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# Energy Efficient Heart Rate Sensing using a Painted Electrode ECG Wearable

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**Abstract**—Many countries are facing burdens on their health care systems due to ageing populations. A promising strategy to address the problem is to allow selected people to remain in their homes and be monitored using recent advances in wearable devices, saving in-hospital resources. With respect to heart monitoring, wearable devices to date have principally used optical techniques by shining light through the skin. However, these techniques are severely hampered by motion artifacts and are limited to heart rate detection. Further, these optical devices consume a large amount of power in order to receive a sufficient signal, resulting in the need for frequent battery recharging. To address these shortcomings we present a new wrist ECG wearable that is similar to the clinical approach for heart monitoring. Our device weighs less than 30 g, and is ultra low power, extending the battery lifetime to over a month to make the device more appropriate for in-home health care applications. The device uses two electrodes activated by the user to measure the voltage across the wrists. The electrodes are made from a flexible ink and can be painted on to the device casing, making it adaptable for different shapes and users. In this paper we show how the ECG sensor can be integrated into an existing IoT wearable and compare the device's accuracy against other common commercial devices.

**Index Terms**—Body sensor networks, Low power sensors, Electrocardiography, Heart rate, Heart rate variability.

## I. INTRODUCTION

More than 28% of people in the UK will be over 60 by 2033 [1], with an associated increase in old age and degenerative disorders. Keeping those with long term conditions functioning in the community, out of hospital, is a strategic priority for the NHS and social care systems. Wearable devices for monitoring a range of physiological parameters are becoming widely known and available, and are having a transformative impact on personalised and preventative health and social care to enable this priority. To date by far the most successful wearables have been ‘*fitbit type*’ ones for activity monitoring, and significant research effort is being applied to enabling other sensing modalities in wearable form with sufficient accuracy, robustness, and ease of use for real-world application by non-specialist users.

For heart monitoring Photoplethysmography (PPG) is a well known noninvasive method based upon shining light into the body and measuring the amount of reflection, which varies

with blood flow. It is easy to perform at peripheral sites such as the wrist and the set up is also straight forward, and as a result PPG sensors are finding substantial new applications in wearable devices and current smartwatches for heart rate monitoring. However, raw PPG signals are severely corrupted by motion artifacts. These arise from a number of sources, principally a relative movement between the skin and the PPG light source/detector [2], and these obscure the heart related information. They necessitate complex signal processing to extract a reliable heart rate, for example [3], with debate present on the accuracy of current methods [4], [5].

In addition, the PPG light source intrinsically must consume a large amount of power, of the order of 1 mW, limiting the battery lifetime of a wearable device. Although minimum light-on duty cycles [6] and compressive sensing techniques [7] have been proposed to help overcome this, for smartwatches and similar wearable devices battery lifetime remains one of the primary concerns of end users [8]. The need for frequent battery recharging is a major obstacle to the wider take up and use of the technology, particularly by vulnerable users, and significant improvements are required for *green* Internet Of Things (IoT) devices.

For heart monitoring this can be achieved by using the Electrocardiogram (ECG) as the sensing basis. The ECG is the well known alternative method for monitoring the activity of the heart and is used widely clinically. It operates by placing small metal electrodes on the body to sense the micro-Volt sized electrical activity from the sino-atrial node and heart muscle contractions that cause the heart pumping action. Although lower power because the sensing element itself is just an unpowered metal electrode, wearable ECG monitoring is much more challenging than wearable PPG due to the electrode contact required with the body, the difficulty in maintaining this robustly over time, and the motion artifacts that are introduced.

Moreover, the signal is normally recorded from the chest with electrodes placed on either side of the heart. If electrodes are placed on just one side of the heart the collected signals reduce in amplitude and get increasingly small further away from the heart, Fig. 1. Placing electrodes on just one arm,

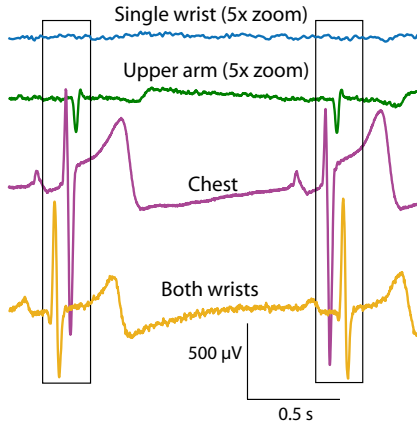


Fig. 1. The ECG signal showing each heart beat is large when the electrodes are on either side of the heart, and get smaller if the electrodes are placed only on one side of the heart. Note measurements are not simultaneous.

the time domain ECG signal reaches a 0 dB Signal-to-Noise Ratio at (approximately) the elbow, and so a single site *watch type* measurement of the ECG is not possible. It is possible, however, to place one electrode on one wrist, and then touch a second electrode with the other hand. As the two sensing connection points are on either side of the heart a high Signal-to-Noise Ratio ECG can be collected [9], see Fig. 1.

