Battery-Free Multichannel Digital ECG Biotelemetry using UHF RFID Techniques

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***Abstract*—We propose to leverage UHF RFID techniques to yield a continuously wearable, battery-free wireless multichannel ECG telemetry device that is potentially disposable, low-cost and suitable for integration with multiple electrodes in a flexible circuit assembly. Such a device could have broad applicability, ranging from initial patient assessment by first responders, to continuous monitoring in various clinical settings. We employ a recently described single-chip data acquisition system including RF power harvesting to eliminate the need for a battery. The single-chip system includes 14 channels of integrated biopotential amplification, an 11-bit ADC, and a 5 Mbps digital backscatter telemetry link. We present an initial characterization of the telemetry chip in this application including battery-free, wireless 3 and 5 channel ECG recordings made from an ambulatory human subject at a range of ≈ 1 meter.**

I. Introduction

The electrocardiogram (ECG) is one of the most common forms of noninvasive diagnostics. In an ECG recording, the electrical activity of the heart is inferred from low-amplitude electrical potentials present on the surface of a patient's skin during actuation of the heart muscle. Multiple sensing electrodes are often used to observe certain linear combinations of potentials from the heart muscle. These potentials are weighted and summed given a model of body conductivity to partially isolate activity to particular regions of the heart. ECG supports the diagnosis of cardiac abnormalities, and in some cases can pinpoint the affected area of the heart and the type of abnormality present. There are several different ECG configurations, using different numbers of electrodes and different electrode placement, each providing somewhat different views of heart activity [1]. The most common ECG configurations are 3-lead and 5-lead configurations, which are used in all settings ranging from emergency medicine to ambulatory monitoring. Less common is a 12-lead configuration which is used for detailed diagnosis in patients with certain indications. Regardless of the type of ECG, monitoring a patient with typical ECG equipment generally requires a wiring harness that connects electrodes on the patient's skin to an external device containing signal conditioning, digitization, and processing. Existing ECG monitors require some type of battery or line-connected power source to provide these functions.

The wiring harness connecting a patient to an external ECG monitor poses a significant problem for monitoring ambulatory activities and in long-term monitoring, due to the potential for discomfort and impeded movement. A particular risk for small infants or unresponsive patients is the potential for strangulation in the wiring harness. The need to disconnect and reconnect the wiring harness when transferring a patient from one work area to another (e.g. from an ambulance to a hospital exam room) adds inconvenience and may result in lost data.

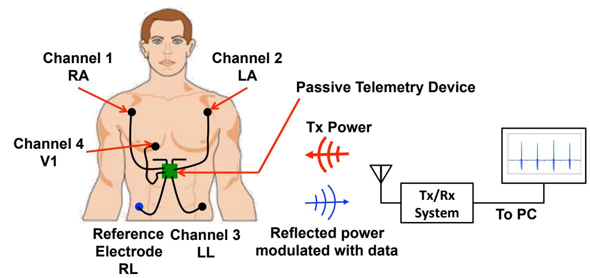


Fig. 1: System diagram with proposed 5-lead ECG setup

To avoid the problems posed by an ECG wiring harness, we propose a continuously wearable, wireless, battery-free multichannel ECG sensor. Like a passive RFID tag, the device extracts its operating power from the incident RF field of the external system. This sensor physically decouples the patient from an external ECG monitor. A diagram showing the passive telemetry device system is shown in Figure 1. The entire system is suitable for packaging as a flexible circuit, yielding a low-cost, light-weight, and potentially disposable sensor. The ECG device can remain attached to the patient during transport, when moving from one domain of care to another, in bed, etc.

II. Related Work

Wearable wireless ECG sensors such as those in [2], [3], focus on monitoring an individual in his or her home, and are capable of communicating with the hospital if necessary. Reducing the power consumption of wearable wireless ECG systems has also been investigated. Using body tissue to conduct current to the wireless transmitter was explored in [4]. Park *et al* [5] focused on the application of capacitive sensing for ECG as a means of power reduction. These systems all require a battery for operation.

Reliance on a battery comes with the disadvantages of more maintenance, a larger and heavier device, and a more expensive sensor system. A passive sensor device would remove the need for a battery, creating a low-cost, low-power, low-maintenance, potentially disposable continuously wearable ECG monitoring system.

