Accessing Differential Measures with a Conjugate Coil-pair for Wireless Resistive Analog Passive (WRAP) ECG Sensors

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***Abstract*—Wearable devices and mobile-based biomedical applications have increased the demand of finding an alternative easy-to-use method for monitoring body signals. One approach towards this challenge is a zero power (battery-less), fully-passive electronic patch sensors that can be attached to the body. We have previously reported a novel technique for Wireless Resistive Analog Passive (WRAP) sensors for various body signals such as core body temperature, heart rate, etc. This passive sensor consists of a loop antenna, a tuning capacitor, and resistive-based biopotential transducer. The wireless scanner transmits an RF signal at frequency 8.37 MHz, which is amplitude modulated based on the resistive changes by a transducer at the WRAP sensor. The envelope of the modulated signal represents the body signal to be captured and can be analyzed on the scanner or downstream on the user's smartphone. In this work, we first demonstrate the capability for differential signal capture, such as electrocardiogram (ECG or EKG) using a novel conjugate coil pair technique. The WRAP sensor uses a dual-gate MOSFET (depletion mode) to convert the biopotential signal to the correlated resistive variation of the source to drain resistance. These two circuits power the conjugate coil pair, which cancels common mode signal and only transmits differential mode signals (viz. ECG). The results show that connecting a pair of sensors in this way could allow accurate measurement of a differential biopotential. This work demonstrates voltage sensitivity down to 40 µV towards realizing a battery-less, body-worn WRAP ECG sensor for monitoring ECG signals.**

***Index Terms*— Wireless sensor; Passive sensor, Body-worn sensor, ECG signal.**

I. Introduction

Wireless and fully passive sensors can be suitable in situations where a wired connection is inconvenient or impractical such as body-worn sensors, implantable sensors, and remote sensors [1]. A passive wireless transensor was developed back in 1967 for measuring the intraocular pressure using a pair of spiral coils implanted inside the eye [2]. Passive sensors can eliminate contact wires and batteries, therefore, continuous patient monitoring in day-to-day life would be more practical and can be designed as body-worn sensors to collect physiological signals unobtrusively [3]-[5]. Our developed Wireless Resistive Analog Passive (WRAP) sensors [6,7] can transmit analog signals without any digital chips, which are required in other types of wireless digital passive sensors. The novel WRAP sensors use inductive coupling between two printed spiral coils (PSCs), and have been demonstrated to capture various physiological signals, such as heart rate, respiration rate, and core body temperature [6-9]. However, some of the physiological signals, such as ECG signal, require a differential input of two ECG electrodes. Differential amplifiers reject most of the common mode noise, while they also amplify differences between two signals such as an ECG signal. In this work, we have designed a novel WRAP sensor for measuring differential ECG signals using one primary coil and two conjugate coil-pairs as secondary. The differential inputs are connected to two MOSFETs that change loading of the two secondary coils. As a result, the common signal of both sensor coils are canceled out while differential signals are transmitted and lead to the ability to access only differential biopotential signals that modulate the carrier signal at the primary. The primary and secondary coil designs have been optimized and improved by our research group previously [8].

II. Hardware Description

The concept of WRAP sensor for biopotential signal access is shown in Fig. 1a, while the WRAP sensor equivalent is a parallel combination of RLC (Fig. 1b). Three spiral coils have been used in this work as shown in Figure 1c. Two identical coils (L2 and L3) were used as passive sensors and one coil (L1) as a scanner device. Dual gate depletion mode MOSFET was used for biopotential capture due to its high sensitivity for small input voltages (Vin). The input voltage is converted to a correlated resistive variation of source-drain resistance of the MOSFET (R_{SD}). The primary board has two SMA connectors, one for the carrier input that oscillates at an RF carrier wave (8.37 MHz used in this work), and the other one is across the coil L1, that carries the modulated signal to another other PCB with electronic circuits for detecting, filtering, and amplification. The capacitor C1 is used to match the antenna to 50Ω and C2 is used to adjust the resonance frequency with L1. The loop antennas (L1, L2, and L3) are designed as a planner spiral coil (PSC) and can be modeled as inductors. The passive sensors have identical circuits where C3 and C4 are used to tune the antenna at the resonance frequency. Gate-1 of the MOSFETs M1 and M2 were connected to the body (e.g. two channels to capture an ECG signal), while Gate-2 was connected to the coupled signal. This configuration increased sensitivity for Vin sensing (µV range) connected at Gate-1.

