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The Non-Contact Heart Rate Measurement System for Monitoring HRV

Ji-Jer Huang, Sheng-I Yu, Hao-Yi Syu, and Aaron Raymond See

***Abstract*—A noncontact ECG monitoring and analysis system was developed using capacitive-coupled device integrated to a home sofa. Electrodes were placed on the backrest of a sofa separated from the body with only the chair covering and the user's clothing. The study also incorporates measurements using different fabric materials, and a pure cotton material was chosen to cover the chair's backrest. The material was chosen to improve the signal to noise ratio. The system is initially implemented on a home sofa and is able to measure noncontact ECG through thin cotton clothing and perform heart rate analysis to calculate the heart rate variability (HRV) parameters. It was also tested under different conditions and results from reading and sleeping exhibited a stable ECG. Subsequently, results from our calculated HRV were found to be identical to those of a commercially available HRV analyzer. However, HRV parameters are easily affected by motion artifacts generated during drinking or eating with the latter producing a more severe disturbance. Lastly, parameters measured are saved on a cloud database, providing users with a long-term monitoring and recording for physiological information.**

Keywords：Noncontact electrocardiogram, Noncontact electrode model, Heart rate variability

I. INTRODUCTION

The refinement of people's diet and increasing work pressure significantly increases the risk for cardiovascular disease. In 2012, the Department of Health, inferred that cardiovascular disease is the 2nd highest cause of death in Taiwan. And according to the World Health Organization, the world population is also aging rapidly, aggravating the existing problems in the community. Therefore, a non-conscious type of physical measurement system will be developed with some intelligence function. It includes a user friendly function for automated capturing, deciding and recording. The long-term and non-conscious recording of HRV parameters will aid people to maintain a healthy lifestyle.

Noncontact physiological measurement techniques can be adapted to acquire bio-electrical signals such as ECG and EMG measurement. Richardson in 1968 made use of the isolated capacitive electrodes technique to measure bio-electrical signals [1]. In recent years, with the rapid development of semiconductor technology, making miniaturized electronic circuits has become easier. As a result, numerous authors are studying the use of noncontact electrode technology and integrating it to daily life. In 2008, Steffen et al. published an article on the use of noncontact electrodes [2]. The article gave a detailed description on the theory behind the isolated electrode model with its equations. It also made a clear description of the model's electrode and circuit structure diagram. The electrodes are placed on the backrest and on the seat of the chair to acquire ECG and physiological signals. It was noticed that this method can easily be affected by movement noise thus resulting in an inaccurate measurement. In 2007, Maruyama et al. researched noncontact electrodes placed on a seat belt [3]. Their research made use of diverse structures for the electrode placement to initiate practical measurements. It was found that even with only two noncontact electrodes ECG can be measured providing drivers with information relevant to their current physiological condition. In 2006, Kato et al. performed a study on sleeping infants to measure the pressure distribution induced by the back on the bed as well as designed noncontact electrode and ground electrode patterns [4]. Then in 2006, Lim et al. made use of noncontact electrodes on the backrest and seat of a chair to measure ECG and also calculated the value of the signal to noise ratio. Consequently, they performed experiments using different materials for the backrest. Results showed that cotton provided the best response as compared to acrylic, sheep wool, and Teflon (PTFE) for frequencies ranging from 0.1Hz to 500Hz [5]. In addition, they made use of noncontact electrodes applied to bathroom equipment such as toilet bowls and bathtubs among others [6] while monitoring the user's physiological information. Then, HRV is a measure of the variation between heart beats and HRV parameters serves as an indicator of a person's physiological and psychological response [7]. It can also document a person's response to a sudden environmental stimulus as well as how the heart adapts to such an abrupt event. Henceforth, the use of HRV analysis can evaluate the effect of the autonomous nervous system (ANS) in controlling the heart's output [8]. In addition, the sympathetic nervous system (SNS) and parasympathetic nervous system (PNS) have been topics of interest for researchers in the past 20 years, in relation to HRV and the heart condition [9].

II. MATERIALS AND METHODS

A. *Noncontact electrode model*

For physiological measurements using a sofa, the electrode and body are separated by a cushion. The relationship of the measured signals from both the measurement end and the feedback electrodes is shown in Fig. 1. In this figure, C_{cloth}

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represents the capacitive value between the cloth and sofa coversheet and the isolated electrode. C_m and C_g represent the stray capacitance between the instrument end and reference. Moreover, the capacitance of C_{cloth} changes with the use of various materials and thickness of the cloth and the sofa cover.