Yeager *et al* present a battery-free low-power wireless telemetry device that is able to record an *in-vivo* muscle temperature from a hawkmoth in flight [6]. This device consumes 9 μA while active, and 16.2 μW of power while actively transmitting a muscle temperature from a flying hawkmoth. The telemetry device transmits a single temperature measurement and is not capable of multichannel operation. A low-cost disposable tag for heart monitoring has been developed by Mandal *et al* [7]. This device utilizes microphones to listen to the heart's activity, recording a phonocardiogram (PCG) rather than an ECG. Unlike the present work, [7] does not telemeter the full PCG waveform. Instead, it detects spikes in the heart's activity and transmits only this information. The battery-free device presented in this work transmits full ECG waveforms on multiple channels while being worn by the user, and can be integrated with flexible circuit packaging for a low-cost disposable sensor.

III. Nature of ECG Signals

Electrocardiograms can be classified by the number of electrodes used, with typical recordings being 3-lead, 5-lead and 12-lead ECG. Three and 5-lead ECG are preferred for continuous or bedside monitoring, while 12-lead ECG is used for diagnostic monitoring to provide a more complete picture of heart activity [1].

This work is focused on 5-lead ECG. Electrodes are placed on the torso as shown in Figure 1. The electrode naming scheme is the US standard for ECG [1]: right arm (RA), right leg (RL), left arm (LA), left leg (LL), with V1-V6 being precordial leads close to the heart. Four electrodes are placed on the torso to avoid noise from muscle movement. The fifth electrode is V1, which is placed in the 4th intercostal space on the right side of the chest.

Signals are derived from the measurement of the voltage between pairs of electrodes. The standard limb leads provide 3 different views of the heart and are computed as follows:

$$\begin{aligned} \text{Lead I} &= LA - RA \\ \text{Lead II} &= LL - RA \\ \text{Lead III} &= LL - LA \end{aligned} \tag{1}$$

The precordial leads are referenced to an average point over the body known as Wilson's Central Terminal, a combination of the electrodes placed over the extremities defined in [8]:

$$\text{WCT} = \frac{1}{3}\left(LA + RA + LL\right) \tag{2}$$

Each precordial lead is compared to Wilson's central terminal to produce the ECG waveform for that point.

A typical ECG waveform is shown in Figure 2. The **P**-wave, a slight bump before the first negative or positive deflection, represents atrial depolarization. The **Q**, **R** and **S**-waves make up what is known as the **QRS** complex, and represents the rapid depolarization of the right and left ventricles. The **T**-wave represents the repolarization, or recovery, of the ventricles.

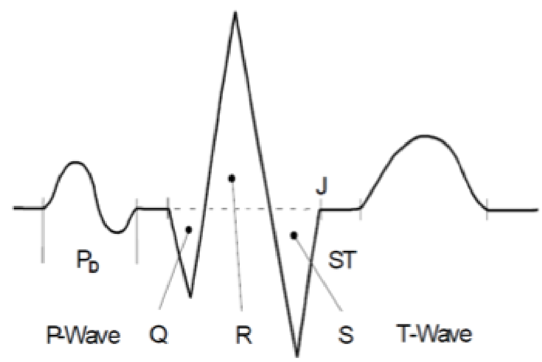


Fig. 2: Typical ECG waveform with labeled features, reproduced from [9]

IV. Digital Telemetry Chip

The main component of the passive telemetry device is a custom amplification and telemetry chip, developed at Intan Technologies, which is responsible for harvesting power from the incident RF field and regulating power to the integrated components, amplifying multiple channels of data and digitizing and encoding the data to transmit back to the external system. A detailed description of the telemetry chip can be found in [10].

A die photo of the telemetry chip is shown in Figure 3. The die measures 2.36 mm x 1.88 mm x 250 μm. The chip has an array of 14 differential amplifier inputs, capable of amplifying weak biopotentials, as well as two DC amplifiers. The data to be transmitted is digitized with an 11-bit ADC, deriving its clock from an external 20 MHz quartz crystal. The telemetry device employs an on-chip RF power harvester, consisting of a 4-stage Schottky diode voltage multiplier. An LDO voltage regulator maintains the regulated voltage at 1.23 V. A binary phase shift keying (BPSK) technique is used for communication, and the modulation depth is configurable. The entire chip consumes 1.23 mW of power while actively transmitting data from all 16 channels at 5 Mbps.

V. Telemetry of ECG Signals

A. Experimental Setup

To provide power and a means for backscattered communication for the passive telemetry device, both a transmit and receive system are necessary to produce the proper interrogation field and a means to decode the backscattered data.

