Passive RFID Tag based Heart Rate Monitoring from an ECG Signal*

Shrenik Vora, Kapil Dandekar and Timothy Kurzweg

Abstract— **In this paper, we propose a monitoring system that employs a passive RFID tag to transmit heart rate using an ECG signal as its source. This system operates without a battery and has been constructed with easily available commercial components. Here, an RFID tag is used as an on-off keying device, wherein it is normally transmitting, but turns off every time a heart beat is detected. Heart beats ranging from 30BPM through 300BPM are successfully measured using our device. It is shown that the system is capable of providing accurate heart rate measurements up to a distance of ten feet with a standard deviation of less than one beat per minute without a local power source. The proposed system is also found to be resilient in the presence of an additional RFID tag.**

I. INTRODUCTION

Over the last decade, there has been a big push towards the development of wearable health monitoring devices [1]. The underlying belief is that wearable devices will allow for affordable round-the-clock monitoring which will enable early detection and prevention of many diseases. Heart rate monitors and electrocardiogram (ECG) devices are a group of devices that have received wide attention from the wearable devices community. However, research on ECG monitors has been focused on making better wearable sensors and integrating them with a minimal profile. To wirelessly transmit the cardiac information, researchers have largely continued using battery-powered methods like bluetooth[2] and cellular technologies[3] or prototype RF transmitters[4]. Batteries add to the size and weight of wearable systems, making them obstrusive and cumbersome. A wearable system benefits from being battery-free as the absence of batteries reduces the system down time for battery replacement and charging.

Passive RFID (Radio Frequency Identification) tags work on wireless power harvested from a RFID reader and thus do not require batteries for operation. RFID tags have been conventionally used as product identification tools similar to barcodes. For these tags to be used in sensor networks, they have to be capable of transmitting sensor data alongwith the tag ID. In its simplest form, an RFID tag can be used as a one bit transmission device by turning the RFID tag on/off and having an RFID reader detect the tag's state. For instance, the reader detects a '1' when the tag is on and a '0' when the tag is off. A similar device that uses two RFID tags is proposed in [5]. Contemporary RFID tags are capable of transmitting multiple bits of optional data alongwith the default tag ID. However, additional circuitry like analog to digital converters (ADCs) and microcontrollers are required to digitize and embed the sensor data with the tag ID. These additional components not only increase the size of the system but also add to its power requirements. Another drawback of such systems is that they require significant transmitted data redundancy to achieve a degree of reliability. For example, the electroencephalogram (EEG) system proposed in [6] requires 92% data overhead and has a range of only 0.8m. The requirement for data overhead adds to the system power consumption. If this power is harvested wirelessly, higher power demand significantly degrades the system's range. Our goal is to design a RFID based heart monitor by leveraging battery-free transmission capabilities of RFID tags while using minimal additional components and not requiring any data overheads. Such a monitor can be used to wirelessly track a person's heart rate and location while he is mobile and could potentially limit the need for a patient to be tethered to a cardiac monitor on the hospital bed.

In this paper, we lay the foundations for an unobtrusive, battery-free, wearable system for monitoring heart rate from an ECG signal using passive RFID tags. Section II explains the system's basic principle of operation, Section III describes the system components and Section IV discusses our results.

II. PASSIVE RFID TAG BASED HEART RATE MONITORING

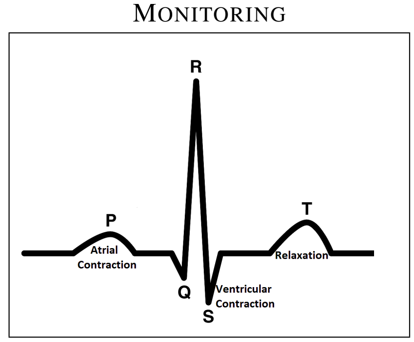

Fig. 1. PQRST complex in an ECG Signal

A typical ECG signal (Fig. 1) has sub-waves that represent parts of the cardiac cycle like atrial contraction, ventricular contraction and relaxation. As the 'R' wave is the most prominent of these waves, the heart rate can be easily measured by calculating the time between successive 'R' waves [7].

