On Implementing an Unconventional Infant Vital Signs Monitor with Passive RFID Tags

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***Abstract*—Enabling continuous and comfortable monitoring for newborn infants continues to remain a challenge. The circuitry and power required to monitor an infant's vital signs both with wired and wireless implementations are the main obstacles in realizing a seamless wearable infant monitor. In this paper, an unconventional battery-free and wireless infant heart and respiration rate monitor that uses passive RFID technology is proposed. Heart rate information is transmitted by turning an RFID tag on and off based on detection of heart beats and respiration rate is measured by monitoring the change in the received signal strength from the RFID tag due to respiratory activity. The system affords tremendous reduction in hardware and makes an integrated unobtrusive infant monitor possible. Such a monitor can contribute to reducing infant mortality by enabling continuous monitoring of potentially life threatening conditions like apnea and bradycardia.**

I. Introduction

Neonatal monitoring plays an important role in reducing infant mortality. Continuous monitoring of vital parameters allows for immediate intervention by enabling quick detection of dangerous situations like cessation of breathing, irregularities in heart beats or slowing of heart rate. It has also been shown that continuous cardiac monitoring can reduce mortality risks especially for low birth weight infants [1]. Such monitoring thus helps in improving survival rates and provides support for the infants' developmental growth [2]. Infants are typically monitored for conditions like apnea and bradycardia as these conditions can be symptoms for possible life-threatening issues. For infants, bradycardia happens if their heart rate goes below 80 BPM (beats per minute) while they are awake or 60 BPM while they are asleep [3]. Bradycardia can be the initial or the only symptom of a host of health conditions [3], thus, real-time detection of this condition is imperative for any infant monitor. An apnea for infants encompasses cessation of breathing for a period greater than 20 seconds or a lower duration when accompanied by bradycardia [4]. A reduction of respiratory motion by 95% over a period of 10 seconds is also considered an apnea [5].

Currently, infants are monitored with wired cardiac and respiration rate monitors. Such monitors are not only cumbersome for infants but also risky as the newborns can get entangled in wires [6]. Hence, a wireless and comfortable solution for continuously monitoring infant vital signs parameters is desirable. Additionally, if monitoring equipment is compact and affordable, an infant can be continuously monitored from home, which may also help in shortening their hospital stay.

Considering the clinical importance of continuous infant monitors and the necessity to make them less cumbersome, significant research efforts have been made in creating unconventional infant monitors. Several approaches for realizing wearable infant monitors are described in literature [7], [8]. However, these efforts mainly include adapting conventional methods of sensing and data transmission into wearable platforms. Resultingly, these wearable monitors remain cumbersome and uncomfortable. Novel methods of sensing have also been investigated, for example, several non-contact technologies that attempt to monitor vital signs without physically touching a subject have been explored by employing a video camera [9], [10], a laser vibrometer [11] or ultra wideband radio [12]. These technologies are dependent on expensive equipment and are not as reliable as contact sensing methods.

Passive RFID tags communicate data by backscattering signals in response to interrogation by the RFID reader and thus do not require any batteries for operation. Hence, by combining low power contact sensing methods and transmitting biosignal data with passive RFID tags, dependable continuous monitoring can be achieved without wires or batteries.

In this paper, an infant cardiac and respiration monitor is proposed that employs passive RFID tags in unconventional ways. The paper is organized as follows: Section II describes the proposed RFID based unconventional biosignal sensing methods for infants, Section III explains the setup for collecting experimental data, Section IV discusses detection of infant health conditions and Section V investigates sensor data fusion.

II. Infant Monitoring with Unconventional RFID Sensors

The Fig. 1 shows a high level diagram for the proposed RFID infant monitor that employs two RFID based sensors to independently monitor heart and respiration rates. RFID tags are used to transmit biosignal data unconventionally wherein sensor data is not directly embedded in the bits backscattered by the tag. The details of these unconventional biosignal sensors are described in this section.