TABLE I: Forward-Link Budget

Channels Used	Estimated Power Savings	RF power	Estimated Read Distance
16 (All)	(reference)	+7.76 dBm (measured)	1.20 m
4 (5-lead ECG)	0.24 mW	+6.77 dBm (estimated)	1.34 m
3 (3-lead ECG)	0.26 mW	+6.68 dBm (estimated)	1.36 m

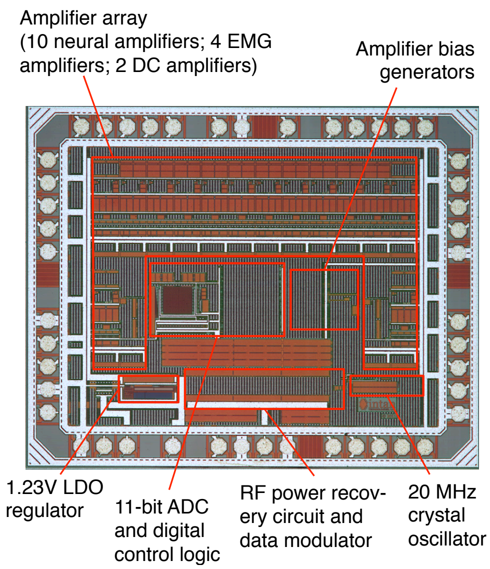


Fig. 3: Die photo of telemetry chip

1) Transmit Subsystem: To create the interrogation field to provide power and a carrier for the passive telemetry device to backscatter ECG data to the external subsystem, the transmit subsystem in Figure 4 was used.

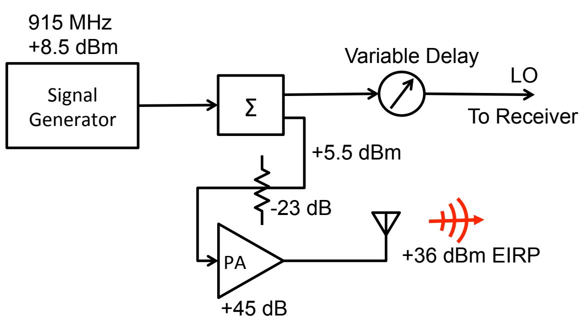


Fig. 4: Diagram of transmit system

The telemetry chip operates in the UHF band, so a signal generator (Agilent N9310A) was used to generate a 915 MHz, +8.5 dBm signal. This signal is then directed through a splitter (Mini-Circuits ZN2PD-920W-S+), providing a local oscillator (LO) signal to the receive subsystem. The LO signal passes through a variable delay before reaching the receive subsystem, which allows for rotation of the IQ constellation. The other half of the split signal is amplified before reaching the transmit antenna (L-Com HG908P) such that the output power is +36 dBm, which is the limit according to the FCC rules Part 15.247 [11].

Table I presents the forward-link budget for the test system, assuming a gain of 5 dBi for the dipole with body reflector and based on the measured harvester and mismatch efficiency of 20.6%, as well as a transmitted power of +36 dBm EIRP. The operating range of approximately 1 m is limited by the power consumption of the chip and the antennas and transmitter employed. In our experiments, the chip transmits data from 16 total channels, 12 of which are unused in the 5 lead ECG. The operating range could be improved by disabling the unused channels to reduce power consumption, by improving antenna gain, or by increasing the transmitter power. Operating range could also be improved by reducing the sampling rate below the 26.1 kSps currently employed, at some loss of fidelity in the **QRS** complex.

2) Receive Subsystem: A block diagram of the RF receiver is shown in Figure 5. This is a direct conversion design with a digital baseband section. We expect our system to be forward-link limited, so the receiver's analog front end (AFE) is optimized for large-signal performance at the expense of noise figure. The AFE has sensitivity better than -70 dBm at a BER of 10^{-5} at a data rate of 5 Mbps [10]. The baseband signals are filtered by an identical pair of bandpass filters such that out-of-band noise is rejected. The receive signal is then decimated and rotated to allow bit-slicing of the BPSK-encoded signal. The clock and data are extracted before being transmitted to the PC to be decoded by a digital phase locked loop and symbol correlators. The base station interfaces with a PC through a USB 2.0 port, for the PC serves as a platform to run software that is responsible for decoding the incoming bit stream.

