A WBAN Based Cableless ECG Acquisition System

Rui Pan[1], *Student Member, IEEE,* Dingjuan Chua[1], *Student Member, IEEE,*
Jaya Shankar Pathmasuntharam[2], and Yong Ping Xu[1], *Senior Member, IEEE*

***Abstract*— A Wireless Body Area Network (WBAN) based 3-lead cableless electrocardiography (ECG) acquisition system is described. To enable truly cableless ECG monitoring, a new ECG measurement configuration and method that acquires ECG signals at individual lead locations referenced to a localized ground is proposed. The synthesized ECG signals are evaluated against the standard wired 3-lead configuration on the same test subject. Average Pearson correlation coefficients of 0.82, 0.95 and 0.86 have been achieved for Lead I, II and III signals respectively, demonstrating a high degree of similarity in the synthesized signals. Measurements are obtained via a custom wireless network platform utilizing a TDMA-based MAC protocol supporting the star topology and a proprietary front-end ECG acquisition system.**

I. Introduction

In recent years, considerable efforts and developments have been made towards wearable devices that are suitable for long-term daily ECG monitoring. Most devices make use of a number of wired electrodes connected to a multichannel interface and signal acquisition circuits with one wireless node to transmit the data [1], [2]. Although the term "wireless ECG" is used, cables are still needed to connect the electrodes to the interfacing circuit, which is inconvenient for wearable devices, especially when more electrodes are needed. Wearable ECG monitoring devices that are of the patch type, where usually two or three electrodes are placed closely beneath the patch, do not need such cables [3]. Such devices typically provide only one lead signal, which is sufficient for dysrhythmia monitoring. However, as each lead provides information specific only to the region of the heart that is viewed, this may be inadequate for identifying symptoms such as myocardial ischemia which may not show up on all leads [4]. Attempts have been made to eliminate the cables that connect the electrodes and make the wearable ECG system truly wireless [5], [6]. The true wireless ECG system offers great flexibility and scalability, and is potentially of lower costs and improved user comfort.

The challenges inherent to constructing a fully wireless ECG system are manifold. First, with no common body reference ground, signal acquisition may be prone to DC offset drifts and sensitive to noise saturation. Second, the sampling processes at the various detached electrode locations need to be synchronized to an adequate accuracy. Last, post-processing methods need to be developed to obtain the useful lead signals that can be easily interpreted by medical personnel. All these challenges need to be addressed while achieving low power consumption, high efficiency, long-term reliability and a small form factor.

Previously demonstrated systems [5], [6] have shown promising performance in creating true wireless and compact systems. However, the proposed system in [5] is sensitive to electrode placement, which may prevent good reproducibility. In addition, the electrodes are centered around the precordium which may hinder application in female users. The authors in [6] have proposed a transformation method that is able to synthesize 12-lead signals from the proposed lead system. However, some of the synthesized signals show large differences in *S-T* segments while some do not preserve the *QRS* complex.

This paper presents a true wireless ECG measurement configuration that has no wired electrodes and common ground connections. In the proposed electrode placement configuration, familiar Mason-Likar [7] electrode locations are used and all other electrodes are referenced to these locations. The ECG signals obtained from these electrodes can be converted to the standard 3-lead ECG with high correlation coefficients. To isolate the individual nodes fully, the prototype ECG acquisition system is implemented on a battery-powered platform. The platform consists of a custom designed time division multiplex access (TDMA) based media access control (MAC) protocol which utilizes the star topology.

The rest of this paper is organized as follows. Section II provides background information on conventional ECG measurement configurations. Section III presents the proposed Wireless Body Area Network (WBAN) based ECG acquisition system which includes the front-end acquisition system, the MAC protocol and the wireless radio frequency (RF) module. Section IV describes the experimental setup and presents the experimental results. The conclusion is given in Section V.

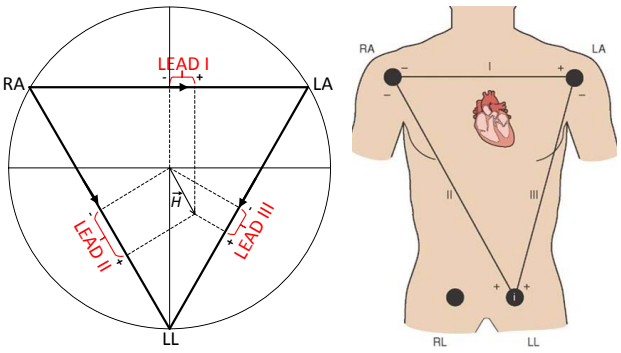
II. Background of ECG measurements

The standard 12-lead diagnostic ECG is commonly used in hospitals to specifically locate the cause of a heart arrhythmia. However, in mobile systems such as the monitoring of patients during transport to the hospital and for daily long-term usage, a simplified 3-lead system is more often used as it is more convenient and is still able to provide a lateral and two inferior views of the heart and thus can indicate the presence and onset of arrhythmias. The source of electrical activity in the heart is the depolarization of the