We propose a system that employs on-off keying (OOK) [8] to RFID communications to transmit heart rate. The RFID tag can be turned off for a specified duration every time a 'R'

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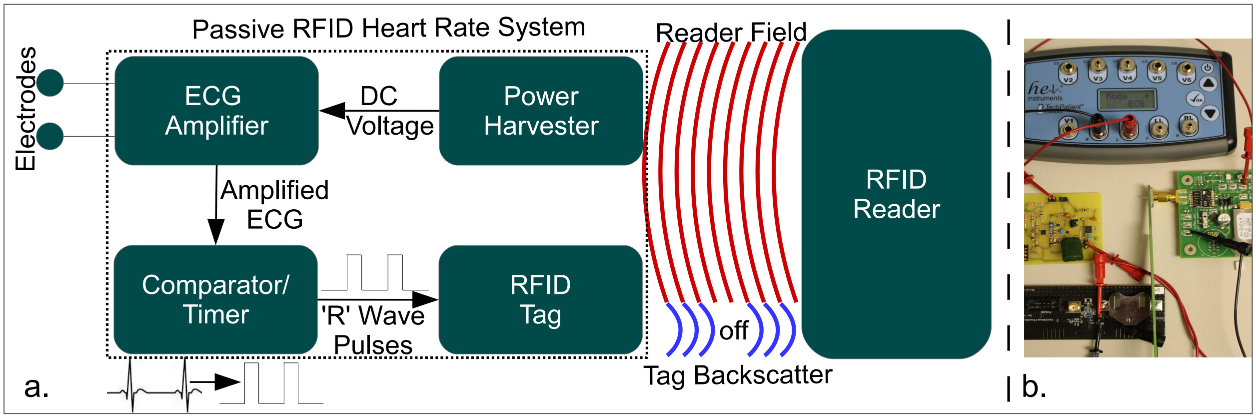

Fig. 2. a) Block Diagram of the Passive RFID Tag based Heart Rate Monitoring System and b) Picture of the system (Clockwise from top): ECG Simulator, Power Harvester, RFID Tag and ECG Amplifier with Timer

wave is detected. This action of turning the tag off results in a succession of RFID outages corresponding to the observed ECG pulses. These outages can be detected by the RFID reader and used to calculate the heart rate. Equation 1 gives the heart rate in beats per minute (BPM) where '$T_{R\text{-}R}$' is the time between successive RFID outages in seconds.

$$HeartRate = \frac{60}{T_{R-R}} \tag{1}$$

III. System Description

A. Components and Setup

The components of our system are shown in Fig. 2a. The ECG signal was generated using the HE Instruments' Tech-Patient Cardio ECG simulator. The amplitude of the ECG signal is typically less than a couple of milivolts and thus needs to be amplified to a level that can be used for analysis. However, such amplification requires power and we get that in our system by using a wireless power harvester. Since we want the harvester to work off the energy transmitted by the RFID reader, we chose a device (Powercast P2110) that harvests energy from the same frequency as our RFID tag (about 920 MHz). A low power amplifier design is imperative as the system is expected to run on harvested power. We chose a simple two lead ECG amplifier circuit based on the design provided in [9] and optimized it to ensure full scale range, lower power and smaller footprint. The amplified ECG signal is fed to the comparator/timer block which consists of a micropower 555 timer IC that detects the 'R' waves in the ECG signal. The timer's comparator is set to look for a signal that exceeds a set threshold such that the 'R' wave is correctly detected. Once the 'R' wave is detected, the timer sets its output high for a period of 100ms. This duration is user controlled and limits the maximum detected heart rate to above 500BPM. After the 100ms duration, the comparator output again becomes low and it starts looking for the next 'R' wave.

The RFID tag needs to turn on and off based on the signal from the timer. The Impinj Monza X-2k UHF RFID chip has a DCI input which suppresses all RF communication from the tag when that input is set to high. This tag allowed us to switch the tag on and off without using any external circuitry. We used Impinj's Speedway RFID Reader (IPJ-R1000) with a Laird Technologies S9028PCLJ antenna to read the UHF (Ultra High Frequency) RFID tag. Our system is designed to operate with an input voltage between 1.5V and 3V with a peak current consumption of less than 80μA. A partial picture of our assembled system without the RFID reader is presented in Fig. 2b.

B. Measurement Setup

We used the ECG simulator to generate five discrete heart rates between 30BPM and 300BPM. Each measurement was taken for a period of three minutes. The RFID reader and the system were kept approximately three feet away from each other. The reader was used to record the times at which a response from the tag was received. These time-stamped data points were then analyzed to determine the heart rate. We analyzed the range of the system by increasing the distance between the RFID reader and the rest of the assembly. We also studied the performance of the system in the presence of an additional dummy tag.

IV. Results and Analysis

A. Heart Rate Calculation

The amplified ECG signal for a 30BPM wave is shown at the top of Fig. 3. For a 30BPM signal, the ECG pulse appears every two seconds. The amplified 'R' wave triggers the comparator in the timer causing it to turn the output high for 100ms. This detection of an ECG pulse is demonstrated in the middle plot of Fig. 3. Every pulse in the middle plot turns the tag off for 100ms. We calculated the time difference between successive tag reads and plotted them at the bottom of Fig. 3. It can be clearly seen that there is a spike for every ECG pulse which denotes the duration of each tag outage. Similar results were observed for other heart rates as well.

