Wireless potential difference electrocardiogram constituted by two electrode-pairs wearing comfort

Hsin-Yen Hsieh,\textsuperscript{h,2} Ching-Hsing Luo\textsuperscript{a,1,2} and Cheng-Chi Tai\textsuperscript{b}

\textsuperscript{a}School of Data and Computer Science, Sun Yat-Sen University (East Campus),
Guangzhou 510006, China
\textsuperscript{b}Department of Electrical Engineering, National Cheng Kung University,
Tainan 70101, Taiwan

E-mail: luojinx5@mail.sysu.edu.cn

ABSTRACT: Recently proposed is a newly Electrocardiogram (ECG) with separate powered electrode-pairs without common ground between two pairs instead of the traditional ECG with a reference electrode and the necessity of common ground creating wearing uncomfortable. Even the novel two-electrode-pair ECG can highly improve the wearing comfort, it must become wireless by overcoming wireless difference problem. This study introduces two analog transceivers of two amplitude modulation (AM) frequencies combined with two front-end amplifiers, two electrode-pairs, and one difference amplifier to implement two-electrode-pair wireless ECG. The experimental results demonstrate that the wireless difference is made possible, i.e., ECG can be constructed after wireless transmission. Two-electrode-pair wireless ECG can greatly improve wearing or measurement comfort without many wires around the body, even Lead I is clear with the acceptable signal-to-noise ratio and only clear QRS complex waves of Lead II and III are detectable for wearing comfort in the health applications.

KEYWORDS: Analogue electronic circuits; Data acquisition circuits; Front-end electronics for detector readout

\textsuperscript{c}Corresponding author.
\textsuperscript{d}First authors.
1 Introduction

As the progress of humans to the 21st century, the more people care about their health and the more attention pays to the ubiquitous wireless biopotential acquisition system. Non-invasive and portable wireless ECG is one of the best ways to monitor heart activity. In order to achieve the purpose of carrying a wireless electrocardiogram, it has become extremely important research scope. The convenience and the comfort of a wireless ECG acquisition system are useful in many situations even at home. When designing a portable ECG system and also having a wireless transmission function, it is necessary to have a driven right leg circuit (DRL) or a reference electrode in order to reduce common-mode noise and have a reference potential for both electrodes [1]. Thus, the body is covered with wires and causes inconvenience to the user.

A clinical ECG system pastes electrodes in precise positions on the hands, legs or chests. It requires nursing personnel with specialized medical background to accurately position the electrode in the correct position. Therefore, many researchers have actively studied the convenience of a wireless electrocardiogram, trying to make wireless ECG more convenient and comfortable. A chest lead system using three electrodes usually proposed, the communication method can be Zigbee [2] or Bluetooth [3]. In order to allow portable ECG to be used on curved skin, Winokur et al. [4] proposed L shaped flexible printed circuit board (PCB), which made the portable ECG free to use on the chest but required 5 electrodes.

Non-contact capacitive electrodes technology using adaptive network topology protocol [5] and Bluetooth [6] communication could avoid skin irritation caused by contact with wet electrodes, but DRL was also required. Bluetooth was used to wirelessly transfer ECG data to mobile phones based on the technique of non-contact capacitive electrodes. A Bluetooth wireless ECG adopted concentric ring electrodes with DRL (driven right leg circuit) for common references [7]. The medical cloud service was carried out by uploading the data to the health monitoring system of the web server [8]. The capacitive electrodes were adopted in a single-arm [9] to greatly improve wearable convenience, but DRL was required for both medical cloud services and single-arm applications. A capacitor
electrode designed as a two-electrode ECG was innovative in the development of the two electrodes ECG system [10]. However, two electrodes placed on the left and right wrists must be connected to the host through wires for common ground. A dry foam electrode wearable ECG system was proposed to send an alert to the medical server via Short Message Service (SMS) [11], in which it still required DRL (driven right leg circuit). A DRL was also required for wireless ECG with a dry flexible electrode [12]. Architecture using AC coupling to design a wireless ECG requires four electrodes [13, 14], in which three of the electrodes were used to detect the ECG signals of lead I, II, and III with the fourth electrode for DRL. Most commercial two-electrode heart rate sensors on the market adopt AC coupling architecture but common ground is still used [15, 16]. A two-electrode wireless ECG was developed by using ZigBee transmission with common ground needed [17].