B. Two-lead Wearable Wireless Passive ECG Recording

An initial test using the device in a 2-lead wearable setup was performed to validate signal quality, prior to setting up a more complex 5-lead configuration. The telemetry chip was placed on an evaluation board so the transmit channels could be easily accessed, as well as providing a platform for external components. A dipole antenna was used on the evaluation board to harvest power and backscatter ECG data to the external subsystem. A picture and block diagram of the passive telemetry device can be seen in Figure 6 and Figure 7, respectively.

Two electrodes were used for this experiment. One lead was placed on left leg (LL), the other on the right arm (RA). These electrodes were connected to two channels on the telemetry chip. The entire evaluation board was then mounted on a styrofoam block to use the body as a reflector backing the dipole antenna, as shown in Figure 8. The subject stood 1 meter from the transmit antenna with the electrodes on the body connected to the passive telemetry device's channels.

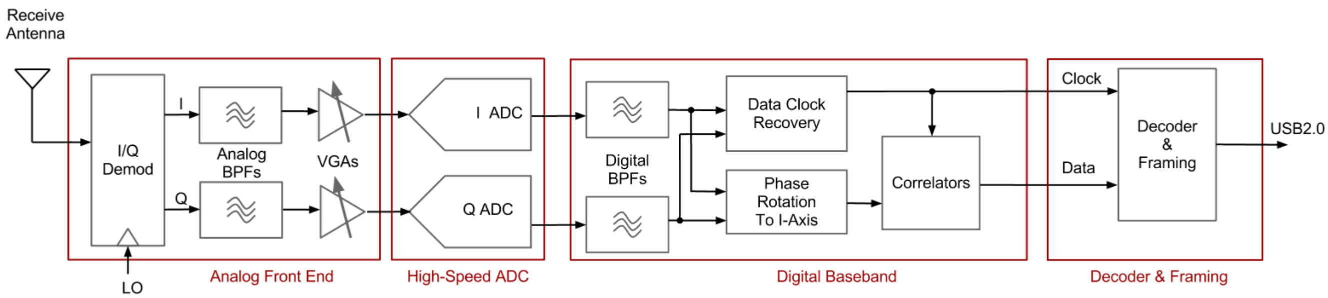
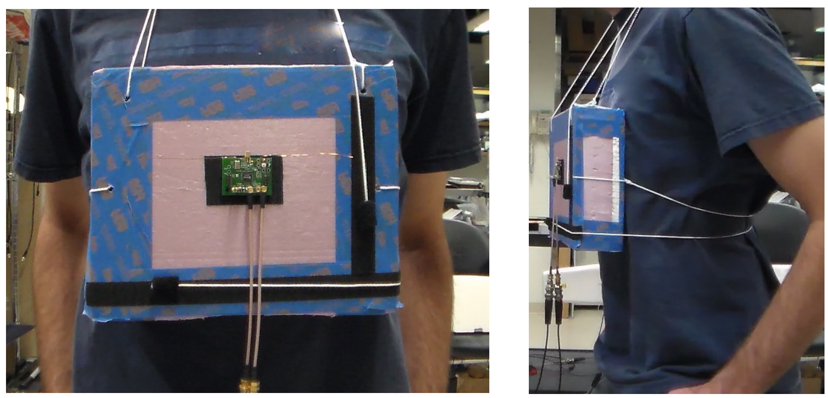


Fig. 5: Receiver architecture



(a) Front view

(b) Side view

Fig. 8: Use of the passive ECG telemetry device in a 2-electrode wearable configuration

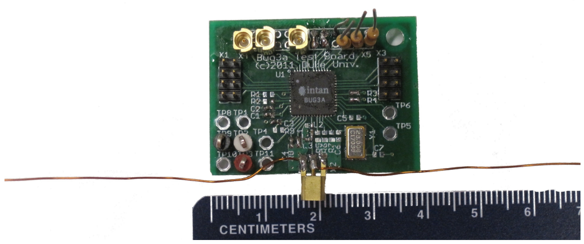


Fig. 6: Telemetry chip evaluation board with dipole antenna

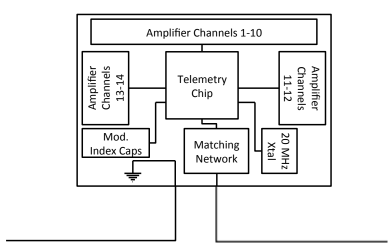


Fig. 7: Block diagram of the telemetry chip evaluation board

Data was recorded in this manner and processed offline using Matlab. Each electrode (channel) was recorded separately and then combined in post-processing to form ECG lead II. This data was filtered with an FIR bandpass filter from 0.05 - 40 Hz. Post-processing was employed to maximize experimental flexibility. While analog or digital filtering of the ECG waveform could be added to the telemetry chip, the added circuitry would lead to increased complexity and power consumption. This tradeoff will be considered in future work.