[1]Rui Pan, Dingjuan Chua and Yong Ping Xu are with the Department of Electrical and Computer Engineering, National University of Singapore, Singapore, e-mail: (g0900211,elechuad,yongpingxu@nus.edu.sg).

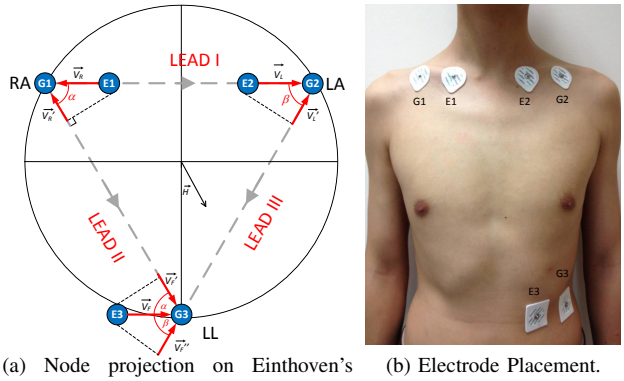
[2]Jaya Shankar Pathmasuntharam is with the Institute for Infocomm Research, Agency for Science, Technology & Research (A*STAR), Singapore, email : (jshankar@i2r.a-star.edu.sg).

978-1-4244-7929-0/14/$26.00 ©2014 IEEE



(a) Einthoven's Triangle [4]. (b) 3-lead ECG [8].

Fig. 1. Conventional 3-lead ECG measurement.



(a) Node projection on Einthoven's Triangle. (b) Electrode Placement.

Fig. 2. Proposed 3-lead ECG measurement.

heart muscle caused during each heartbeat. On average, the depolarizing wave moves along a vector at any given time instance. This is denoted as the heart vector $\vec{H}$ in Fig. 1a [4]. By assuming that the heart can be represented by a dipole located at the center of a homogeneous sphere representing the torso, it is also at the center of the equilateral triangle formed by the limb leads. Thus, the differential voltages measured across the limbs, termed Leads *I*, *II*, and *III* signals are proportional to the projections of the heart vector $\vec{H}$ on the sides of the Einthoven triangle, as shown in Fig. 1a.

These 3 lead signals are effectively differences of the signals at the electrode positions *RA*, *LA* and *LL* as shown in Fig. 1b, where the *RL* electrode serves as a common body ground. This is shown in (1). To obtain these differential signals from the subject, the connecting cables have to span across the entire torso, often rendering the subject immobile as DC drifts are prone when these cables are disturbed.

$$\text{Lead} \quad I = V_{LA} - V_{RA} \tag{1a}$$
$$\text{Lead} \quad II = V_{LL} - V_{RA} \tag{1b}$$
$$\text{Lead} \quad III = V_{LL} - V_{LA} \tag{1c}$$

III. Proposed Measurement Configuration

To counter the problem of immobility, we obtain the ECG signal at individual lead locations referenced to a localized ground. For ease of measurement, the Mason-Likar modification has been adopted and Fig. 2b shows the proposed electrode placement with their individual referenced grounds. Electrodes *E1* and *E2* are placed at the left and right shoulders below the clavicle and near the border of the deltoid muscle while electrode *E3* is placed above the left hip. The individual ground electrodes are placed approximately at a distance of 5cm away in a straight line to obtain a good compromise between signal-to-noise ratio (*SNR*) of the signal and the spacing between the electrodes [9]. Signals are obtained across each pair of electrodes and we define the differential signals as node signals V_R, V_L and V_F as follows.

$$V_R = V_{E1} - V_{G1} \tag{2a}$$
$$V_L = V_{E2} - V_{G2} \tag{2b}$$
$$V_F = V_{E3} - V_{G3} \tag{2c}$$

As illustrated in Fig 2b, the obtained node V_R, V_L and V_F signals can be represented by vectors $\vec{v_R}$, $\vec{v_L}$ and $\vec{v_F}$. Due to the unique locations of the electrodes, $\vec{v_R}$ and $\vec{v_L}$ reside on the Einthoven Lead *I* vector while vector $\vec{v_F}$ is parallel to it. Due to variances in actual distances between the electrodes, the gains of the front-end acquisition system and human variability, a scaling factor C_N is introduced for each node signal. Subsequently, simply by projecting the signals on the Einthoven triangle, the original Einthoven vectors can be derived accordingly as a linear combination of two node signals as follows.