In the process of transmitting signals of the wireless ECG, the most important is how to effectively reduce the transmission power [18]. The Holter monitor [19] recording heart activity for consecutive 24 hours involves the use of DRL (driven right leg circuit) electrode and wiring over the patient’s body limiting the patient’s regular activity. The novel wearable clothing provided excellent convenience without any wire connection [20–23]. It is convenient to wear on special personnel such as soldiers and firefighters, but the general people may not prefer and afford the wearable clothing, because the wearable clothing requires special materials and techniques to design the wires, electrodes, and modules inside the articles of clothing. A wireless ECG was fabricated by three-stage amplitude modulation but with DRL required [24].

For wearing comfort, our previous work [25] proposed a two-electrode-pair ECG without common ground between two measurement electrode-pairs to save wiring over around the body. Each electrode pair was supplied with isolated power but its own ground must be attached on the skin as a ground electrode near the measurement electrode to form an electrode pair. Both ground electrodes without any wire connected together were attached on the skin to share the body fluid (named as body electrode reservoir) via skin and tissue impedance as common ground. Both measurement signals were read out and differentially amplified to form ECG with power and ground separated from two electrode-pairs. This design did create the potential of wearing comfort without any wire all over the body with each electrode-pair attached on the skin by using its own power source and ground. Nonetheless, it still needs wires to connect both amplified biopotential to get potential difference for ECG formation. This study further cuts the connection wires by implementing wireless modules to transmit the amplified potentials and form ECG by the difference between the wirelessly received potential entitled as wireless potential difference. The experimental results show that wireless potential difference does work and two-electrode-pair wireless ECG is made possible for wearing comfort in the health applications.

2 Methods and materials

2.1 Systematic description of wireless potential differential ECG

The general wireless ECG is made by either the bipolar limb leads (figure 1(a)) or the chest lead (figure 1(b)) with DRL (driven right leg circuit) or reference electrodes [26]. And, importantly, common ground is needed and an ECG is formed by potential difference before wireless transmission.

This study proposes wireless potential differential ECG with two electrode-pairs, two front-end amplifiers, two wireless transceivers of two amplitude modulation (AM) frequencies (500 MHz and
Figure 1. General Wireless ECG in which $\downarrow$ is common ground.

1 MHz), and one instrumentation amplifier for potential difference to form Lead I ECG shown in figure 2. Each electrode-pair with one measurement and one ground electrodes shares the separated power supply and ground (GND 1 and GND 2) with one front-end amplifier and one transmitter module. Two receivers get two measurement potentials wirelessly differenced by one instrumentation amplifier (IA) to form Lead I ECG with the third power supply and ground (GND 3). In total, three sets of power supply and ground are adopted without any wire to create common ground for the testification of wireless potential differential ECG. In order to avoid co-channel interference, it is necessary to use the AM transmitters of different frequencies for wireless transmission.

Figure 2. Two-electrode-pair wireless ECG (Lead I). The components of Boards 1-a, 1-b, 2-a, 2-b and 3 shown in figure 6 are marked in the dotted rectangles.

