Battery-Free RFID Heart Rate Monitoring System
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Abstract—The goal of this research was to design an RFID based wearable platform capable of continuous heart rate monitoring for infants. Two TechniTex P180+B conductive fabric electrodes with interconnects were integrated onto a baby onesie and connected to an RFID heart rate detection circuit that used an on-off keying modulation scheme to transmit heart rate data. The quality of the output signals obtained from the TechniTex P180+B electrodes had 98.90% correlation with that obtained from a standard MediTrace-230 foam electrode. An antenna connected to an RFID reader captured the RFID tag information and fed the data into a Raspberry Pi processor that was programmed to compute the heart rate. The real time heart rate was calculated by using a four-heartbeat moving average window with an acceptable error rate of 3 bpm. The calculated heart rate was then communicated to a mobile app via Bluetooth. The app displayed the heart rate as a discrete value and also as a real time graph. A local alarm system integrated with the MediTrace P180+B was used as an emergency backup system to alert in case the baby’s heart rate was not in the range of 80 bpm to 180 bpm and/or in case there was a communication failure between the RFID tag, RFID reader, processor or the mobile device.

I. INTRODUCTION
Heart rate is an accessible clinical variable that is used for evaluating and monitoring cardiac health. Heart rate has been shown to be a good indicator for determining cardiovascular mortality and in the general population, there is a correlation between resting heart rates and cardiovascular deaths [1]. An abnormal increase or decrease in the heart rate has been linked to cardiovascular diseases, including heart rate failure [2, 3]. Medical conditions such as tachycardia, characterized by an abnormal increase in heart rate, and bradycardia, characterized by an abnormal decrease in heart rate, are preliminarily diagnosed through heart rate monitoring [4]. The need for consistent heart rate monitoring of such cardiovascular diseases among at-risk populations, especially infants, is crucial [4].

When it comes to infants, studies have shown that continuous heart rate monitoring has led to an increase in survival rate especially in Neonatal Intensive Care Units (NICU) [5, 6]. By continuous monitoring of Electrocardiogram (ECG) signals, which provide reliable heart rate data [7], critical conditions such as bradycardia and tachycardia in infants can be detected [4]. This possibility of using a heart rate monitor in detecting these two medical conditions has prompted the need for developing wearable technologies capable of continuous heart rate monitoring of infants. Most of the current baby heart rate monitors are wired to heavy devices such as Holter monitor [8, 9] and, though they are accurate and efficient for long term continuous heart rate monitoring, they cause discomfort to the baby and have wires that can get tangled up easily [9]. The other baby heart rate monitors, that operate wirelessly using Bluetooth [10] and cellular technologies [11] or prototype Radio Frequency (RF) transmitters [12], are battery powered and therefore cumbersome to recharge. In addition, there is also the risk of failures in heart rate monitoring in case the batteries are not replaced promptly.

Most of these heart rate monitors also use standard Ag/AgCl wet electrodes that have been shown to have a few limitations such as not-being reusable, causing skin irritation, and supporting bacterial growth [9,13]. Also since the wet electrodes are gel based, they can dry up and cause abrasion that further lead to increase in electrode impedance, signal noise and motion artifacts [9, 13]. This has helped intensify the research into conductive textile electrodes and dry foam electrodes to be used for telemedicine sensors capable of monitoring the heart rate for long periods of time [13, 14].