In order to ascertain the voltage produced by the minute induced charge by the isolated electrode it is imperative to utilize high input impedance and low noise operation amplifier and the sensed charges through the resistor (R_B) and operation amplifier is converted into a voltage signal. Outside the electrode and the circuit is a metal shielding layer used to isolate outward noise so as not to affect the electrode and is also used as the reference for the front-end circuit signal. Consequently, it also generates the capacitance (C_{shield}). Two synonymous shielded electrode circuits are shown in Fig. 2, excluding the stray capacitance of C_{cloth}, C_m and C_g. The shielding and signal end also act as a capacitive shielding (C_{shield}). The noise developed in the front-end consists of the voltage noise (V_n) and current noise (I_n). The noise signal is mainly from radio interference and thermal noise radiated by the circuit board. In the opening amplifier the input end is composed of the input impedance (R_{in}) and input capacitance (C_{in}). Moreover, in addition to the cloth and body impedance, the skin impedance (R_{skin}) should also be considered. It is usually in the range of several kΩ to more than 10 MΩ. Hence, the skin humidity has a large effect on the system's front end.

Accordingly, the induced current is very small and is converted to a voltage signal using resistor R_B. In order to find the voltage across R_B we needed to use high input impedance operation amplifier. An opening amplifier utilizing the Field Effect Transistor (FET) was used to produce high-input impedance. As a result the input impedance can exceed 10^{13} Ω and the capacitive impedance can be brought to a minimum value. Afterwards, a function is derived from the circuit's input signal (V_{ECG}) and output signal (V_o). During the calculation process noise was ignored in the computation. Later on the noise is considered in the calculation of the Signal to Noise Ratio (SNR). First, C_{shield} is paralleled with the C_{in} to produce C_p as shown in the equation. Next, C_{cloth}, $C_g/2$, $C_m/2$ are put in the series as shown in (2). The circuits of the two shielded electrodes are then converted into a single shielded electrode circuit before deriving the function shown in (3).

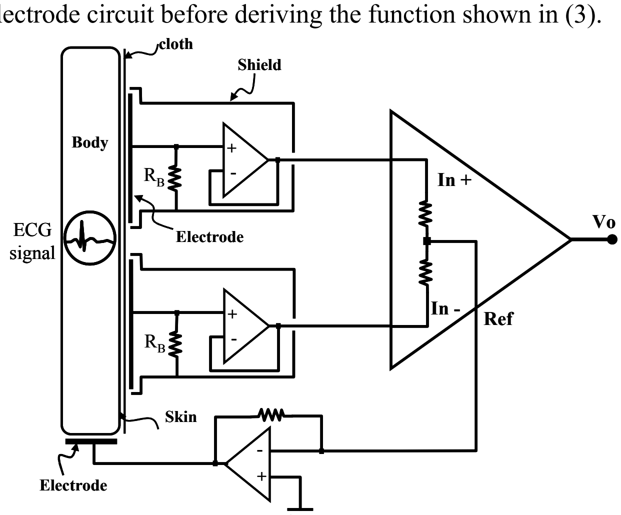


Figure 1. Diagram of shielded electrode and body surface

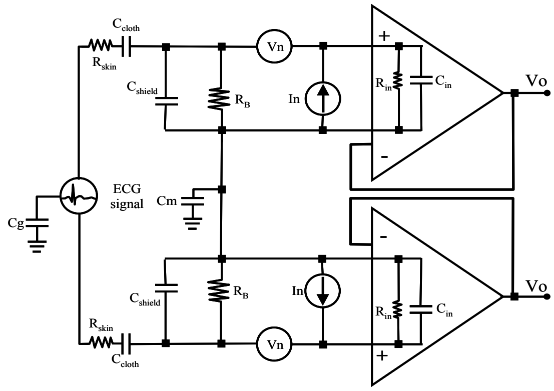


Figure 2. Equivalent circuit diagram of a double isolation electrode

$$C_p = C_{shield} + C_{in} \tag{1}$$

$$C_s = \frac{C_{cloth} C_g C_m}{C_g C_m + 2C_g C_{cloth} + 2C_m C_{cloth}} \tag{2}$$

$$A(jw) = \frac{jwR_B R_{in} C_S}{R_B + R_{in} + jw[R_B R_{in}(C_S + C_P) + R_{skin} C_s (R_{in} + R_B)] - w^2 R_{in} R_B R_{skin} C_p C_S} \tag{3}$$