Figure 9 shows a few heartbeats, as evidenced by the clear **R**-wave spikes. The heart rate of the individual can be determined by the **RR**-interval, the distance between **R**-waves, and extrapolated to determine the heart rate in beats per minute (bpm). For this particular recording of ECG lead II, the subject's heart rate is approximately 75 bpm, a normal resting heart rate.

To verify that the passive telemetry device is recording waveforms resembling those of a standard ECG, a closeup of one of the heart rate "blips" can be viewed. Each individual heart rate waveform should resemble that of the standard waveform for lead II. The salient features of a typical lead II ECG waveform can be seen in Figure 10a and are labeled accordingly. The most prominent feature of this waveform, the **QRS** complex (section formed by the **Q**, **R** and **S** waves), can be seen clearly. Additionally, the **P** and **T** waveforms can be observed in the closeup of the waveform. Figure 10b shows the noise during a similar time interval with the electrode grounded. The RMS voltage of the ECG signal is 10.7 μV, and the noise RMS voltage is 0.35 μV, leading to an SNR of the filtered received signal of approximately 29.6 dB.

A 2-lead ECG was recorded while the subject simultaneously wore a commercial heart rate monitor from Timex®.

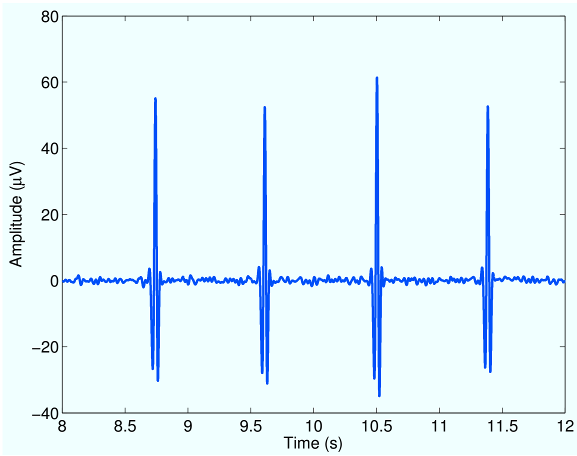
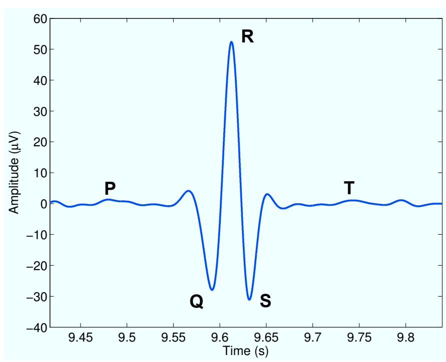
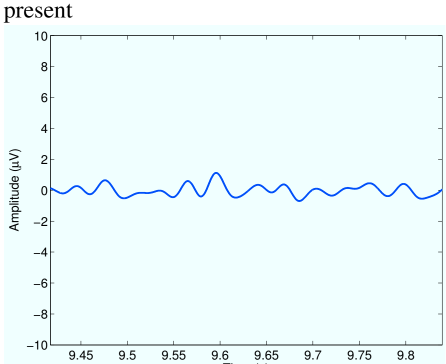


Fig. 9: Sample of fully passive wireless 2-lead ECG, filtered and processed offline



(a) Filtered signal at lead II, heart beat present



(b) Filtered signal at lead II, electrode grounded (zoomed-in voltage scale)

Fig. 10: Recordings from Lead II, single heart beat interval.

While data was being recorded from the passive telemetry device, the heart rate reading from the commercial heart rate monitor was recorded as well. The ECG waveform was processed offline, and the heart rate as measured by the passive telemetry device was determined by computing the time difference between consecutive peaks, **R**-waves, and converting this period into heart beats per minute. The comparison between the heart rate as measured by the ECG waveform from the passive telemetry device and that from the commercial heart rate monitor can be seen in Figure 11.