Figure 3 shows a circuit diagram of wireless potential differential ECG to understand how GND 1 and GND 2 are connected to the body electrode reservoir (GND B) as reference ground via skin and tissues [25, 27, 28]. $E_{eq}$ is half-cell potential. $R_d$ and $C_d$ represent the impedance of the electrode-electrolyte interface and the polarization effect. $R_g$ is the equivalent resistance of the conductive gel between the electrode and the skin. $E_{eq}$ is the potential difference produced by the difference in ion concentration between the stratum corneum and the conductive gel. $R_{ep}$ and $C_{ep}$ are the resistance and capacitor of the epidermal layer. $R_{ds}$ is resistance in the dermis and subcutaneous layers [29]. An adult $R_{tissue}$ impedance from the sternum to the calf was approximately 200 $\Omega$ [30], and the impedance of the electrode and the skin equivalent circuit is about 1 M$\Omega$ [29, 31].
Based on the circuit diagram in figure 3, the electrocardiac potential $V_{ECG}$ can be modeled as one potential $\frac{V_{ECG}}{2}$ at one measurement electrode subtracting the other potential $-\frac{V_{ECG}}{2}$ at the other measurement electrode. The output voltage at the front-end amplifier should be $A_1 \left( -\frac{V_{ECG}}{2} - \varepsilon_1 \right)$ or $A_2 \left( \frac{V_{ECG}}{2} - \varepsilon_2 \right)$. The input voltage of the instrumentation amplifier can be obtained as $A_1 A_{1M} \left( -\frac{V_{ECG}}{2} - \varepsilon_1 \right)$ or $A_2 A_{500K} \left( \frac{V_{ECG}}{2} - \varepsilon_2 \right)$ in addition to $\varepsilon_{1M}$ or $\varepsilon_{500K}$ noise introduced from the air interference. As a result, the output voltage $v_O$ of the instrumentation amplifier can be expressed in eq. (2.1).

$$v^{(IA)}_O = A_{IA} \left[ \frac{V_{ECG}}{2} (K_1 + K_2) - (\varepsilon_2(t) K_2 - \varepsilon_1(t) K_1) + (\varepsilon_{500K} - \varepsilon_{1M}) \right]$$  \hspace{1cm} (2.1)

If $K_1 = K_2 = K$, eq. (2.1) can be simplified as follows:

$$v^{(I)}_O = A_{IA} \left[ K (V_{ECG} - V_{t2} + V_{t1}) + \varepsilon_{500K} - \varepsilon_{1M} \right]$$  \hspace{1cm} (2.2)

where

$A_{IA}$: the gain of instrumentation amplifier
$K_1 = A_1 \times A_{1M}$ and $K_2 = A_2 \times A_{500K}$
$A_1$: the gain of the front-end amplifier 1
$A_2$: the gain of the front-end amplifier 2
$A_{1M}$: the gain of 1 MHz AM transmitter and receiver
$A_{500K}$: the gain of 500 KHz AM transmitter and receiver
$\varepsilon_1(t)$: the noise from GND 1 to GND B
$\varepsilon_2(t)$: the noise from GND 2 to GND B
$\varepsilon_{1M}$: the noise induced by 1 MHz wireless transmission
$\varepsilon_{500K}$: the noise induced by 500 KHz wireless transmission
$V_{t1}$: the potential between GND 1 and GND B
$V_{t2}$: the potential between GND 2 and GND B

Figure 3. Circuit diagram of wireless potential differential ECG. The components of Boards 1-a, 1-b, 2-a, 2-b and 3 shown in figure 6 are marked in the dotted rectangles.
2.2 Body surface potential mapping (BSPM) analysis

As a heart beats, it incites the depolarization and repolarization of the cardiomyocytes, which creates a potential change on the surface of the skin called as body surface potential [32]. Body surface potential mapping (BSPM) shown in figure 4 is the potential distribution on the surface of the chest skin at the given moment of a cardiac cycle [32], in which two electrode pairs ((E₁, GND 1) and (E₂, GND 2)) are located.

![Figure 4. BSPM in response to the heart dipole.](image)

2.3 Device implementation

The proposed system is divided into two parts: a measurement and transmitter circuit as well as a receiver and ECG formation circuit shown in figure 5. In the measurement and transmitter circuit (figure 5a), the body surface potential E₁ or E₂ is amplified by a front-end amplifier TL082 and then transmitted to the air via 1 MHz or 500 KHz AM modulator MC1496 with an AM ring antenna. TL082 comprises a unity-gain buffer and a noninverting amplifier to ensure high input impedance.