In function, (3) displays a band-pass filter circuit and if the skin impedance R_{skin} is very small $R_B R_{in}(C_S + C_P) >> R_{skin} C_s (R_{in} + R_B)$ ignoring $w^2 R_{in} R_B R_{skin} C_p C_S$. $A(jw)$ from (3) can then be simplified as (4).

$$A(jw) = \frac{jwR_B R_{in} C_S}{R_B + R_{in} + jw[R_B R_{in}(C_S + C_P)]} \tag{4}$$

Assuming an Op Amp with an internal $R_{in} \to \infty$, (4) can be further simplified to (5) integrating equation 1 and 2 into equation 5. If C_{cloth}, $C_g/2$, $C_m/2$ are put in series and $C_g/2$, $C_m/2$ is larger than C_{cloth} hence forming equation 6.

$$A = \frac{C_S}{C_S + C_P} \tag{5}$$

$$A = \frac{C_{cloth}}{(C_{cloth} + C_{shield} + C_{in})} \tag{6}$$

Based from equation 6 it would be best to reduce C_{in} and C_{shield} or raise C_{cloth} to obtain a satisfactory signal. In the event of a large R_{skin}, $w^2 R_{in} R_B R_{skin} C_p C_S$ in (3) cannot be ignored because it will greatly attenuate the gain of A.

Based from Fig. 2, we can formulate the SNR function. And base from the equivalent circuit diagram the input V_{ECG}, noise voltage (V_n) and current noise (I_n) can be deduced. In addition, since R_B has a large value, it would also generate a large thermal noise that cannot be ignored. The derived equation is shown in (7). Wherein, K is the Boltzman constant, T, temperature and B is the bandwidth of the band pass filter.

$$V_{Thermal}^2 = 4KTR_B B \tag{7}$$

Now, there are three noises under consideration namely $V_{thermal}$, V_n and I_n. Applying the superposition theorem, it can be derived to the SNR of the front-end circuit [6]:

$$SNR = \frac{|V_{ECG}|}{\sqrt{I_n^2 Z_s^2 + V_n^2 \left(1 + \left|\frac{Z_s}{Z_p}\right|^2 + \left|\frac{Z_s}{R_B}\right|^2\right) + V_{Thermal}^2 \left|\frac{Z_s}{R_B}\right|^2}} \tag{8}$$

When the value of $C_m/2$ and $C_g/2$ in series is larger than C_{cloth}, $Z_s = \frac{1}{jwC_s} + R_{skin}$ becomes $Z_s = \frac{1}{jwC_{cloth}} + R_{skin}$ and $Z_p = \frac{1}{jwC_p} = \frac{1}{jw(C_{shield} + C_{in})}$. When comparing the gain and noise, as C_{cloth} and R_B increase a higher gain and SNR can be obtained. Subsequently, an increase in R_{skin} would attenuate both the gain and SNR.

B. Hardware system architecture

The design of the front-end circuit of the noncontact electrode requires very high input impedance. Figure 3 shows the block diagram of the shielded electrode ECG measurement system. First, the noncontact electrode circuit consists of a pre-amplifier, electrode face and plate for shielding. The size of active electrode was 176 cm^2 (16 cm x 11 cm) and thickness was 1.5cm. The reference electrode (30cm x 30cm) was placed under the seat cushion of sofa.

The input impedance of the low noise operation amplifier (OPA121) should be greater than $10^{13}\Omega$ to serve as the front-end buffer to acquire a very minute change in physiological signal. The signal obtained is then sent through an instrument amplifier with a high CMRR (>120dB) to obtain the differential signal between the two noncontact electrodes. The signal will then be approximately equivalent to the standard lateral ECG measurement. Consequently, it goes through a high pass analog filter with cut-off frequency at 0.5Hz before going through a low-pass analog filter with a cut-off frequency of 105Hz. Later it would pass through a notch analog filter to remove the 60Hz electrical noise. Afterwards, it will be input into a MCU (PIC32MX575F) programmable gain and offset adjustment circuit by MCU.

The resulting ECG is then input into a 10 bit Analog to Digital converter (ADC) with a sampling rate of 1 kHz. The output digital signal then goes through an IIR digital filter inside the DSP/MCU to remove the trend and calculate the average energy amplitude. The average energy amplitude will then be used as a feedback to the input to adjust the programmable gain and offset adjustment circuit. The process can then reduce the difference in the signal produced by different materials and clothing and obtain a better SNR. The circuit board and the sofa with integrated noncontact electrode are shown in Fig. 4.