Cadence Allegro (Cadence Design Systems Inc., San Jose, CA, USA) was used to design the Printed Circuit Boards (PCBs). The outer diameter of the primary and the secondary coils were 40 and 20 mm, respectively. The planner coils of the sensor boards were placed to face each other and Kapton tape was used as an isolation material as shown in Figure 2 (a). A fixture was designed to keep the boards in parallel at a co-axial position Figure 2 (b).

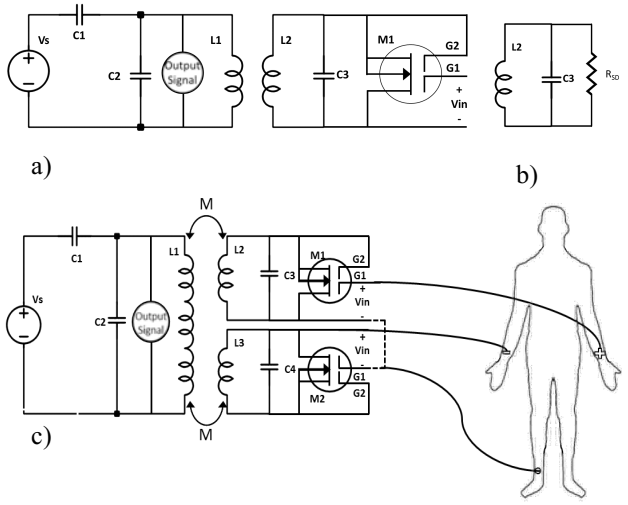

Fig. 1. a) Schematic of the primary and the secondary coils for biopotential sensing. b) The equivalent circuit of secondary. c) Two WRAP sensors are used for sensing of the differential input voltages to access ECG signals from a human body.

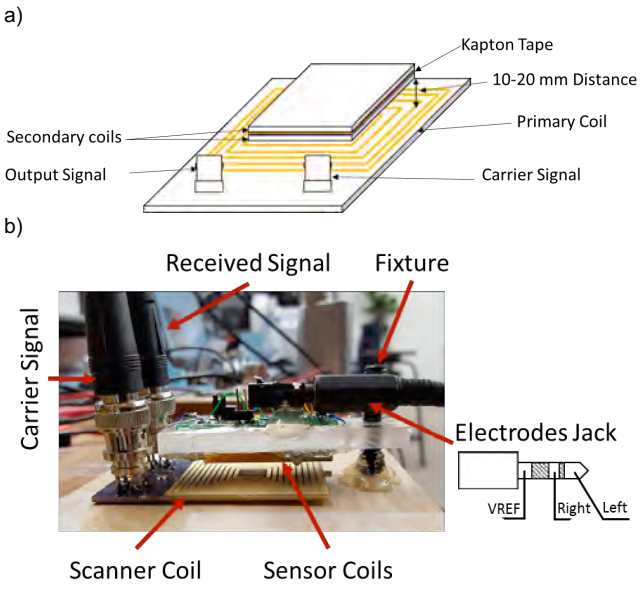

Fig. 2. a) Diagram depicting the conjugate coil paper (separated by Kapton tape), and the scanner coil positioned at the co-axial position. b) A photograph of the setup used for the scanner and the sensors to capture signals in differential mode.

III. Theory and Method for Data Collection

Each passive sensor has an equal impedance as given in Equation 1 and 2 [10]. In our case, R_{SD1} and R_{SD2} are the corresponding resistive variations of MOSFET 1 (M1) and MOSFET 2 (M2), respectively.

$$Z_{sen1} = \left(jwL_2 \,||\, \left(\frac{1}{jwC_3}\right) || \, R_{SD1}\right) \quad (1)$$

$$Z_{sen2} = \left(jwL_3 \,||\, \left(\frac{1}{jwC_4}\right) || \, R_{SD2}\right) \quad (2)$$

As both sensors are identical ($Z_{sen1} = Z_{sen2} = Z_{sen}$), each sensor's impedance can be seen from the primary side as given in Equation 3, where *M* is the mutual inductance.

$$X_{sen} = \left(\frac{(wM)^2}{Zsen(w)}\right) \quad (3)$$

Therefore, the total impedance that can be seen by the output signal is equal to Equation 3:

$$Z_{Out} = \left(X_{sen1} + X_{sen2} + jwL1 \,||\, \frac{1}{jw(C1+C2)}\right) \quad (4)$$

Equation 3 shows that any change in the resistive load creates a change in the received signal amplitude (provided all other variables remain fixed). In our setup, the PSC for sensor 1 is in reverse of that of PSC for sensor 2, forming a conjugate coil pair. If the polarities of V_{in1} and V_{in2} are opposite, the received signal will be modulated by the voltage difference as X_{sen1} and X_{sen2} changes in the same way due to conjugate coil pairs. On the other hand, when the polarities are the same, the effect cancels out as X_{sen1} and X_{sen2} changes in opposite way, thus all of the common mode voltages are suppressed. This unique technique allows us to capture differential signals like ECG.