Radio Frequency Identification (RFID) is a form of wireless communication that uses radio waves to identify and track tags attached to objects. The tags contain electronically stored information. Passive RFID tags work on wireless power harvested from an RFID reader and do not require batteries for operation. In this paper, an RFID based wearable platform capable of long term continuous heart rate monitoring is proposed. A heart rate detection circuit developed by Vora et al [15] that amplifies ECG signals and transmits heart rate data using an RFID tag was used and integrated. The proposed system consists of conductive fabric electrodes, passive RFID technology heart rate detection circuit, local data transmission and processing unit, and a front end mobile app as a user interface.
II. SYSTEM DESIGN AND ARCHITECTURE

The system design and implementation is illustrated in Fig. 1. The wearable component of the system consists of a onesie integrated with two electrodes and its interconnects for detecting the ECG pulse. A onesie is chosen because of its simplicity, design and mainly since it is the kind of clothing that is worn at most times by a baby. The connectors on the onesie are connected to an RFID heart rate detection circuit [15] which uses an on-off keying modulation scheme to transmit heart rate data. An antenna connected to the RFID reader captures the RFID tag information, and then feeds the data into a processing unit. A Raspberry Pi 2 Model B was chosen as the processing unit. The real time heart rate is calculated through developed algorithms programmed on the processing unit. The calculated heart rate is then communicated via Bluetooth to a mobile app which allows the user to interpret the heart rate data. A local alarm system integrated with the processing unit is used as an emergency backup system to alert people nearby in case the baby’s heart rate is not in the normal range of 80 beats per minute (bpm) to 180 bpm [4] or if there is a communication failure between any of the components of the system.

III. WEARABLE PLATFORM DESIGN

A. Electrode Characterization

Five conductive fabric electrodes (TechnikTex P180 + B, TechnikTex P130 + B, Nora Dell-CR, Bremen RS, Berlin-LX) obtained from V Technical Textiles Inc. were tested and their characteristics were compared with that of a standard MediTrace-230 foam electrode used as a reference. Each electrode was characterized on the basis of its impedance, wearable comfort and the effect on performance of the electrode after washing in a household detergent.

B. Electrode Impedance Measurements

An electrode is an electrical conductor that can be used to record the electrical activity of the heart over a period of time. Contact area affects the skin–electrode interface and its impedance which strongly affects the acquired ECG signal [16]. Hence it was important to measure the electrode contact impedance of each electrode where the dimension of each conductive fabric electrode was 4x3 cm², comparable to the size of the standard foam electrode.

Fig. 2 shows the block diagram of the setup for this test consisting of a base copper plate, skin dummy, the electrode to be characterized, a Velcro strap and a high-precision LCR meter (GW Instek LCR-819). The skin dummy was made of agar (a gelatinous substance derived from seaweed), distilled water, disinfecting agent and salt. To prepare the skin dummy, distilled water was mixed with the salt and disinfecting agent. 7g/100 ml agar-gar was mixed with the solution to obtain good stability and flexibility [13]. Since the properties of the dummy were very stable, the measurements could be repeated for several days [13]. A two-point measurement was performed to measure electrode impedance by using the LCR meter. The electrode impedance was measured in the frequency range of 12 Hz to 1 kHz [16]. This frequency range was selected because the biological ECG frequency range is from 0.5 Hz to 300 Hz [17]. The measured impedance result for each setup consisted of the net
impedance of the test electrode, the skin dummy and the copper plate. Since the impedance of the skin dummy and copper plate stayed constant for all measurements, any impedance change incurred thus depended only on the electrode’s contact impedance [13]. The lower the impedance the better the electrodes would acquire the ECG signals when placed on the human skin [16].

The impedance of the electrodes decreased as frequency increased. Fig. 3 shows the impedance results of two of the five conductive fabric electrodes and it is observed that the conductive fabrics had lower impedance than the standard foam electrode. This could be explained by the fact that conductive fabrics do not need adhesive conductive gel and are conductive across their whole area unlike the foam electrode. TechnikTex P180+B had lower impedance compared to the other four conductive fabric electrodes. TechnikTex P180+B is a silver plated knitted fabric with its raw material being 94% Nylon + 6% Elastomer. The fabrication of the materials could account for the difference in the impedance of the electrodes.

C. Electrode Comfortability

Fifteen random college students were surveyed for this test to characterize the five conductive fabric electrodes on the basis of their wearable comfort. The participants were asked to place each of the conductive fabrics on their hand and rank the fabrics based on the comfort they felt. TechnikTex P180+B was picked as the most comfortable electrode.