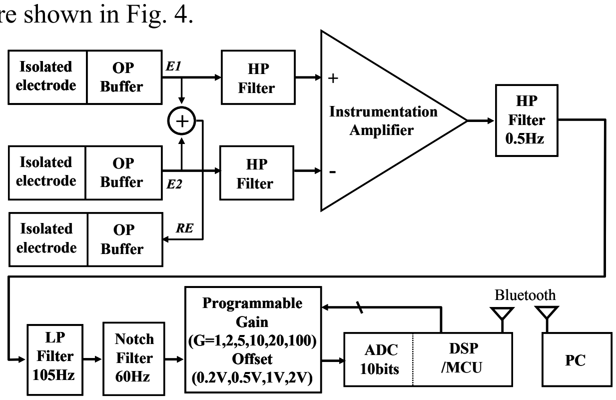
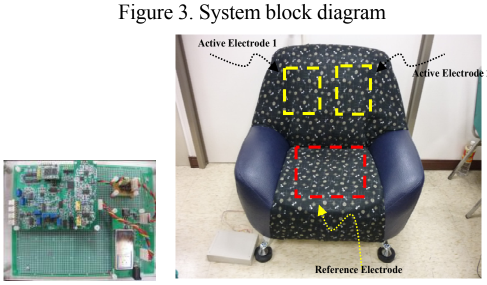


Figure 3. System block diagram



(a) (b)

Figure 4. (a) System circuit board and (b) Noncontact electrode sofa

C. Heart Rate Variability (HRV) parameters

HRV examines the heart signal to obtain the HRV parameters and its index values. In this study, time domain analysis initially calculates the heart rate and the R-R intervals. R waves are detected using slope method. The acquired ECG waveform S(n) is substituted into (9) to determine the QRS waveform. Through locating the largest value nearby the R peak can be known. This in turn allows the RR interval to be calculated. Then by utilizing the time interval and by the application of statistical methods can be used to determine the HRV. HRV parameters are shown in Table I.

$$Slope(n) = -2S(n-2) - S(n-1) + S(n+1) + 2S(n+2) \quad (9)$$

TABLE I. HRV PARAMETERS

R-R interval	$RT_i = T_{R_i} - T_{R_{i-1}}$
MeanRR	$(\sum_{i=1}^{n} RT_i)/n$
SDNN	$\sqrt{[\sum_{i=1}^{n}(RT_i - MeanRR)^2]/n-1}$
RMSSD	$\sqrt{[\sum_{i=1}^{n-1}(RT_i - RT_{i-1})^2]/n-2}$
SDSD	$(SDNN / MeanRR) \cdot 100\%$

D. Software system architecture

The software system architecture is composed of two parts namely the DSP microcontroller unit DSP/MCU and the HRV analysis algorithm. First, the non-ECG signal is received. Then it goes through to a second order IIR Butterworth digital notch filter with a 60Hz cutoff frequency. From there it undergoes further filtering using a 5^{th} order low pass and 8^{th} order high pass IIR digital filter with low cutoff frequency at 100Hz and 0.5Hz high cutoff frequency respectively. These data are then transmitted via Bluetooth to the HRV analyzer tools implemented by using LabVIEW software. Upon receiving the data from the Bluetooth interface, the R wave is detected and enhanced. Subsequently, the time domain analysis to determine the HRV parameters is calculated. The software will calculate the SDNN, RMSSD, and SDSD before displaying and recording the data.

III. EXPERIMENTAL SETUP

A. Testing of various cloth materials

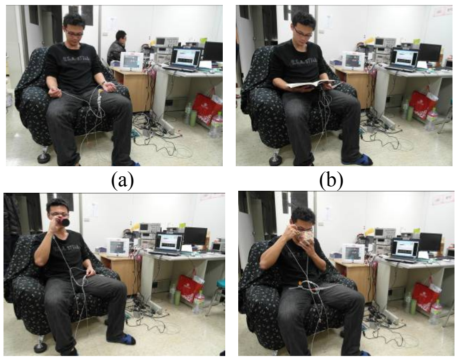
The capacitive coupling electrode is affected by different fabric materials. Hence, the 3532-50 LCR by Hioki Company was utilized to determine the capacitance of different materials. The scanning frequency is set at 42Hz to 300Hz. The electrode surface is 176cm^2 (16cmx11cm). Different fabrics of different materials and thicknesses were inserted between the upper and lower electrodes.

B. Different measurement conditions

The system was tested by performing daily living tasks such as sleeping, reading, eating and drinking. The standard ECG was measured using SpaceLabs' Ultraview SL2200 and our self-developed system that can measure ECG from a sofa with the person's clothes on. Fig. 5 shows an illustration of the actual measurement.