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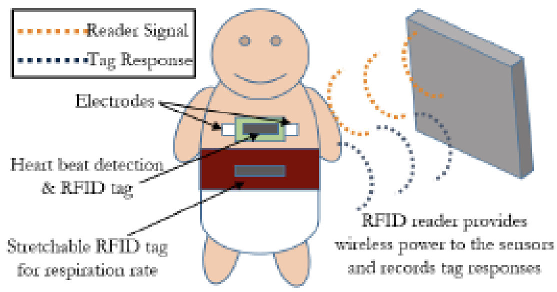

Fig. 1: RFID infant monitor with separate RFID sensors for heart and respiration rate monitoring

A. *Heart Rate Monitoring*

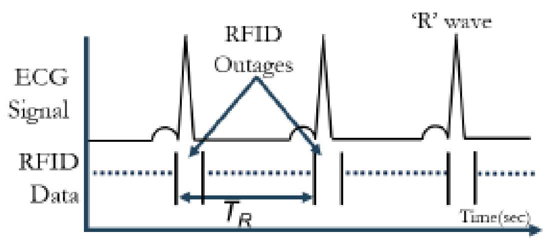

Fig. 2: Heart rate monitoring from RFID data by turning an RFID tag on and off

The heart rate part of the infant monitor is based on the system described in [13]. The principle of operation for this system is shown in Fig. 2. The system relies on an electrocardiogram (ECG) signal as the source which is picked up using contact electrodes. Every heart beat is indicated by a spike ('R' wave) in the ECG signal and the heart rate can be found by calculating the time between these spikes. In the absence of an ECG spike, the RFID tag attached to the system continues transmitting its unique tag number. However, when an ECG spike is detected, the tag is turned off momentarily thereby creating an RFID outage. Such outages are repeated every time a heart beat is detected. As the RFID responses are time stamped, heart rate can then be calculated by simply finding the time between RFID outages ('T_R'). Such a system has several benefits. There is no need to store data locally on the sensor which eliminates the need for analog to digital converters and memory devices. Transmitting heart beat data by turning the RFID tag on and off eliminates the need for any local power for data transmission. Hence, the only components that need power are an ECG amplifier and a simple heart rate detection circuit, both of which can be run using the energy harvested wirelessly from the RFID reader. Such a system not only makes the monitor very small but also eliminates the need for batteries. Proof of concept tests for such a device using commercially available equipment are described in [13].

B. *Respiration Rate Monitor*

The strength of the reader signal received by an RFID tag is a function of the tag's antenna characteristics among other parameters. If the RFID tag antenna changes shape due to respiratory activity, it would cause variation in the signal strength received by the tag. Hence, a flexible antenna that can change shape with respiratory activity is required. A fabric strain gauge that can be used in this fashion is realized by knitting conductive threads on a wearable garment [14]. When this garment (bellyband) is placed on the abdomen, it stretches as the abdominal wall expands with every breath. The signal received by the tag can be monitored using Received Signal Strength Indicator (RSSI) bits contained within the RFID tag communication packet. The operating principle of the respiratory monitor is shown in Fig. 3. The RSSI parameter changes with every breath and cyclical change in the RSSI value corresponds to the respiration rate. Again, an unconventional method is used to monitor respiratory activity using RFID tags without batteries in a comfortable form factor. Details about this sensor can be found in [14].

Fig. 3: Respiration rate monitoring from RFID data by correlating the change in received signal strength to antenna size modulation due to respiratory activity

C. *Integration for Infant Monitoring*

The basic systems described above have been tested independently to demonstrate their general viability [13], [14]. However, the systems need to be modified and adapted for an integrated infant monitoring application. It is clear that the heart rate monitoring method relies on time between tag reads. The respiration rate monitor also needs multiple reads within a respiration cycle to identify a change from the baseline RSSI. Having multiple tags in the environment not only increases time between tag reads but also adds irregularity in read times. In the case of the heart rate sensor, the presense of the respiration rate tag in the field increased the incidence of 'false outages' which are periods of heart rate tag silence in the absence of a heart beat. To address this, it was necessary to configure the RFID singulation protocol to interrogate consistently between the heart and respiratory tag to the extent possible. This was accomplished by modifying the reader settings in the RFID interrogation library. The final settings are described in Section III-B. Additionally, placing the two tags very close to each other caused highly irregular read rates due to the interference caused by the other tag's field. It was found that placing the two tags about 4 cm away from each other on the baby's body eliminated this problem. However, the placement issue motivated an alternate approach of using the same tag for both the measurements. A first attempt at unifying the two sensors into a single tag is described in Section V-B.