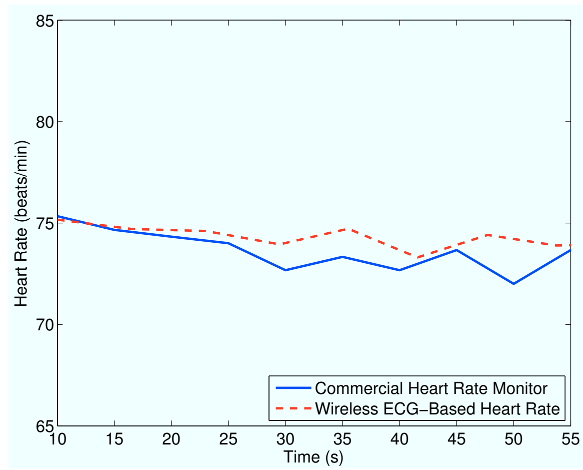


Fig. 11: Comparison between heart rate measured from wireless ECG data and ground truth measured by Timex® heart rate monitor

To compare the heart rate from the ECG waveform, a moving average filter was applied to the instantaneous heart rate from the ECG waveform such that a heart rate recorded every 5 seconds was produced. During the interval of recorded data, the heart rate as measured by the passive telemetry device falls within approximately 2.5 bpm of the commercial heart rate monitor, or approximately 3.5%.

C. 5-lead ECG Recording

To record multichannel ECG data, the electrodes as shown in Figure 1 are used for a 5-lead ECG. In this 5-lead ECG setup, the electrode on the right leg (RL) is used as a common reference for the other 4 channels. The waveform for each electrode referenced to the right leg (RL) is recorded, and the waveforms are then combined and filtered in post-processing to form leads I, II, and III, and precordial lead V1 as described in Section III. While the data is currently recorded and post-processed offline in the prototype, we believe it is feasible to implement realtime filtering and processing of ECG data.

The device was placed on a plastic cart while the subject, attached appropriately to the device, stood behind the cart. A rectangular patch antenna (L-Com HG908P) was used for the passive telemetry device, and placed 1 meter from the transmit antenna.

All 4 channels were transmitted simultaneously in this 5-lead ECG (the 5th lead being the RL, common ground). After the data was recorded and post-processed offline, the data shown in Figure 12 results. The same filtering techniques used for the 2-lead data were applied here. The **QRS** complex representing each heartbeat can be clearly seen in each lead and the precordial lead, and the subject's heart rate can be tracked by the **RR**-interval as **R**-wave timing is preserved by the telemetry device. Moreover, **R**-wave timing among channels is consistent, as shown in Table II. The average **RR**-interval and equivalent heart rate is nearly identical among the four recorded channels. The peak **R**-wave amplitude is reported, as the relative amplitude among lead traces is an important element of ECG waveforms and can lead to various medical diagnoses.

A closeup of the 5-lead ECG data is shown in Figure 13. The prominent features of a standard ECG waveform are visible among the recorded data and labeled accordingly. The recorded waveform for Lead II as shown in Figure 13b for the multichannel setup appears similar to that recorded in the two-lead setup, as shown in Figure 10a. These two traces were taken using the same two electrodes, and are both Lead II as defined by standard ECG terminology. In moving from a 2-lead to a 5-lead setup, there is minimal difference in comparable traces; the standard ECG features are visible in both Figure 13b and Figure 10a, especially the prominent **QRS** complex. The peak of the **R**-wave is approximately 30 μV in the 2-lead setup and 34.2 μV in the 5-lead setup.

VI. Discussion

The proposed passive wireless multichannel ECG system provides for a low-cost, low-maintenance, disposable ECG sensor. While the data presented here show 2-lead and 5-lead ECG measurements, this can easily be extended to a full 12-lead ECG system, as the passive telemetry device supports up to 14 channels of simultaneous data.

A passive wireless multichannel ECG system would remove the need for a patient to be directly connected to an external machine, reducing the mass of bulky wires typical of ECG systems. This would reduce the bulky wire harness used during ambulatory monitoring, where the wire harness can become a burden and prevent normal motion. Small infants or unresponsive patients would also benefit, as there will be a reduction of the potential for strangulation. First responders could quickly and easily attach a passive wireless ECG device to a patient, intermittently querying the patient's state during hospital transport.