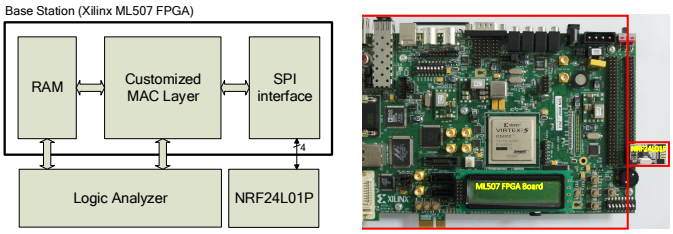
$$\text{Lead} \quad I = C_L V_L - C_R V_R \tag{3a}$$
$$\text{Lead} \quad II = C_F V_F \cos\alpha - C_R V_R \cos\alpha \tag{3b}$$
$$\text{Lead} \quad III = C_L V_L \cos\beta - C_F V_F \cos\beta \tag{3c}$$

where C_R, C_L and C_F are scaling constants for node voltage signals V_R, V_L and V_F respectively.

To obtain the transformation scaling coefficients, the waveforms obtained in this measurement scheme are compared against those obtained in the conventional standard 3-lead wired ECG measurement configuration. As a metric for assessing the similarity of two ECG signals, the Pearson's correlation coefficient function as shown in 4 is calculated.

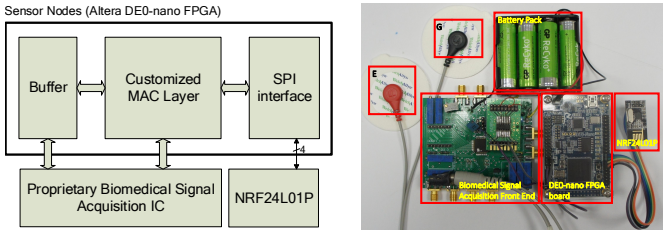
$$r = \frac{\sum_{n=0}^{N-1}(x(n)-\overline{x})(y(n)-\overline{y})}{\sqrt{\sum_{n=0}^{N-1}(x(n)-\overline{x})^2 \sum_{n=0}^{N-1}(y(n)-\overline{y})^2}} \tag{4}$$

where r is the Pearson correlation coefficient, the time series $x(n)$ is the synthesized ECG lead signal and the time series $y(n)$ is the reference ECG signal with $0 \leq n \leq N$ and N represents the number of function values in the current signal window. Subsequently, through sweeping combinations of scaling factors, the transformation scaling coefficients can be optimized by maximizing the correlation coefficients between the synthesized and reference signal waveforms.



(a) Block diagram. (b) Prototype Implementation.

Fig. 3. Base Station.



(a) Block diagram. (b) Prototype Implementation.

Fig. 4. Sensor Node.

IV. Experimental Setup and Results

A. Experimental Setup

1) System Overview: To validate the proposed measurement method, a prototype WBAN system was implemented, in which each node having two electrodes attached continuously samples ECG data at different sites of the test subject's body, and transmits the data to the base station for post-processing.

The proposed system supports the star network topology and implements a customized TDMA MAC protocol, in which a base station centrally controls all sensor nodes for data sampling, transmission and reception. Each node is granted a time slot for data transmission. This eliminates collisions in the wireless channel within the network, minimizes transmission failure and reduces power consumption. Each sensor node is running on a high frequency system clock of 5-MHz to minimize the slot timing differences among sensor nodes and the base station. Each node samples ECG data at 1-kHz, to help achieve greater time resolution and accuracy in the correlation computation process. To support the wireless communication of the proposed system, a low power high data rate transceiver, NRF24L01P from Nordic [10], is used. With a high RF data rate of 2-Mbps, system latency is minimized.

2) Base Station: Fig. 3 shows the block diagram of the base station design. In the proposed system, the base station centrally controls all sensor nodes for data sampling, transmission and reception. Data collected by the base station is stored in the Random-Access-Memory (RAM) block and sent to a logic analyzer for post processing. The base station is implemented in a FPGA development board, ML507, from Xilinx [11].