D. Electrode Washability

The washability test was performed on the electrodes to determine the effect on the performance of the electrodes after household washing. The electrodes were soaked in a household laundry detergent for two days and three hours as an extreme test. In each test, after five, ten, twenty-four and forty-eight hours, the electrodes were checked for any fading in the color of the fabric and if the texture felt different but none of that was observed. The electrodes were rinsed under running water and allowed to air dry before any tests were conducted. The impedance test on the electrodes was performed following the same setup and procedure as specified in III B.

The conductive fabric electrodes looked and felt the same after 2 days and 3 hours. The new washed electrode impedance data was compared with the dry unwashed electrode impedance data. Fig. 3 shows the impedance comparison between the unwashed and washed case for TechnikTex P180+B electrode and it was observed that the unwashed electrode had lower impedance than the washed electrode. Though the washed electrode had higher impedance, washing didn’t deteriorate the performance of the electrode (as will be discussed using Fig. 6).

E. Onesie Fabrication

The chosen TechnikTex P180+B electrode had low impedance when compared to the foam electrode and the other conductive fabrics, household washing did not destroy it and, above all, felt the most comfortable when placed on the skin. In the fabrication of the onesie, two Techniktex P180 + B electrodes were positioned to match the lower part of both sides of the musculus pectoralis major on the front bodice of the baby for best recording of the ECG pulse [18]. Since the electrodes were to be connected to the heart rate detection circuit, each electrode was cut out such that an electrode strip (interconnect) of 3.5 cm reached as far as the center of the chest of the baby. A gap of 1 cm was maintained between the two strips coming from each electrode thus maintaining the bipolar nature of the electrodes. A connector was soldered to 1cm2 of fabric electrode and cemented with epoxy onto the end of each electrode strip. Fig. 4(a) shows the cutout of the electrode and the soldered connector cemented with epoxy. Fig. 4(b) shows the electrodes sewn to the interior of a baby onesie. A small amount of padding, as seen in Fig. 4(a), was added between the electrode and the interior of the onesie to the region around the chest to provide for a better electrode contact area and also to provide comfort. As seen in Fig. 4(b), a non-conductive white cloth was sewn over the electrode strip extending to the center of the chest of the baby to ensure the ECG pulse was recorded only from the musculus pectoralis major and not from across the whole chest. The connectors, as seen in Fig. 4(c), were used to make connection with the heart rate detection circuit. To provide safety to the baby the connectors would be covered with caps when the heart rate isn’t being monitored.

Figure 4. Baby onesie with electrodes and connectors (a) individual components used (b) interior of onesie with electrodes (c) front view of onesie with electrodes and connectors sewn
F. Output Signal Quality

Fig. 5 shows the block diagram of the setup for this part. One skin dummy was placed in contact with each electrode on the interior of the baby onesie. One copper plate was placed on each skin dummy and connected to the HE Instruments’ Tech-Patient Cardio ECG simulator. A copper plate was used to provide adequate contact between the ECG simulator probe and the skin dummy. The connectors from the onesie were connected to the heart rate detection circuit which was then connected to an oscilloscope to obtain the display of the ECG pulse.

The test was also done with the standard MediTrace foam electrode to serve as a reference for comparison. In both cases, both signals had similar amplitudes (correlation of 98.90%) and the results are shown in Fig. 6. In the case of the washed TechniKTex P180+B electrode the correlation with the reference electrode was 97.98%. Thus the quality of the output ECG signal obtained through the TechniKTex P180+B electrodes integrated onto the onesie was of good quality and the data collected was accurate for further processing.