![Figure 5. Block Diagram of Wireless ECG: (a) Transmitter, (b) Receiver.](image)
required for the measurement [6]. In the receiver and ECG formation circuit (figure 5b), the received signal ($\tilde{E}_1$ or $\tilde{E}_2$) is demodulated by an envelope detector and a low pass filter (LPF) is used for noise reduction before entering the instrumentation amplifier (IA, AD620). ECG signal is created by the differential amplification in the IA and then filtered by a bandpass filter (0.5–100 Hz).

Figure 6 presents the implementation of wireless potential differential ECG device with two electrode-pairs: (a) measurement and transmitter and (b) receiver and ECG formation. In figure 6a, the transmitter circuit (Board 1-b or 2-b) and measurement circuit with an electrode pair (Board 1-a or 2-a) are placed on the top and bottom of the measurement and transmitter box separately. The measurement and transmitter box was made by a 3D printer. In figure 6b, the receiver circuit is implemented in the left half of Board 3 and followed with the ECG formation circuit in the right half of Board 3.

![Figure 6](image.png)

**Figure 6.** Wireless potential differential ECG device with two electrode-pairs. (a) Measurement (Board 1-a, 2-a) and Transmitter (Board 1-b, 2-b), ruler in cm unit. (b) Board 3: receiver (upper for 1 MHz and lower for 500 KHz) and ECG formation (Instrumentation amplifier and signal processing circuits), ruler in cm unit.

3 Results

Figure 7 shows the measurement environment for Leads I, II, III. ECG signals are shown on the screen (Multifunction I/O Device, National Instruments, Austin, TX) and transmitted to the laptop for data storage via LabView software. RA and LA electrode pairs are placed between 5th and 6th ribs to avoid muscle movement interference. The distance between RA and LA electrode pairs is 27 cm. A 60 Hz filter made with a 2nd-order Inverse Chebyshev Notch Filter in the LabView is used to eliminate line interference. The sampling frequency is set at 1 kHz. Power consumption of
1 MHz transmitter or 500 kHz transmitter is 1 W or 0.93 W separately. Power consumption of the receiver is 187.4 mW.

Figure 7. Measurement environment (a) Lead I, (b) Lead II, (c) Lead III. Both antennas (receiver and transmitter) are close (10 cm far) enough for sufficient power transmission.

Figure 8 presents the ECG waveform of Lead I, in which the one with large noise (solid line) is made by the proposed method while the one with almost no noise (dotted line) is served as the control signal measured by a commercial device (MP36, BIOPAC inc.) with regular DRL (driven right leg circuit) and common ground. It can be seen that the noise of the proposed system is even large but its Q, R, S waves can be seen clearly and have the similar wave patterns as those in the control signal. Signal-to-noise ratio (SNR) of Lead I is 24.86 dB.

Figure 8. Lead I: the proposed wireless potential differential and control ECG signals.

In figures 9 and 10, the noise is too big to identifying the ECG waveforms of Leads II and III by the proposed method except the clear QRS waves. This is because LL lead is far away from the heart, so the body surface potential is quite weak against the noise. SNR of Leads II and III are 22.13 dB and 21.18 dB, respectively.
Figure 9. Lead II: the proposed wireless potential differential and control ECG signals.

Figure 10. Lead III: the proposed wireless potential differential and control ECG signals.

4 Discussion & conclusion

With electrode-pair sharing the common body electrode reservoir, wireless potential difference is made possible to form ECG after wireless transmission. Similar to our previous work [20], the proposed wireless potential differential ECG obtains a clear Lead I with acceptable signal-to-noise ratio (SNR) and clear QRS waves of Leads II and III. The surface potential on the chest is significant in comparison to noise, yielding a clear Lead I ECG, while the surface potential on the ankle is quite weak to give a clear Lead II or III ECG. This limits the applications of the proposed wireless potential differential ECG even it creates wearing comfort without any wire connection around the body. For homecare or exercise control, heart rate is a major monitoring parameter by wearing the chest belt or grabbing two handles for wired connection while riding a bike. For a wrist pulse
signal measurement by a wearable watch, it is difficult to get the accurate heart rate and heart rate variability even it is quite comfortable to be worn.