Data for all heart rates measured between 30BPM and 300BPM is summarized in Table I. The second and the third columns in this table list the total time for each measurement and the number of heart beats detected by the RFID system in that time. The fourth column is populated

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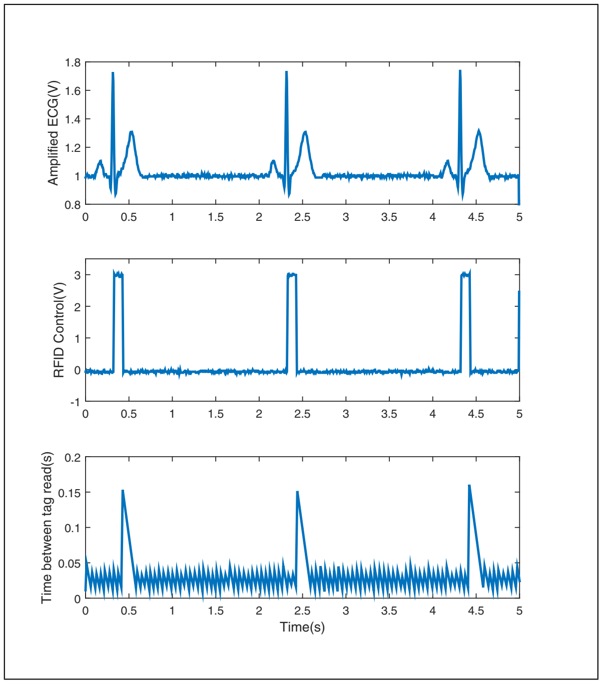

Fig. 3. Heart beat detection for a 30BPM ECG signal: amplified ECG signal(top), timer output to turn RFID tag on/off(middle), time between successive tag reads (bottom)

TABLE I

ANALYSIS OF TAG READ MEASUREMENTS AT SEVERAL HEART RATES

HR(BPM)	Time(min)	Beats	Average HR (BPM)	Beat-to-beat std dev
30	2.99	90	30.00	0.20
60	3.00	180	59.95	0.82
120	3.00	360	119.99	3.67
180	2.99	540	180.04	6.90
240	2.99	720	240.02	15.86
300	2.99	899	299.92	25.65

by dividing the number of beats in the third column by time in the second column. The beat-to-beat heart rate is determined by measuring the time between successive tag outages and using that in Equation 1. The last column lists the standard deviation of all such beat-to-beat heart rate measurement in the given data set. Heart rates are always presented as an integer, but we have included fractional values to compare accuracy. The average heart rate calculated for all measurements are accurate when sampled over three minutes. For 30BPM and 60BPM, the standard deviation of the beat-to-beat measurement is less than 1, which means that for these heart rates one could simply measure the time difference between successive tag outages and expect to get a very accurate heart beat calculation. However, the standard deviation starts increasing steadily after that. It is important to note that the accuracy of the tag timing measurements does not get worse with increasing heart rate but the same error in measurement has a larger significance for a higher heart rate. For example, a 20ms overestimation in a 30BPM measurement will cause an error of 2% while a 20ms overestimation in a 300BPM measurement will cause the calculated heart rate to be lower by 9%.

The primary source of this error can be explained using the bottom plot in Fig. 3. We expected that while the tag is on, the time between successive tag reads would be constant. However, that is not the case, and it alternates between approximately 0.015s and 0.03s. Another issue is that the time taken to turn the tag on/off could be anywhere in between 2ms and 20ms which causes additional variation. Besides, there could be instances when tag reads are missed which lead to a longer than expected measured time between successive tag reads. These factors create variability in the time measured between successive ECG pulses by the system. Thus, the heart rate determined by simply measuring the time between two successive heart beats as per Equation 1 may be slightly off. Inspite of these inaccuracies created by the tag read time variability, the calculated heart rates over a longer period are very accurate. For instance, the 300 BPM signal in Table I had nearly 900 beats detected in three minutes even though the standard deviation was significant. Thus, averaging is necessary for faster heart rates. However, the three minute sampling window used in Table I is both impractical and excessive. Hence, the next logical step is the determination of a minimum sampling time window within which we can expect to get a fairly accurate heart rate calculation.