G. Heart Rate Detection Circuit

The heart rate detection circuit [15] would receive the ECG pulse signals from the onesie connectors connected to it. The circuit amplifies the signals and when it detects a heartbeat, it would turn off the RFID tag (shown as the gap between the two dashed lines in Fig. 7) that is connected to the circuit. And when the heartbeat disappears, it would keep the RFID tag on (shown as the stars in Fig. 7). The circuit uses an on-off keying (OOK) scheme to send the data to an antenna which is connected to the RFID reader. The heart rate detection circuit would be detachable such that it would be connected to the onesie connectors only when heart rate monitoring is needed.

IV. DATA PROCESSING AND COMMUNICATION

A. Data Acquisition

Fig. 8 shows the block diagram of the setup for data processing and communication. The S8658PLJ (LHCP) Indoor RFID antenna is connected to the IMPINJ Speedway RFID reader for data collection. In order to acquire data from the IMPINJ Speedway RFID reader, Low Level Reader Protocol Electronic Protocol (LLRP EPC) was implemented on the Raspberry Pi based on [19]. The data acquired included Tag ID, absolute time stamp, and most importantly, relative timestamp which was used to calculate the real time heart rate.

B. Heart Rate Calculation

Relative timestamp was utilized to calculate the real time heart rate based on two consecutive heartbeats using

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\text{Heart rate} = \frac{60}{T_2 - T_1}\]

as seen in Fig. 7. \(T_2\) refers to the timestamp of the current heartbeat and \(T_1\) refers to the timestamp of the previous heartbeat. The heart rate calculated from (1) is the real time heart rate at the moment that a heartbeat is detected as seen in Fig. 7. In order to get a stable heart rate, a moving average window was applied to each calculated heart rate from (1). This window was applied to minimize the error that came from ambient noise and in the case when there was a momentary loss in connection between the RFID tag and the RFID reader.

Figure 5. Block diagram of setup to test quality of output ECG pulse

Figure 6. ECG pulse signal obtained from TechniKTex P180+B sewn on onesie compared with standard foam electrode

Figure 7. Heart rate calculation procedure using relative timestamp

Figure 8. Block diagram of setup for data processing and communication
### C. Determination of Heartbeat Window Size

The size of the moving average window depended on how many heartbeats were included in the window. In order to determine the optimum window size, hundred heartbeat samples were collected at 60 bpm, 80 bpm, 120 bpm and 180 bpm each and the standard deviation in the case of each window size was obtained. The window size of 1, 2, 4, 6, 8 and 10 beats were tested at 60 bpm, 80 bpm, 120 bpm, 180 bpm respectively based on the fact that normal infant heart rate ranges from 80 bpm to 180 bpm [4]. The test result is seen in Fig. 9.

All moving average windows of different sizes outputted an accurate heart rate when the heart rate was relatively low; the standard deviation for all tested window sizes was within 3 bpm when the heart rate was below 100 bpm. A greater standard deviation was present when the heart rate increased. Increasing the window size lowered the standard deviation of the resulting heart rate. However, this would cause the initial waiting time to increase since enough heartbeats needed to be included in the window to perform the computation. A larger window size would also cause the real time variation in the heart rate value to be lost. For example, if the heart rate were to drop from 130 bpm to 70 bpm, it is considered as an abnormal condition. With a window size of 12 beats, it would take 11 heartbeats to detect the abnormality. However, if a window size of 4 beats was used it would take 4 beats to detect this abnormality.

The standard deviation dropped by a factor of 46.5% when the window size went up from 2 beats per window to 4 beats per window at 180 bpm. In addition, from 60 bpm to 180 bpm, the resulting standard deviation using a 4 beats moving average window was within 3 bpm, which brought the error due to the RFID data down to this acceptable level. Based on the results from Fig. 9, the moving average window of 4 heartbeats was used in the heart rate calculation algorithm with an acceptable error rate of 3 bpm.