The forward-link limited operating range of the telemetry device is approximately 1 meter. Adding a battery to operate semi-passively could increase the operating range. Assuming a receiver sensitivity of -70 dBm, and a backscatter ratio of -15 dB, the estimated return-link-limited range of the telemetry device would be $\approx$ 7.8 m in free space. Unfortunately the cost, size and weight, and limited lifetime of a battery may be undesirable in many scenarios.

A proposed application of the passive wireless ECG system is shown in Figure 14. In this scenario, the patient is decoupled from the external system, removing the need for a bulky wire harness as well as allowing the patient to move about freely without the need to disconnect and re-connect themselves to the ECG monitor. Moreover, transport of the patient is simplified, as they can be moved from one location to another without the need to remove and reconnect wiring leads, provided the new location has the proper antenna setup. Removal of the wiring harness also gives medical personnel more room to work.

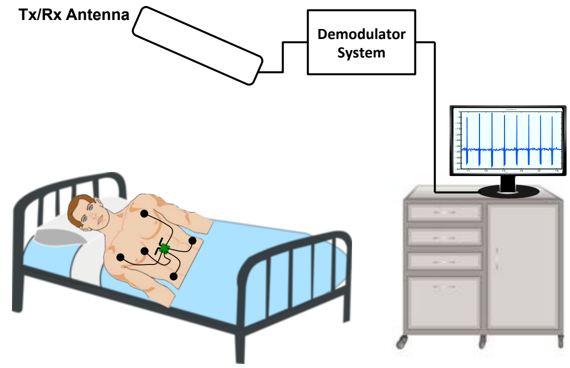


Fig. 14: Proposed application of passive wireless ECG system

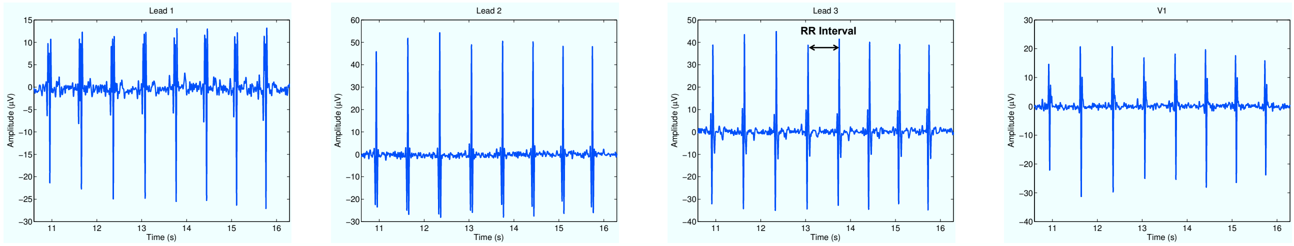
VII. Conclusion and Future Work

In this work, we have presented a passive wireless multi-channel telemetry device capable of transmitting an ECG to an external system. The telemetry device is battery-free, and supports up to 14 channels of simultaneous data transmission. The functionality of the passive telemetry device has been verified in both 2-lead and 5-lead ECG scenarios, and it is able to successfully record typical ECG waveforms simultaneously, as well as accurately track a subject's heart rate. The telemetry chip has a small footprint, 2.36 mm x 1.88 mm x 250 μm, only consumes 1.23 mW of power, and is currently capable of operating at a distance of 1 meter.

In its current form, the telemetry device is comprised of a rigid PC board that is approximately 3 x 4 cm. It is wearable in the 3-lead ECG configuration. A miniaturized antenna that can be worn on or close to a patient's skin could be integrated, allowing for improved read range and eliminating the need for the foam spacer. Additionally, the telemetry device can be integrated with flexible circuit packing, which would reduce the overall size and cost of the sensor, allowing for a disposable device. The results presented here were taken in a lab setting, far different from the typical hospital environment, the device is not yet suitable for clinical use. Further validation of the system in an actual healthcare environment will be needed. Consideration of potential interference with other medical electronics would be required, including a detailed electromagnetic compatibility assessment to ensure that the forward-link transmission from the base station does not interfere with other devices. Assessment of patient RF safety [11] would also be required.