The proposed wireless potential differential ECG can provide an accurate Lead I with acceptable SNR without any wire around the body for easy wearing comfort. It is expected that the next research should focus on the increase of SNR and arrhythmia detection with many subjects for practical applications, such as homecare or exercise health control. For instance, a digital wireless transmission instead of analog wireless transmission can largely decrease the noise from the air but it is introduced the unwanted delay that may interfere the wireless difference function.

References

[1] B.B. Winter and J.G. Webster, Driven-right-leg circuit design, IEEE Trans. Biomed. Eng. 30 (1983) 62.
[2] H. Cao, H. Li, L. Stocco and V.C. Leung, Wireless three-pad ECG system: Challenges, design, and evaluations, J. Commun. Netw. 13 (2011) 113.
[3] G. Gargiulo et al., An ultra-high input impedance ECG amplifier for long-term monitoring of athletes, Med. Devices (Auckl.) 3 (2010) 1.
[4] E.S. Winokur, M.K. Delano and C.G. Sodini, A wearable cardiac monitor for long-term data acquisition and analysis, IEEE Trans. Biomed. Eng. 60 (2013) 189.
[5] E. Nemati, M.J. Deen and T. Mondal, A wireless wearable ECG sensor for long-term applications, IEEE Commun. Mag. 50 (2012) 36.
[6] J. Wannenburg, R. Malekian and G.P. Hancke, Wireless capacitive-based ECG sensing for feature extraction and mobile health monitoring, IEEE Sensors J. 18 (2018) 6023.
[7] Y. Ye-Lin, J.M. Bueno-Barrachina, G. Prats-Boluda, R.R. de Sanabria and J. Garcia-Casado, Wireless sensor node for non-invasive high precision electrocardiographic signal acquisition based on a multi-ring electrode, Measurement 97 (2017) 195.
[8] E.-M. Fong and W.-Y. Chung, Mobile cloud-computing-based healthcare service by noncontact ECG monitoring, Sensors 13 (2013) 16451.
[9] V.P. Rachim and W.-Y. Chung, Wearable noncontact armband for mobile ECG monitoring system, IEEE Trans. Biomed. Circuits Syst. 10 (2016) 1112.
[10] S. Majumder, L. Chen, O. Marinov, C.-H. Chen, T. Mondal and M.J. Deen, Noncontact Wearable Wireless ECG Systems for Long-Term Monitoring, IEEE Rev. Biomed. Eng. 11 (2018) 306.
[11] K.C. Tseng, B.-S. Lin, L.-D. Liao, Y.-T. Wang and Y.-L. Wang, Development of a wearable mobile electrocardiogram monitoring system by using novel dry foam electrodes, IEEE Syst. J. 8 (2014) 900.
[12] C. Lou et al., Flexible graphene electrodes for prolonged dynamic ECG monitoring, Sensors 16 (2016) 1833.
[13] Y. Wang, R. Wunderlich and S. Heinen, Design and evaluation of a novel wireless reconstructed 3-lead ECG monitoring system, in proceeding of the IEEE Biomed. Circuits Syst. Conf. (BioCAS), Rotterdam The Netherlands (2013), pg. 362.
[14] Y. Wang, R. Wunderlich and S. Heinen, A low noise wearable wireless ECG system with body motion cancellation for long term homecare, in proceeding of the IEEE 15th International Conference on e-Health Networking, Applications and Services (Healthcom 2013), Lisbon Portugal (2013), pg. 507.
[15] E. Spanò, S. Di Pascoli and G. Iannaccone, Low-power wearable ECG monitoring system for multiple-patient remote monitoring, IEEE Sensors J. 16 (2016) 5452.
[16] E.M. Spinelli, R. Pallás-Areny and M.A. Mayosky, AC-coupled front-end for biopotential measurements, *IEEE Trans. Biomed. Eng.* **50** (2003) 391.