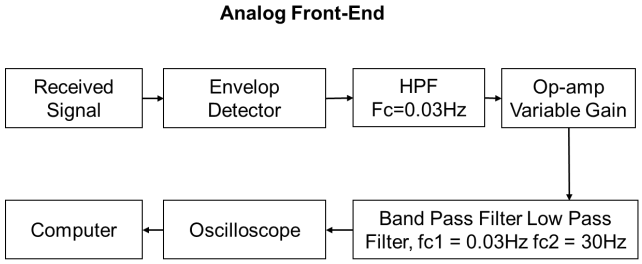

Fig. 3. ECG signal capture process and analysis flowchart

For experiments, a signal generator (Model: DG4162, Rigol Technologies Inc., Beijing, China) was used to generate carrier signal to interrogate the passive sensors. The received signal was detected using an envelope detector followed by a unity gain voltage follower with a high pass filter, a variable gain amplifier, and a bandpass filter with fc1 = 0.03 Hz and fc2 = 30 Hz as shown in Figure 3. The output is connected to an oscilloscope (Model: MDO3024, Tektronix, Inc., Santa Clara, CA,

USA) and the captured data from the oscilloscope is saved in .csv file format. This data is later compiled, analyzed, and plotted in Matlab (Mathworks Inc, Natick, MA).

A. Test Bench experiments

In the first set of experiments, only one sensor board was used as shown in Figure 1 (a) to check the sensitivity of the MOSFET when the applied input voltage is varied. Another set of test bench experiments were conducted to investigate the differential passive sensor setup (two sensors with conjugate coil-pair) and to determine the responses to the differential mode and the common mode input voltages. Two function generators (Model: DG4162, Rigol Technologies Inc., Beijing, China) were used in these experiments to generate V_{in1}, V_{in2} and the carrier signal. All ground terminals of the secondary boards were common.

B. ECG Measurement

The performance of the passive sensor was further investigated by recording ECG signals and compared to the quality of ECG signal measured using an open source commercial hardware EKG/EMG shield (OLIMEX Ltd, Bulgaria) as shown in Figure 4 (right). *In vivo* ECG data was captured using commercial gel type ECG electrodes Ag/AgCl (GS-26) for the passive sensor while dry electrodes for the EKG/EMG shield as shown in Figure 4 (left). The left arm was connected to the V_{in1}, right arm to V_{in2}, and right leg to the reference point as shown in Figure 1 (c). The shield EKG/EMG converts the analog differential signal attached to CH1_IN+/CH1_IN- inputs into a single stream of data as output. The output signal was connected directly to the oscilloscope. All the ECG measurements were taken at the same time from both sensors.

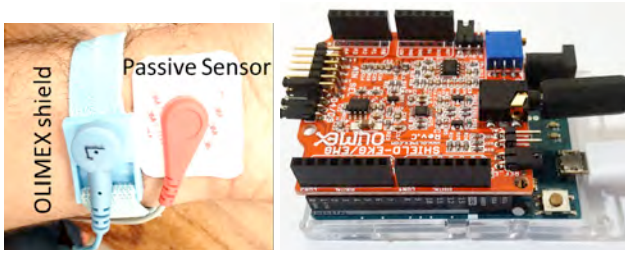

Fig. 4. ECG electrode setup for signal capture (left) and EKG/EMG shield from OLIMEX connected on an Arduino Uno board (right).

IV. RESULTS

A. Test Bench Experiments

In the first set of experiments, only one sensor board was used where 100 and 40 µV were applied to the input voltage using 40 dB BNC attenuator at a frequency of 100 Hz. Figure 5 shows the received signals when 100 µV was applied before and after processing using Matlab.