Additionally, a compact design that integrates all components of the heart rate sensor (power harvester, ECG amplifier and heart beat detection circuits) into a single board for wearable infant monitoring was designed and used for this application (Fig. 4d). An elaborate test platform to realistically simulate and test critical infant health conditions is also designed in this work as described in Section III.

III. Data Collection Setup and Tests

This section describes the setup and tests conducted to validate the RFID infant monitor.

A. SimBaby

The goal of this paper is to demonstrate the utility of the proposed RFID system in monitoring critical cardiorespiratory conditions in infants. Hence, a SimBaby [15], which is a programmable infant mannequin is employed. Using a connected computer, the mannequin's behavior and vital signs which include its ECG signal, heart rate, respiratory activity and a host of other parameters can be controlled. The ECG activity can be recorded by connecting electrodes to the right and left arm of the baby (shown in Fig. 4c). The SimBaby's abdomen rises and falls with every breath and hence its respiratory activity can be simulated by putting the RFID bellyband on its abdomen. An image of the SimBaby with the bellyband and the heart rate monitoring system is shown in Fig. 4b. The heart rate circuit is shown in Fig. 4d. The overall data collection setup with the RFID reader antenna, SimBaby and the SimBaby controller is shown in Fig. 4a. The RFID reader antenna is kept about 60 cm from the tags.

B. RFID Reader Settings

For an environment with multiple tags, RFID interrogation happens in a way that RFID tags respond successfully only once per interrogation 'round,' which then repeats after all tags have been observed. However, depending on the number of tags and interrogator settings, collisions may occur between tag responses causing variation in the tag read rate. A consistent read rate is, however, desirable for this application as both the heart and respiration rate applications rely on time between tag reads. In order to facilitate a consistent interrogation rate when interrogating multiple tags in the field, the RFID reader (Impinj R420 Speedway) was configured to use the "MaxMiller" configuration with "M4" encoding, which sacrifices interrogation rate for robustness to interference [16]. Because the number of tags to be interrogated is known for a given system in this application, the interrogator is also informed of an overestimate of the number of tags, which results in empty interrogation slots in which no responses are received in exchange for fewer unpredictable collisions among the tags. An overall interrogation rate of about 90 Hz is obtained using this configuration which corresponds to a tag read at approximately every 11 ms.

C. Data Processing

The heart rate tag is turned off for a period of 100 ms every time a heart beat is detected. Hence, at the RFID reader, a heart beat is simply identified by looking for an absence of response from the heart rate tag for a period greater than 100 ms. Heart rate is then calculated by finding the time between the start of successive outages. The processing for respiratory activity, however, is a little more involved.

To maintain compliance with Federal Communications Commission (FCC) regulations [17], the interrogator switches frequencies in the 902-928 MHz UHF RFID band for short periods of time (200 to 400 milliseconds per channel). This behavior is known as 'channel hopping.' These frequent changes in RFID interrogation frequency or 'channel' results in changes to RSSI in-band with the RSSI signal being used to determine respiratory activity. To mitigate these channel effects, the mean of the data points collected during a 200 millisecond period of interrogations of a particular channel ('channel burst') is computed, and each data point is replaced with the difference between the data point and the mean of that channel burst. Because the RFID interrogator does not guarantee a consistent interrogation period or interrogation rate, signal processing is performed on the data to separate the respiratory motion from noise artifacts in the data.

The above processing steps are conducted in real-time so that respiratory detection and rate estimation are facilitated without any delays. Because respiratory motion is oscillatory but possibly irregular in nature, the Short-Time Fourier Transform (STFT) is used to detect these respiratory oscillations in RSSI in successive small time windows of data. The magnitude of the power spectral density of the STFT is used to indicate the magnitude of any oscillatory pattern that exists in that window, which, in turn, detects a respiration [18]. The number of respiration cycles over an adaptive time period are averaged to estimate the respiratory rate per minute.