3) Wireless Sensor Node: As depicted in Fig. 4, a wireless sensor node consists of a battery pack, a proprietary biomedical signal acquisition front end integrated circuit (IC) [12], the aforementioned RF transceiver and a DE0-nano FPGA board from Altera [13], which includes the MAC layer. Two patch electrodes from WelchAllyn are connected to G_X and E_X as mentioned in section III. For each node, ECG data is sampled, buffered and transmitted under the instruction of the base station. With such a configuration, all nodes are independent with no interconnections, ensuring that a true cableless ECG acquisition system is constructed and verified.

TABLE I

Correlation Coefficients between different sets of reference data obtained at different time instances

Reference Lead Signals	Set 1 & 2	Set 1 & 3	Set 2 & 3
LEAD I	0.898	0.989	0.990
LEAD II	0.903	0.995	0.997
LEAD III	0.907	0.993	0.991

*Three sets data are acquired for each lead at different time instances.

B. Experimental Results

1) Reference Data Acquisition: Using only the proprietary biomedical signal acquisition front end, reference data for leads *I*, *II*, and *III* are obtained directly from a healthy test subject via the conventional 3-Lead ECG measurement method. However, as the electrode positions in the conventional 3-lead measurement configuration, denoted as electrodes *G1*, *G2*, *G3* in Fig. 2b, overlap with the electrodes to be used in the proposed configuration, the reference signals cannot be be obtained at the same time for direct one-to-one comparison. Signal conditioning is therefore carried out to normalize for beat variance, signal offset drifts and noise filtering in MATLAB prior to correlation coefficient computation. All correlation coefficients are obtained based on a signal window of 4096 points due to the limited acquisition depth of the logic analyzer used. The length of this signal window corresponds to approximately 4 *PQRST* complexes or heart beats.

To ensure the consistency of the reference lead signals, correlations between reference data sets obtained at different time instances for each lead were calculated. Table I summarizes the correlation coefficients across 3 different reference data sets for leads *I*, *II*, and *III*. The correlation coefficients for lead *I* signals are lower than that of lead *II* and lead *III* signals due to the relatively smaller signal amplitudes and larger noise component. However, the low variability in the correlation coefficients suggest that the ECG signals of a healthy test subject are consistent and do not vary greatly across different time periods. Therefore, it can be assumed that obtaining the lead signals at different time instances have a negligible effect on the comparison.

2) ECG Acquisition using Proposed Measurement Method: Based on equations 3 and 4 the optimized values of C_R, C_L, C_F for this test subject are 19, 3 and 1 respectively. Applying this set of transformation coefficients across a sample set of 4 measurement sets, Table II

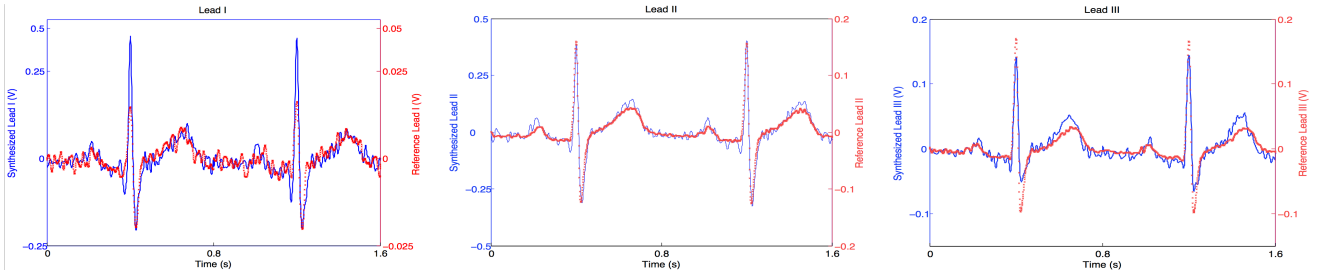
TABLE II

CORRELATION COEFFICIENTS BETWEEN SYNTHESIZED AND REFERENCE DATA FOR LEADS I, II AND III

Lead I				
Syn./Ref.	S1	S2	S3	S4
R1	0.858	0.850	0.812	0.797
R2	0.871	0.864	0.813	0.808
R3	0.836	0.817	0.782	0.766

Lead II				
Syn./Ref.	S1	S2	S3	S4
R1	0.963	0.924	0.938	0.946
R2	0.967	0.923	0.939	0.954
R3	0.967	0.927	0.939	0.951

Lead III				
Syn./Ref.	S1	S2	S3	S4
R1	0.890	0.806	0.846	0.885
R2	0.892	0.805	0.847	0.88
R3	0.890	0.805	0.845	0.884



(a) Lead I with correlation coefficient of 0.85 (b) Lead II with correlation coefficient of 0.97 (c) Lead III with correlation coefficient of 0.89

Fig. 5. Comparison between Synthesized and Reference signal

summarizes the correlation coefficient values between the synthesized lead signals with respect to the reference lead signals.