This paper presents an ultra low power two electrode wrist ECG wearable for collecting heart information in such a manner. The ECG front-end is combined with a previously reported IoT data collection node to automatically integrate the new sensing with established IoT and smart home demonstrators in the UK, and highly optimised circuit design is used to allow more than one month of battery lifetime. In addition, the metal ECG electrodes are made using a flexible Silver/Silver Chloride (Ag/AgCl) ink, which can be painted on to the wearable housing, allowing electrode sizes and shapes to be personalised to different users.

The remainder of this paper is organised as follows. Section II details our design in terms of the system architecture, hardware and software required. Section III reports the performance of the system, comparing it to currently available commercial devices for heart rate monitoring performance, and demonstrating the energy usage in detail. Finally conclusions are drawn in Section IV.

## II. SYSTEM DESIGN

Standard ECG recordings require three electrodes (2 for sensing a differential signal, 1 for rejecting common mode interference signals (50 Hz mains noise)) and a conductive gel between the electrode and the skin to minimise the impedance of the connection to the body. Both factors make conventional ECG approaches unsuitable for wrist based wearable monitors. We combine a number of techniques, described below, to overcome these issues. The final wearable ECG sensor node is shown in Fig. 2. It is made up of three consistent parts, each discussed in detail here.



Fig. 2. The wrist worn wearable sensor node. Top: Outer view showing a painted electrode which is in contact with the wrist when worn. A similar electrode is present on the other side of the case for touching with the other hand. Bottom: Inside view showing the SPHERE PCB board and ECG front-end amplifier circuit.

### A. ECG front-end

A number of techniques have been suggested previously for allowing ECG recordings using only two electrodes, simplifying the equipment set up by requiring just two body contacts. [10] used very high common mode rejection electronics, [11], [12] a DC servo loop for preventing saturation, [13] an active virtual ground, and [14] a common mode follower and a.c. bias approach. We make use of the topology introduced in [10] in 1980, as it requires only 3 active components, intrinsically minimising power consumption as our primary objective. This is achieved due to the very low number of active components required.

Our ECG amplifier circuit is shown in Fig. 3, which uses the LPV542 op-amp due to its ultra low  $1 \mu\text{W}$  power consumption per device. Two high input impedance buffers are used as the subject connection with partial positive feedback employed to increase the input impedance, making the circuit suitable for gel-free recordings (see Section II-C). The a.c. coupling provided by capacitors C1 and C2 allows the user to be d.c. biased via resistor network R1/R2/R3, ensuring the absolute input voltage remains within the input range of the amplifiers, without affecting the biasing of the subsequent circuit stages.

To reduce power no common mode feedback or feedforward is provided [15]. Instead, to ensure the collected ECG signal is within the input ranges of the front-end amplifiers the user is driven to a fixed mid-supply voltage, with resistors sized to ensure that the single point failure current into the user is limited to below  $18 \mu\text{A}$ , ensuring compliance with IEC 60601 safety standards. This arrangement means that more mains interference (50/60 Hz) is collected by the circuit, but as this is a known frequency which does not overlap with the