D. Infant Monitoring Tests

The SimBaby control software was employed to simulate bradycardia and apnea. To simulate bradycardia, the baby's heart rate was varied from 110 BPM to 75 BPM and then finally to 55 BPM. Both the actual heart rate and the RFID system calculated heart rate were recorded. The actual heart rate is measured by connecting a data acquisition module (NI myDAQ) to the ECG leads of the SimBaby and it was found that the actual rate varies slightly around the set heart rate. Based on these measurements, the RFID system error and correlation for heart rate measurement were calculated. For apnea, the baby's respiratory activity was stopped for 60 seconds periodically using the SimBaby program which caused the baby's abdominal movements to cease. These tests were first done independently. Both the heart rate and respiration rate monitoring systems were then placed on the SimBaby simultaneously and the above tests were repeated to check for any deterioration of collected data quality. Finally, data analysis was done to fuse data from both sensors and

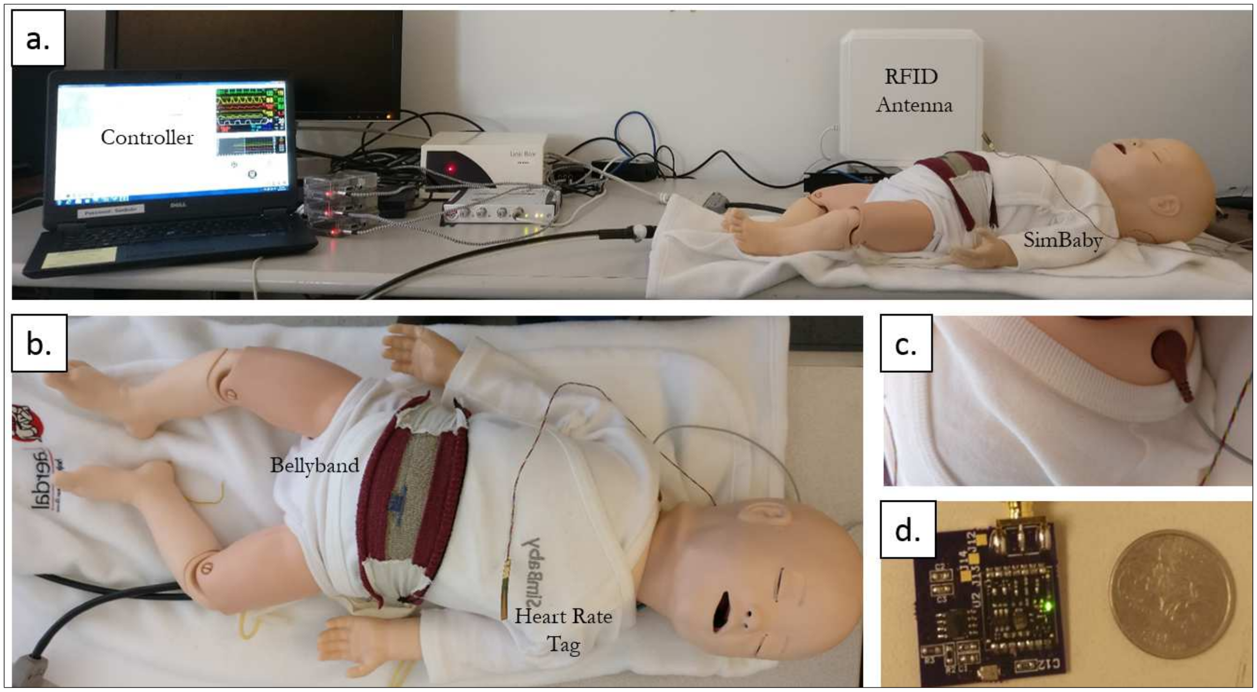

Fig. 4: Setup for data collection: a. Complete Setup; b. SimBaby with respiration rate (bellyband) and heart rate RFID tags; c. Contact for ECG signal; d. Heart rate circuit

simulate the possibility of using a single RFID tag to monitor both the heart and respiration rate.