The average correlation coefficients for leads *I*, *II*, and *III* are 0.82, 0.95 and 0.86 respectively while the highest and lowest correlation coefficients obtained across the board are 0.97 and 0.77 respectively. This shows that a high degree of correlation is obtained between the signals from the new proposed measurement configuration and that through the standard 3-lead configuration. Figs. 5a, 5b and 5c show superimposed plots of the reference and synthesized lead signals for comparison purposes. A high degree of resemblance between the synthesized and reference signal can be seen and the *P* wave, *QRS* complex and *T* wave of a typical ECG signal can be clearly identified in the synthesized signals.

In addition, despite some variability in the *QRS* complex from beat to beat as the signals are obtained at different time instances, the integrity of the *PQRST* waves are retained. This indicates the potential of this system to be a truly wireless alternative to the conventional 3-lead ECG monitoring system.

V. CONCLUSION

A true wireless WBAN based 3-lead ECG acquisition system is developed. Each electrode in the standard 3-lead ECG measurement is replaced by a sensor node in the wireless network. A method has been proposed to convert the ECG signal measured at 3 nodes to the standard 3-lead ECG. High correlation coefficients for all leads have been achieved. The proposed system can be used for wearable wireless ECG monitoring and provides great comfort and convenience to the users.

REFERENCES

[1] M. Khayatzadeh, X. Zhang, J. Tan, W. S. Liew, and Y. Lian, "A 0.7-V 17.4-μW 3-lead wireless ECG SoC," in *IEEE BioCAS*, Nov 2012, pp. 344–347.

[2] L. H. Wang, T. Y. Chen, S. Y. Lee, T. H. Yang, S. Y. Huang, J. H. Wu, K. H. Lin, and Q. Fang, "A wireless ECG acquisition SoC for body sensor network," in *IEEE BioCAS*, Nov 2012, pp. 156–159.

[3] D. R. Zhang, C. J. Deepu, X. Y. Xu, and Y. Lian, "A wireless ecg plaster for real-time cardiac health monitoring in body sensor networks," in *IEEE BioCAS*, Nov 2011, pp. 205–208.

[4] N. E. Technologies, "An ECG Primer," 2003. [Online]. Available: http://www.nursecom.com/ECGprimer.pdf

[5] R. D. Chiu and S. H. Wu, "A BAN system for realtime ECG monitoring : From wired to wireless measurements," in *IEEE WCNC*, Mar 2011, pp. 2107–2112.

[6] H. Cao, H. Li, L. Stocco, and V. C. M. Leung, "Wireless three-pad ECG system: Challenges, design, and evaluations," *Journal of Communications and Networks*, vol. 13, no. 2, pp. 113–124, 2011.

[7] R. E. Mason and I. Likar, "A new system of multiple-lead exercise electrocardiography," *Am. Heart J.*, vol. 71, no. 2, pp. 196–205, 1966.

[8] T.-C. Publishing, "Diagnostic ECG-The 12-Lead(Clinical Essentials)(Paramedic Care) Part 1," 2013. [Online]. Available: http://what-when-how.com/paramedic-care/diagnostic-ecgthe-12-lead-clinical-essentials-paramedic-care-part-1/

[9] H. Cao, H. Li, L. Stocco, and V. C. M. Leung, "Demonstration of a novel wireless three-pad ECG system for generating conventional 12-lead signals," in *Proceedings of the Fifth International Conference on Body Area Networks*, 2010, p. 29.

[10] N. Semiconductor, "nRF24L01 Product Specification v2.0," 2007. [Online]. Available: http://www.nordicsemi.com/eng/content/download/2730/34105/file/nRF24L01_Product_Specification_v2_0.pdf

[11] Xilinx, "ML505/506/507 Evaluation Platform User Guide," 2011. [Online]. Available: http://www.xilinx.com/support/documentation/boards_and_kits/ug347.pdf

[12] H. Wu and Y. P. Xu, "A 1V 2.3μW Biomedical Signal Acquisition IC," in *IEEE ISSCC*, 2006, pp. 119–128.

[13] Altera, "DE0 Nano User Mannual," 2012. [Online]. Available: ftp://ftp.altera.com/up/pub/Altera_Material/12.1/Boards/DE0-Nano/DE0_Nano_User_Manual.pdf