B. Sampling Window

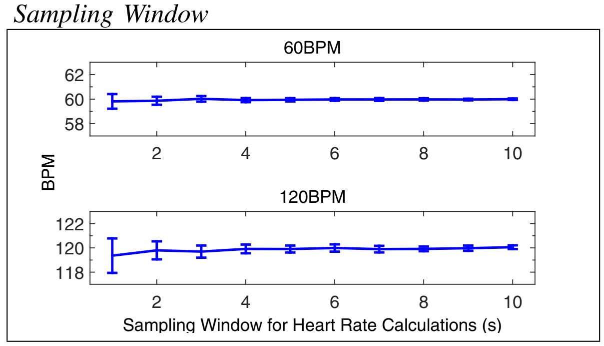

Fig. 4. Mean and Standard deviation (error bars) in heart rate calculations over different sampling windows

We use a modified version of Equation 1 here. Each window starts and ends with a detection of a beat. In equation 2, T_{sample} is the duration of the sampling window in seconds and 'Beats' represents the number of detected heart beats.

$$HeartRate = \frac{60 \times Beats}{T_{sample}} \qquad (2)$$

In Fig. 4, we have plotted the mean and standard deviation (represented by error bars) of heart rate measurements over sampling windows between one and ten seconds for 60BPM and 120BPM signals. If we want to have a heart rate calculation accurate to within 1 BPM, a sampling window of one second is enough for the 60 BPM signal. However, to achieve the same accuracy for the 120BPM signals, we need a sampling time of about 2 seconds. The higher standard deviation with increasing heart rate can be attributed to the

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increasing significance of the same error margin as explained earlier. However, one should note that the standard deviation for 120 BPM is smaller than that observed in beat-to-beat calculations shown in Table I. This lower deviation is because even for a one second sample window the 120BPM signal has two beats, thus taking advantage of averaging over a larger data set.

C. Range

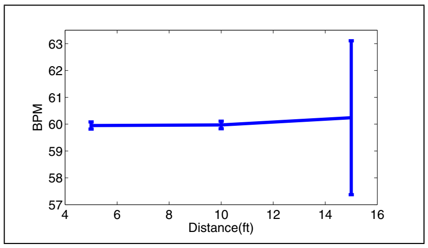

Fig. 5. Mean and Standard deviation (error bars) in heart rate calculations over increasing tag-reader separations

The range of the system is dependent on two factors; the ability of the harvester to supply enough power to run the circuitry and the reader being close enough to the tag to get sufficient data to resolve the heart rate.

To test the harvester range, we moved the reader away from the system while monitoring the output of the timer on an oscilloscope. We observed that the harvester is able to produce enough power to continuously operate the system circuitry upto a distance of 11 feet from the reader. Beyond that distance, the harvester runs the circuit for a few seconds at a time. This range can be enhanced by optimizing the harvester circuitry.

To quantify the range limitations due to the reader, we calculated the standard deviation in measurement while increasing the separation between the tag and the reader. The standard deviation in the calculated heart rate stayed within 1BPM of the actual heart rate for up to 10 feet and got close to 3BPM at 15 feet. One can increase the sampling window if higher accuracy is desired at longer distances.

D. Additional Tag

An additional RFID tag could interfere with reader measurements of our sensor tag. To analyze this interference, a dummy tag and the sensor tag were placed about one and a half feet from each other and the reader was kept at a distance of about three feet from them such that about 50% of the reader measurements were from the sensor tag. Table II shows 60BPM and 120BPM measurements for this setup. This table lists the average heart rate and standard deviation calculated over a four second sampling window and also the percentage of false beats detected in the entire measurement. It is evident that the performance suffers slightly due to the presence of an additional tag. As the reader spends time in reading the dummy tag, the time between measurements of the sensor tag increases thereby creating a larger variation in the time between successive beats detected. It is also possible that the reader may not read the sensor tag for a duration longer than 100ms in which case the system would falsely detect a heart beat. In spite of these errors, the system is able to calculate the heart rate very closely.

TABLE II

HEART RATE IN THE PRESENCE OF AN ADDITIONAL TAG

HR(BPM)	Average HR(BPM)	Std. Dev.	False Beats
60	60.37	0.92	0%
120	120.1	2.4	0.5%

V. CONCLUSIONS

In this paper we have demonstrated that an OOK modulated passive UHF RFID tag can be used to accurately determine the heart rate from an ECG signal. There is no additional data embedded on the RFID tag communications and the system can be easily integrated with existing RFID equipment. The design consumes less than 200μW of peak power and is suitable for battery-free operation using a wireless power harvester. Though the beat to beat variability is low for normal heart rates, averaging over short time windows of a few seconds may be essential for getting reliable results. We determined that the system maintains its performance till a distance of about ten feet from the reader and even in the presence of an additional tag.

In the future, we envision making a practical wearable device with a small form factor by integrating the components shown in Fig. 2b on a single flexible circuit board. We believe that the entire system with antennas can fit on a flexible board with a size no larger than a standard credit card. Such a system could be seamlessly integrated into garments without the user ever having to replace or recharge batteries or perform any maintenance. This device could be used as a mobile heart rate monitor for patients in healthcare facilities or a comfortable cardiac monitor for infants in cribs or even a wireless monitor for exercise equipment like treadmills.

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