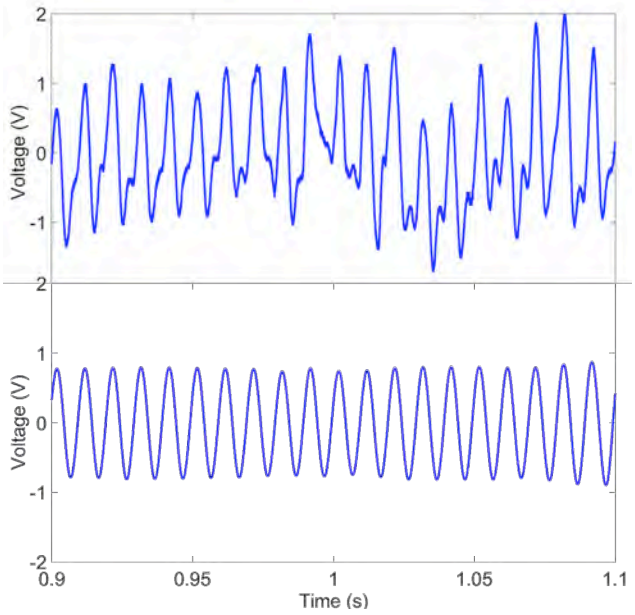

Fig. 5. The output signal of applying 100 µV. Raw (Top) and filtered data (Bottom).

The first experiment of the differential mode setup was conducted to measure the CMRR for the differential sensor. For this experiment, the received signal was connected directly only to a unity gain low pass filter to remove the carrier signal. The average common (A_{CM}) and differential (A_{DM}) mode gains are 0.1703 and 1.24 respectively, therefore, the CMRR is 17.24 dB. Figure 6 shows a linear response of the sensor when one input is fixed and the other input is linearly changed.

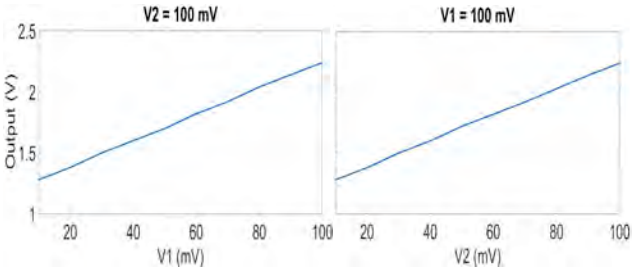

Fig. 6. The response of the differential sensor when one input fixed and the other one changes.

Figure 7 shows the received signal (Blue) as the phase between V_{in1} (Green) and V_{in2} (Red) is equal to 0, 90, and 180 degrees, respectively. Both V_{in1} and V_{in2} are sinusoidal signals with a peak-to-peak voltage of 100 mV and frequency of 100 Hz. The results show that the output signal is significantly attenuated when both input signals are identical, while the output signal is amplified when they have a maximum difference.

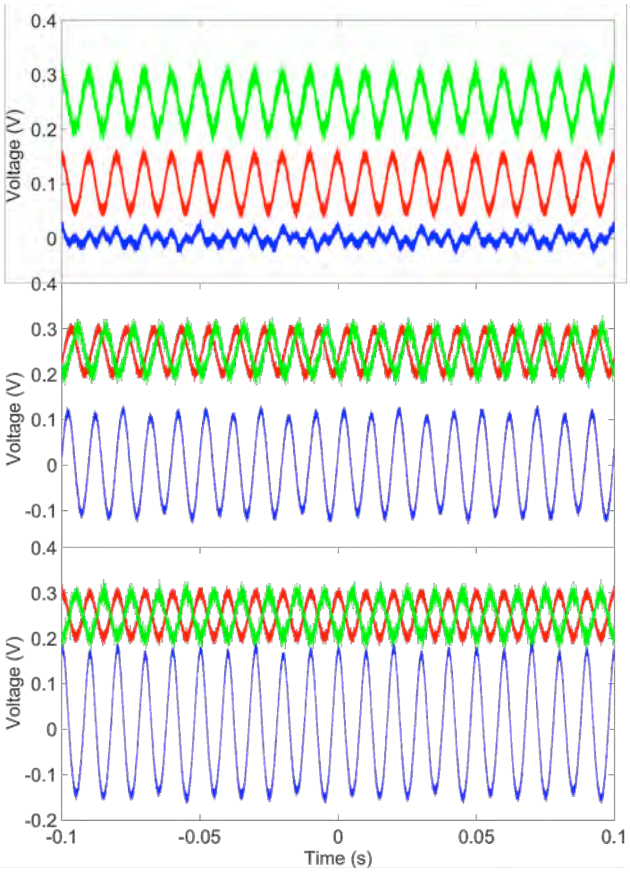

Fig. 7. The output mode of the differential mode setup. The phase difference between V_{in1} and V_{in2} a) 0 b) 90 degree and c) 180 degree.