IV. Infant Cardiorespiratory Conditions Detection

A. Bradycardia Detection

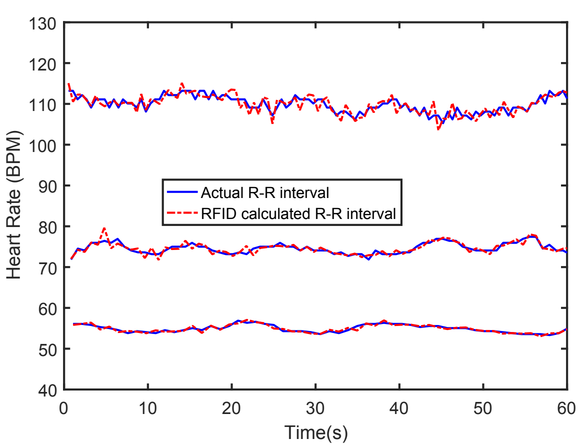

Fig. 5: Comparison of actual and RFID calculated heart rates for ECG signals sourced from the SimBaby

The first 60 seconds of actual and calculated heart rates for three different heart rate settings; 110 BPM, 75 BPM and 55 BPM are shown in Fig. 5. It should be noted that the last two settings can be classified as bradycardia for an infant while awake and asleep respectively. For all three heart rate settings, the RFID calculated heart rate closely follows the actual heart rate. Table I compares the mean and standard deviation of actual and RFID calculated 'R-R' intervals (interval between 'R' waves representing heart beats). The mean error and the standard deviation of error are also calculated. The mean error is of the order of hundreds of microseconds and the standard deviation of error is less than 10 ms for all heart rate settings.

TABLE I: R-R interval determination accuracy for three different heart rates

Parameter	Heart Rate Setting		
—	55 BPM	75 BPM	110 BPM
Mean R-R(s)	1.093	0.800	0.547
STD R-R(s)	0.017	0.014	0.009
RFID mean R-R(s)	1.093	0.800	0.547
RFID STD R-R (s)	0.018	0.016	0.012
Mean error (s)	-1.97×10^{-4}	-1.12×10^{-4}	-1.17×10^{-4}
STD Err (s)	0.008	0.009	0.009

The SimBaby was also programmed to gradually transition from 110 BPM to 75 BPM to 55 BPM over 20 second intervals in a separate test. The actual and RFID calculated heart rates for this test are shown in Fig. 6. The correlation between the two sets was calculated to be 0.9976.

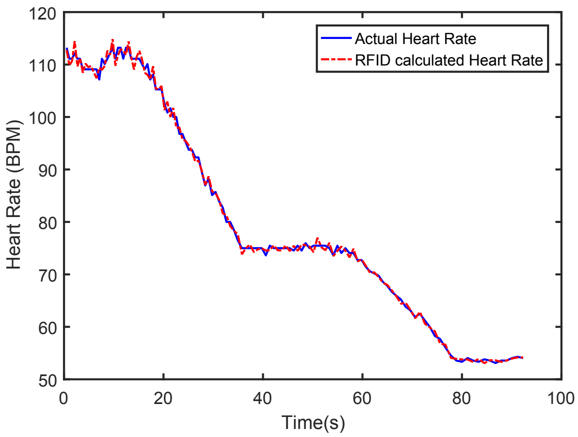

Fig. 6: Comparison of actual and RFID calculated heart rates for bradycardia simulated using SimBaby

B. Apnea Detection

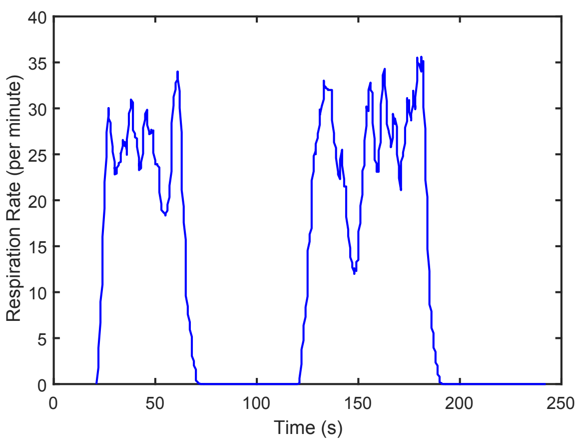

Fig. 7: A respiratory rate and apnea trial in which respiration activity is simulated using a SimBaby with the cycle set as breathing at 30 respirations per minute for one minute followed by no respiratory activity for one minute

