared with those by commercial equipments, and the results are competitive. The presented system has also been utilized to monitor the ECG signals of a subject during sleep. On the basis of measurement results, the presented system is also potential for the researches on sleep, though motion artifact remains to be overcome in such system. The tests under various capacitive areas and distances have been performed to evaluate the feasible size. From the measurement MSE, the sensor is suggested to be with an area greater than 60 cm^2 and a distance smaller than 1.5 mm.

The presented system is operated in isolation mode. That is, the system is powered by DC batteries. For non-isolated configuration, the notch filter for PLI removal is suggested to be added in practical implementation. The results of this study may help the researchers to implement a capacitive ECG system and to evaluate its utility in clinical medicine. The system can be implemented by flexible PCB and may be utilized in ambulatory ECG monitoring.

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