B. ECG Measurements

The setup in Figure 1(c) was used for ECG signal measurement, where the reference of input voltages was connected together and then connected to the right leg. The setup in Figure 1(c) was used for ECG signal measurement, where the reference of input voltages was connected together and then connected to the right leg. Figure 8 shows the raw data of the ECG signal recorded using EKG/EMG shield (Red) and the passive sensor (Blue) for 10 seconds (100k points at sampling frequency 10k sample per second). Some noise can be seen in the unfiltered signal using time-frequency analysis of ECG EKG/EMG shield (Left) and the passive sensor (Right). The ECG signals recorded using the passive sensor had more components of the utility line 60Hz noise and its harmonics. All the ECG components (R-peaks, QRS-complexes, T-waves, and P-waves) are easily recognizable from the raw data without any downstream processing as shown in Figure 9 (Top). A low pass filter was applied to the signal using Matlab for both signals and the results are depicted in Figure 9 (Bottom).

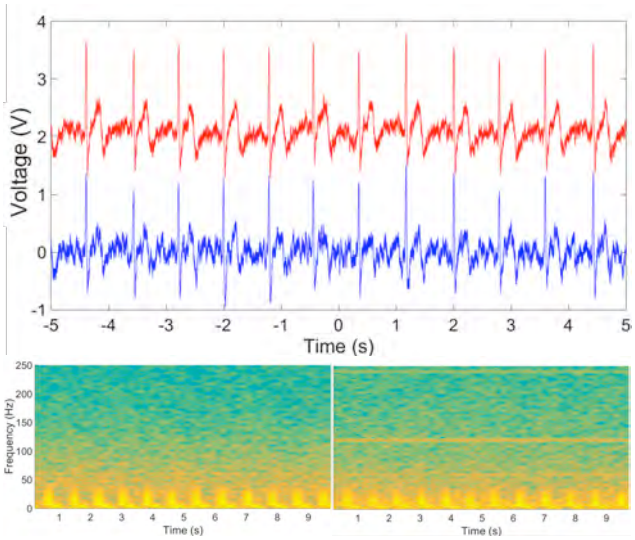

Fig. 8. (Top) The raw data of the recorded ECG signals using EKG/EMG shield (Red) and the passive sensor (Blue) for 10 seconds. (Bottom) the time-frequency analysis of ECG using EKG/EMG shield (Left) and the passive sensor (Right)

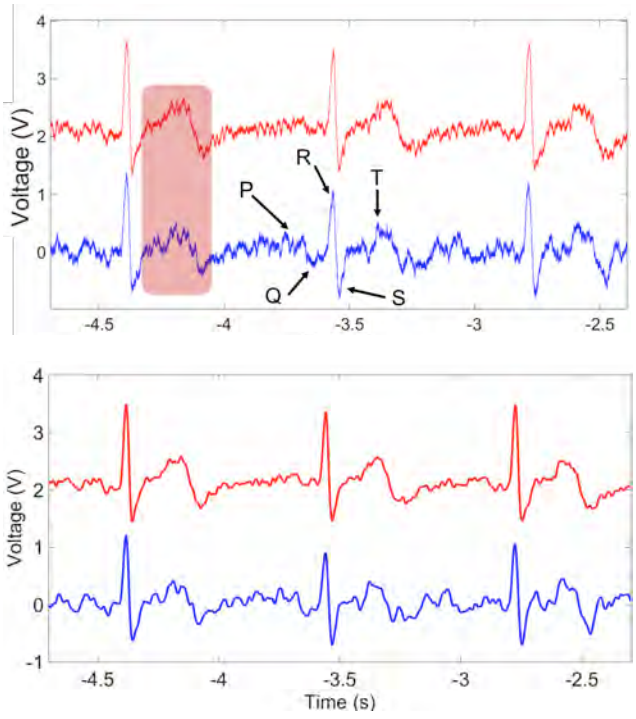

Fig. 9. (Top) The ECG components (R-peaks, QRS-complexes, T-waves, and P-waves). (Bottom) and the filtered ECG signal.

V. Conclusions

In previous works, our group had reported a novel fully passive electronic body-worn patch sensor (WRAP sensor) that was demonstrated to capture heart rate, core body temperature, and other physiological signals, as well as biopotentials using MOSFET. In this work, we described

for the first time a novel setup (conjugate coil-pair) of WRAP sensors with two WRAP sensors utilizing depletion mode MOSFETs for measuring ECG signal from limb lead configuration. The system successfully captured biopotentials as low as 40 to 100 μV, in addition to the ECG signal *in vivo*. The results demonstrate the promise of developing a battery-less WRAP ECG sensor that can be worn on the body or even possibly be implanted inside the human body and would be a suitable for continuous ECG recording.

VI. Acknowledgement

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