### D. Local Processing Unit and Data Transmission via Bluetooth

Raspberry Pi 2 Model B was chosen to be used as the processing unit due to its compatible size and fast processing speed. The Raspberry Pi acquired data from the RFID reader, processed the encoded data and then calculated the real time heart rate. The heart rate was then transmitted to an Android mobile app. A Bluetooth dongle was attached to the Raspberry Pi in order to communicate with the mobile app via Bluetooth. A Bluetooth server was established on the Raspberry Pi using Radio Frequency Communication (RFCOMM) protocol, and would advertise its service using Service Discovery Protocol (SDP) [20]. The service name and Universally Unique Identifier (UUID) would be advertised in the process of establishing communication and the server would wait for the connection request from the Android app. Once the UUIDs returned from the mobile app matched with the server side, the connection would be accepted and the Bluetooth link would be established. After the pairing process is completed, the Raspberry Pi would constantly send the most recent heart rate to the Android app.

### E. Android App

A commercial mobile phone was used as the front end user interface since it provided for an easy and intuitive mobile user interface for parents/guardians and other users to be informed about the baby’s heart rate. The app would display the heart rate as a discrete value and also as a real time graph. The application interface, as shown in Fig. 10(a), incorporates features such as wirelessly receiving data from the processing unit, data visualization, alarm warning and emergency call.

In the Bluetooth pairing process, the mobile device played the role of a client. The Android app established Bluetooth RFCOMM channel and verified UUID with the Raspberry Pi. After data is received, the app would check if the delimiter character was present, and then delimit and

![Figure 9. Variation of calculated heart rate with heart beat window size](image-url)
graphically display the received heart rate. The app would ring an alarm when the heart rate detected was outside the normal range between 80 beats per minute (bpm) and 180 bpm [4] thus alerting the user if the baby had an abnormal heart rate. The user could manually turn the alarm off by the press of a button. An “Emergency” button was also included in the app. Once a user presses the “Emergency” button, the main screen would switch to the default dialing pad with “911” number dialed. Aside from the mobile device alarm, in case of wireless transmission failures between the Raspberry Pi and the mobile device, a local alarm system was set up on the processing unit. This local alarm system would send out an immediate audio warning to a user if there was an abnormal heart rate condition or signal loss. The alarm, as shown in Fig. 10(b), would stay ‘on’ until a button was pressed.

F. Signal Loss Identification

The Signal Loss Identification graph is shown in Fig. 11. In the ideal case, the processing unit should receive 30 readings from the RFID reader in 1 second. If there is loss in connection and the reader drops to 10 readings per second, the processing unit would receive a constant outage and no encoded heart rate information would be received. This is the unacceptable case. With some safety margin, the threshold of signal loss identification was set as 20 readings per second and a reading rate below this value would be identified as “lost connection” and the local alarm would be triggered.

V. CONCLUSION

In this paper, an RFID based wearable platform capable of long term continuous heart rate monitoring of an infant was proposed. The quality of the output signal obtained from the TechnikTex P180+B conductive fabric electrodes used had a 98.90% correlation with that of a standard foam electrode. Two comfortable and wearable TechnikTex P180+B conductive fabric electrodes were sewn onto a baby onesie to match the lower part of both sides of the musculus pectoralis major on the front bodice of the baby to record the ECG pulse data. The connectors from the onesie were connected to the heart rate detection circuit which used an on-off keying (OOK) scheme to send the data to the antenna that was connected to the RFID reader. The RFID reader captured this data and sent it to the Raspberry Pi processor that read every “off” as a heartbeat. The real time heart rate was calculated by using a four-heartbeat moving average window with an acceptable error rate of 3 bpm. The Raspberry Pi microprocessor communicated the calculated heart rate to an Android app via Bluetooth. The app displayed the heart rate as a discrete value and as a real time graph which allowed a user to read and interpret the infant’s heart rate. The app also had the functionality of calling 911 in the case of an emergency. A local alarm system was used as a backup system and was triggered on if the real time heart rate was not in the range of 80 bpm to 180 bpm and/or in case there was a communication failure between the RFID tag, RFID reader, processor or the mobile device.

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REFERENCES