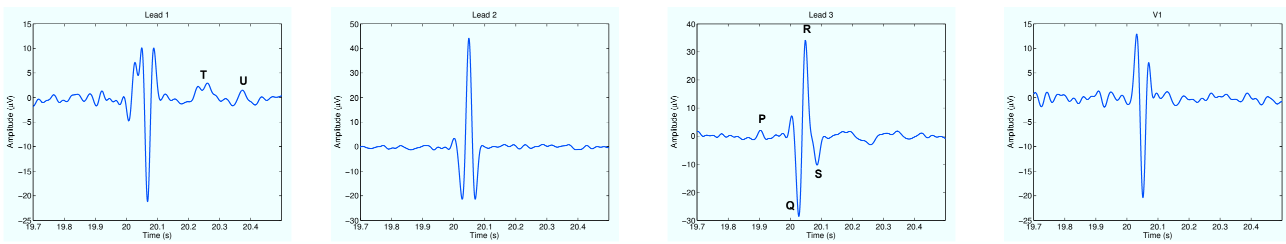
TABLE II: Multichannel Passive Wireless ECG Trace Characteristics

Trace	Peak **R**-wave Voltage (μV)	Average **RR**-interval (s)	Equivalent Heart Rate (bpm)
Lead 1	15.2	0.6674	89.90
Lead 2	56.6	0.6676	89.87
Lead 3	48.0	0.6676	89.87
V1	22.5	0.6676	89.87



(a) Lead I (b) Lead II (c) Lead III (d) V1

Fig. 12: Five lead fully passive multichannel ECG recording showing the three standard limb leads (Lead I, Lead II, and Lead III) and one precordial lead (V1).



(a) Lead I (b) Lead II (c) Lead III (d) V1

Fig. 13: Closeup of five lead fully passive multichannel ECG recording. Recordings were made simultaneously and processed offline.

References

[1] *Cardiac Monitoring Basics for the Adult Hospitalized Patient*. Covidien Inc., 15 Hampshire Street, Mansfield, MA, 2010.

[2] R. Fensli, E. Gunnarson, and O. Hejlesen, "A wireless ECG system for continuous event recording and communication to a clinical alarm station," in *IEMBS '04, 26th Annual International Conference of the IEEE Engineering in Medicine and Biology Society, 2004.*, vol. 1, September 2004, pp. 2208–2211.

[3] R. Fensli, E. Gunnarson, and T. Gundersen, "A wearable ECG-recording system for continuous arrhythmia monitoring in a wireless tele-home-care situation," in *Proceedings, 18th IEEE Symposium on Computer-Based Medical Systems, 2005.*, June 2005, pp. 407–412.

[4] T. Handa, S. Shoji, S. Ike, S. Takeda, and T. Sekiguchi, "A very low-power consumption wireless ECG monitoring system using body as a signal transmission medium," in *1997 International Conference on Solid State Sensors and Actuators. TRANSDUCERS '97 Chicago, 1997*, vol. 2, June 1997, pp. 1003–1006.

[5] C. Park, P. Chou, Y. Bai, R. Matthews, and A. Hibbs, "An ultra-wearable, wireless, low power ECG monitoring system," in *IEEE Biomedical Circuits and Systems Conference, 2006. BioCAS 2006.*, December 2006, pp. 241–244.

[6] D. Yeager, F. Zhang, A. Zarrasvand, N. George, T. Daniel, and B. Otis, "A 9 μ a, addressable Gen 2 sensor tag for biosignal acquisition," *IEEE Journal of Solid-State Circuits*, vol. 45, no. 10, pp. 2198–2209, October 2010.

[7] S. Mandal, L. Turicchia, and R. Sarpeshkar, "A low-power, battery-free tag for body sensor networks," *IEEE Pervasive Computing*, vol. 9, no. 1, pp. 71–77, Jan.-March 2012.

[8] F. N. Wilson, F. D. Johnston, F. F. Rosenbaum, H. Erlanger, C. E. Kossmann, H. Hecht, N. Cotrim, R. Menezes de Oliveira, R. Scarsi, and P. S. Barker, "The precordial electrocardiogram," *American Heart Journal*, vol. 27, no. 1, pp. 19–85, 1944.

[9] "Health informatics–personal health device communication part 10406: Device specialization–basic electrocardiograph (ECG) (1- to 3-lead ECG)," *IEEE Std 11073-10406-2011*, pp. 1–73, 30 2011.

[10] S. Thomas, R. R. Harrison, A. Leonardo, and M. S. Reynolds, "A battery-free multichannel digital neural/EMG telemetry system for flying insects," *IEEE Transactions on Biomedical Circuits and Systems*, vol. 6, no. 5, pp. 424–436, October 2012.

[11] Federal Communications Commission, "OET bulletin 65: Evaluating compliance with FCC guidelines for human exposure to radiofrequency electromagnetic fields," August 1997.