[17] C.D. Pereira and P.M. Mendes, Development of a two-electrode ECG acquisition system with dynamic interference rejection, in proceeding of the 1st Portuguese Biomedical Engineering Meeting, Lisbon Portugal (2011), pg. 1.

[18] A. Sodhro, A. Sangaiah, G. Sodhro, S. Lohano and S. Pirbhulal, An energy-efficient algorithm for wearable electrocardiogram signal processing in ubiquitous healthcare applications, *Sensors* **18** (2018) 923.

[19] T. Inoue, T. Tsujioka, S. Nakajima, S. Hara, H. Nakamura and K. Takeuchi, Evaluation of a wireless vital sensor for ubiquitous health monitoring systems, in proceeding of the IEEE International Workshop on Robotic and Sensors Environments, Phoenix U.S.A. (2010), pg. 1.

[20] P. Shyamkumar, P. Rai, S. Oh, M. Ramasamy, R. Harbaugh and V. Varadan, Wearable wireless cardiovascular monitoring using textile-based nanosensor and nanomaterial systems, *Electronics* **3** (2014) 504.

[21] K. Fujii, Wearable sensing devices for unobtrusive biomedical monitoring, in proceeding of the 2015 IEEE CPMT Symposium Japan (ICSJ), Kyoto Japan (2015), pg. 204.

[22] F. Castano, A. Hernández, C. Sarmiento, A. Camacho, C. Vega and J.D. Lemos, Redundant measurement of vital signs in a wearable monitor to overcome movement artifacts in home health care environment, in proceeding of the 2016 IEEE 7th Latin American Symposium on Circuits & Systems (LASCAS), Florianopolis Brazil (2016), pg. 299.

[23] M. Rapin et al., Wearable sensors for frequency-multiplexed EIT and multilead ECG data acquisition, *IEEE Trans. Biomed. Eng.* **66** (2018) 810.

[24] Y. Yang, X. Zhu, K. Ma, R.B. Simorangkir, N.C. Karmakar and K.P. Esselle, Development of wireless transducer for real-time remote patient monitoring, *IEEE Sensors J.* **16** (2016) 4669.

[25] H.-Y. Hsieh, C.-H. Luo, J.-W. Ye and C.-C. Tai, Two-electrode-pair electrocardiogram with no common ground between two pairs, *Rev. Sci. Instrum.* **90** (2019) 114703.

[26] A. Gabrielli and I. Lax, Measurements on wireless transmission of ECG signals, 2016 JINST **11** C12055.

[27] C. Assambo and M.J. Burke, Amplifier input impedance in dry electrode ECG recording, in proceeding of the IEEE Engineering in Medicine and Biology Society, 2009 EMBC 2009 Annual International Conference, Minneapolis U.S.A. (2009), pg. 1774.

[28] C. Assambo, A. Baba, R. Dozio and M. Burke, Determination of the parameters of the skin-electrode impedance model for ECG measurement, in proceeding of the Proceedings of the 6th WSEAS international conference on electronics, hardware, wireless and optical communications, Corfu Island Greece (2007), pg. 90.

[29] J.G. Webster, *Medical instrumentation application and design*, fourth edition, John Wiley & Sons, New York U.S.A. (2009).

[30] S. Grimnes, Impedance measurement of individual skin surface electrodes, *Med. Biol. Eng. Comput.* **21** (1983) 750.

[31] Y.M. Chi, T.-P. Jung and G. Cauwenberghs, Dry-contact and noncontact biopotential electrodes: Methodological review, *IEEE Rev. Biomed. Eng.* **3** (2010) 106.

[32] L. Nahum, A. Mauro, H.M. Chernoff and R.S. Sikand, Instantaneous equipotential distribution on surface of the human body for various instants in the cardiac cycle, *J. Appl. Physiol.* **3** (1951) 454.