IV. RESULTS AND DISCUSSION

The capacitance values measured using different types of fabrics of specified area and thickness showed in Fig. 6. From this figure one may notice that the cotton material provides the highest capacitive value. It is also less likely to affect the measurement of surface charges going through the capacitive coupling electrode. Fig. 7 displays the measured ECG from four different scenarios. It can be perceived that when the person is sleeping or reading, the current system and the commercial device is able to clearly measure the heart rate. In the event of tea drinking, since there is the arm movement and a noticeable amount of motion artifact distorts the signal. Lastly, when eating the motion artifact is significantly greater.



(a) (b) (c) (d)

Figure 5. (a) Sleeping (b) Reading (c) Drinking (d) Eating

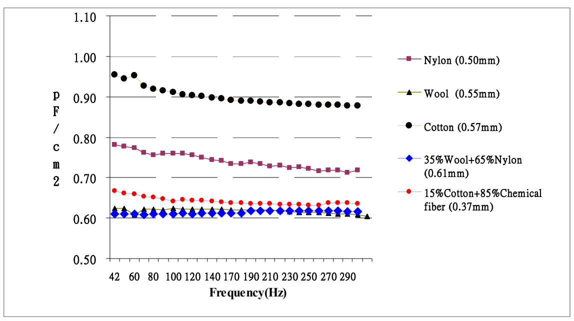
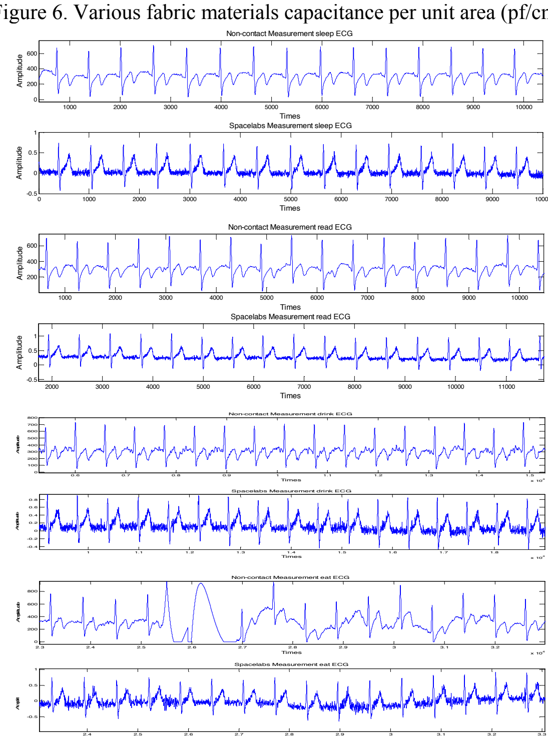


Figure 6. Various fabric materials capacitance per unit area (pf/cm^2)



(a) (b) (c) (d)

Figure 7. ECG during different scenarios (a) Sleeping (b) Reading (c) Eating (d) Drinking (Upper diagram is own system and lower diagram is from standard commercial ECG device)

Then, the recorded one minute data is fed to the system wherein HRV parameters are calculated as shown in Table 2.

TABLE II. HRV PARAMETERS FROM VARIOUS 1-MINUTE CONDITIONS

Unit: millisecond		MeanRR	SDNN	RMSSD	SDSD
Sleeping	Our system	625	27	72	4.32
	Standard ECG	635	70	82	11.02
Reading	Our system	607	69	5	1.14
	Standard ECG	607	85	10	1.40
Drinking	Our system	605	41	68	6.78
	Standard ECG	605	13	7.5	2.16
Eating	Our system	624	120	180	19.23
	Standard ECG	611	58	96	9.49

V. CONCLUSIONS

The experiments done on the noncontact heart rate measurement system, different fabric materials significantly affect the electrode sensing. We found that cotton fabric has the highest capacitive value and signal gain. Next, the current system provides an initial implementation that integrates a noncontact ECG to a home sofa while the subject is wearing a thin cloth and analyzes the HRV parameters. Moreover, measurement under different living conditions was considered stable during reading and sleeping. The system was able to produce HRV parameters consistent with that of the commercial device. However, since motion artifact is quite serious for both drinking and eating it would affect the computed HRV parameters. In the future, improvement in the digital filters will be used to improve the R-wave detection. Subsequently, future advancement on the current research would deal with the noncontact electrodes and dielectric material to further the research and attain more stable results.

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