Detection of apnea is shown in Fig. 7. The respiratory rate is estimated using RFID backscatter power as a signal over a 4-minute period. During this period, the SimBaby mannequin breathes at a rate of about 30 respirations per minute for one minute, followed by a cessation of respiratory activity for one minute, and repeats this process. The first 20 seconds of data is reserved for training the statistical test to classify respiratory activity; correspondingly, no rate estimate is given for this period. Reduction in respiratory motion is detected using the approach described in Section II-B and monitoring for non-oscillatory motion for desired consecutive time windows. The respiration rate was estimated in this test with a root-mean-square error of 8.8, and a cessation of respiration was noted within 10 seconds of each apnea episode by observing the respiration rate fall below 1.0 respiration per minute. It should be noted that alerting to a cessation of respiration is possible as early as 4 seconds by detecting consecutive 0.5 seconds windows of non-respiration during that time [18].

V. RFID based Infant Monitoring

A. Combining Data from Two Sensors

The comparison of the error in RFID calculated 'R-R' intervals and its correlation with the actual intervals with and without the respiration rate tag are shown in Table II. The performance does not deteriorate significantly after the respiration rate tag is added. The standard deviation of error increases from about 7 ms to 12 ms. This increase can be attributed to the time spent in reading the respiration rate tag which marginally increases the time between tag reads of the heart rate tag and causes some tag reads to be missed. The correlation with the actual heart rate continues to remain very high even after the addition of the respiration rate tag which had nearly 55 % of the tag reads over the measurement period.

TABLE II: Comparison of actual and RFID calculated 'R-R' intervals in the absence and presence of the respiration rate bellyband tag

Test	Mean Error (s)	Std. Dev. Error (s)	Corr. Coeff.
No Resp. Tag	-1.28×10^{-4}	0.007	0.9993
With Resp. Tag	-8.02×10^{-4}	0.012	0.9957

Detection of bradycardia and apnea using the RFID based heart and respiration rate monitors is shown in Fig. 8. In the SimBaby simulation scenario, the heart rate is held steady around 110 BPM for one minute and then bradycardia is simulated by dropping the heart rate to 60 BPM for the next minute. This cycle is repeated for a total period of five minutes. For respiration rate, the rate is held around 30 breaths per minute for a minute followed by the same period of no breathing activity. Periods of apnea and bradycardia are clearly visible in Fig. 8. The heart rate plot has a drop in heart rate at around 30 seconds due to a missed beat which may have been caused due to a sudden rise in baseline of the SimBaby ECG signal. Two spikes in heart rate can also be seen around 200 and 240 seconds. These were caused due to the the start of the SimBaby compressor which creates a spurious noise impulse that triggers the heart beat detection circuit thereby creating an outage. Respiration rate was estimated during this time with a root mean square error of between 10 and 12 breaths per minute across multiple data runs. More importantly, a reduction of respiration (apnea) was detected within 10 seconds.

The above tests were performed with the tags kept 60 cm away from the reader antenna. The system continues to have similar performance up to about 100 cm. However, beyond that range, the respiration rate monitor does not have enough resolution to detect the smaller changes in RSSI due to respiratory activity.

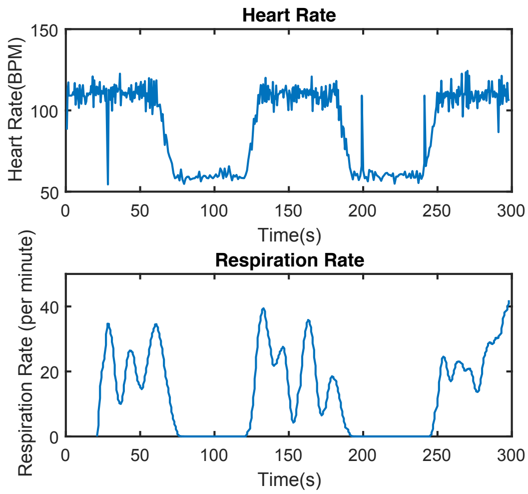

Fig. 8: Detection of bradycardia and apnea using RFID tag data

B. Integrated Single RFID Tag for Respiration and Heart Rate Monitoring

Both the heart rate monitor described in Section II-A and the respiratory rate monitor described in Section II-B utilize RFID technology; it is possible to construct a wearable garment that incorporates both sensors in a single RFID tag. In addition to eliminating interference between the two RFID tags during monitoring, a single tag approach reduces the size and cost of the deployed monitoring system. To do this, the respiratory RFID tag would be interrogated regularly as in Section II-B, but the tag response would be disabled on each heart beat to facilitate heart beat detection as in Section II-A. These breaks or 'outages' in RFID interrogation responses may occur up to 180 times per minute, and can last for about 100 ms. Hence, the respiration rate data may be adversely affected due to the integration with the heart rate tag. To determine the effectiveness of the respiratory rate estimation approach in the presence of RFID outages, first, RFID data was experimentally obtained for a heart beat setting of about 110 BPM. This data had outages associated with heart beat detection. Next, RFID respiratory data obtained for Fig. 7 was reused but data corresponding to the durations of outages from the heart rate tag was deleted. Thus a single data stream that contains outages due to heart rate and RSSI variations due to the respiratory activity is generated. The resulting data for respiration monitoring is shown in Fig. 9. The removal of outages modifies the training data set observed during the first 20 seconds. This removal results in small changes to the rate estimations throughout the data run as can be seen if Fig. 7 and Fig. 9 are compared. However, we continued to detect apnea conditions within 10 seconds; this performance was consistent across multiple data runs and heart rate 'outage' simulations.

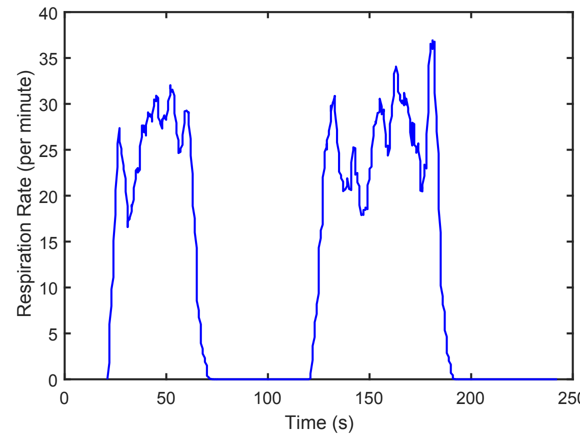

Fig. 9: Respiration rate trial with data points removed during 'outage' periods corresponding to heart beat detection

VI. Conclusions and Future Work

This work showed that the bradycardia and apnea detection are possible by simultaneously using novel RFID based heart and respiration rate monitors. The heart rate monitor displays a correlation of over 99% with the the actual heart rate even in the presence of the respiration rate tag. The respiration monitor is able to detect an apnea within 10 seconds of its onset. Additionally, it was shown that the two technologies can be combined into a single RFID tag to optimize the infant monitor.

One consideration with respect to the deployment of an RFID biological monitor is the minimum safe distance between the interrogator and the human subject. To maintain safe levels of RF exposure, a minimum distance of 50 cm from the human body is recommended [19]. Future efforts will focus on increasing the effective range of the RFID monitoring system to increase read resolution and processing accuracy while further reducing RF exposure levels.

One might correctly argue that the system may stop working if the baby were to turn over. However, such turning should be avoided for the baby's well-being and the resultant system outage can be used to trigger an alarm for intervention. In next steps, the first goal is to physically integrate the heart rate monitoring system hardware with the respiration rate system to create a single wearable infant monitor. Even though an episode of apnea is accurately detected using the system, further data analysis needs to be done to improve respiration rate calculation accuracy. Human tests need to be completed to validate the monitoring system. Additionally, practical aspects of system deployment need to be considered in future development. For example, the use case for home infant monitoring is demonstrated in this paper but multiple concurrent systems might be deployed in a nursery. Methods to mitigate the resultant tag read time irregularity need to be investigated. Other practical aspects like washing and spillage on the life of the fabric band also need to be studied.

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