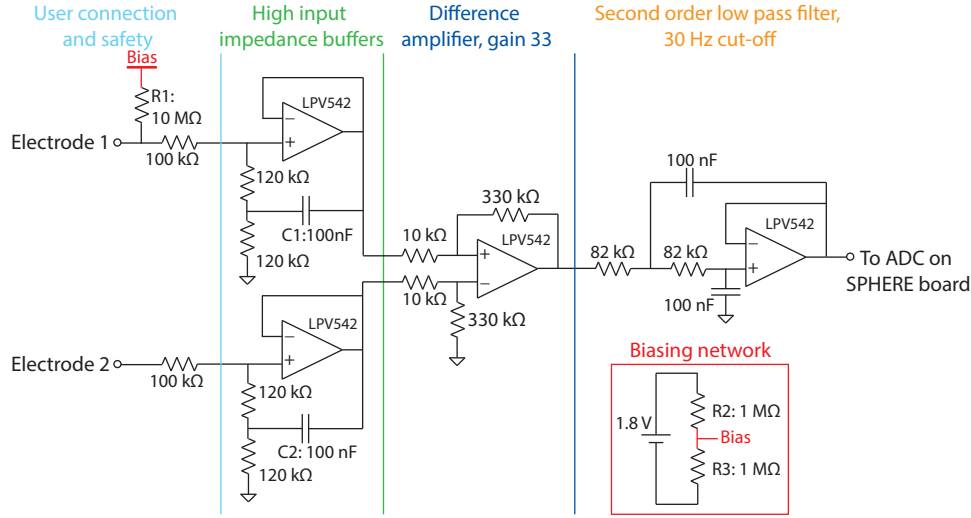


Fig. 3. Two electrode ECG front-end design to minimise power consumption.

wanted ECG components it is easily removed by hardware and software filtering.

Beyond the above, resistor sizes, particularly for the biasing, are chosen to minimise power consumption, at the cost of interference pick-up due to high impedance nodes being present, and introducing more thermal noise. All resistors are carefully sized to keep the final system noise to an acceptable level (see Section III). The front-end circuit is completed with a standard difference amplifier made with high precision (0.1%) resistors to ensure that a high common mode rejection ratio is maintained and a single op-amp second order low pass filter is used for anti-aliasing prior to digitisation.

The complete front-end circuit requires only four active components and approximately  $8 \mu\text{W}$  from a 1.8 V supply for ultra low power operation.

### B. IoT back-end

For digitisation and wireless transmission the new ECG front-end circuitry is connected to the latest version of the SPHERE wearable [16] which is an ultra low power body sensor node based upon the Texas Instruments CC2650 chip. The device incorporates wireless power transfer, two months of battery life, accelerometers for motion/gait analyses, and integrates with smart home infrastructure [17]. This paper adds the potential for heart sensing to this platform for the first time.

To minimise power we make use of the *Sensor Controller* on the CC2650 which allows continuous digitization without turning on the main processor core. ECG from the front-end circuit is sampled at 128 Hz and stored in a buffer up to 6 seconds in length, meaning that the main microcontroller has to wake up only once every 6 seconds in order to fetch the acquired ECG data. This leads to significant reduction in power consumption as usually the main microcontroller wakes up for every ADC sample (or in some cases every few samples if there is a FIFO buffer present). The estimated power consumption of the CC2650 system while running the

ADC sampling algorithm at 128 Hz is approximately  $200 \mu\text{W}$ . Running the same task without the sensor controller would easily exceed several milli-Watts in power consumption.

The SPHERE wearable provides the option of data collection in a *connected* or *not connected* state. In the *connected* state the raw ECG data is transmitted over a Bluetooth Low Energy (BLE) link, allowing all of the data to be saved and analysed offline or potentially in real-time as part of the smart home set up. A simple GUI has been created for this purpose, and the data can also be analysed in MATLAB or any similar software. In this *connected* mode the power consumption for the complete system is low enough to allow a month of data collection and streaming (see Section III-C).

For even longer term operation, in the *not connected* state the SPHERE CC2650 is programmed to run an adapted version of the Pan-Tompkins algorithm [18] for heart beat detection from the ECG. This allows the user's heart rate to be extracted on the wearable itself, with this data transmitted making use of the Bluetooth advertising packets as described in [19]. This allows the heart rate information to be wirelessly transmitted without having to power up and connect the full Bluetooth stack.

For the heart rate measurement algorithm to operate in real-time on the CC2650 the ECG signal is initially filtered using a low pass filter with a cut-off of 25 Hz and a baseline removal filter. A single point numerical derivative is then calculated and the output squared to amplify the QRS complex, the part of the ECG signal which has the highest rate of change and corresponds to each heart beat. This squared signal is integrated with a moving average of approximately 200 ms and a peak detection algorithm is used to identify each individual heart beat. The difference between the Pan-Tompkins algorithm and the algorithm implemented on the SPHERE wearable is that we do not store any of the results in memory and so have a fixed rather than two adaptive thresholds for heart beat detection. This has some disadvantages in terms of accuracy



Fig. 4. Typical ECG recording set up with two ECG electrodes used, one on the back side of the SPHERE wearable touching the wrist, and one on the front for touching with the other hand. A PPG device is worn on the user's other wrist to provide a reference heart rate measurement.

but it is saves great deal of processing time and memory.

### C. Case and electrodes

The electronics are housed in an off-the-shelf Minitect plastic housing [20] and held on to the wrist in the position of a standard smartwatch using an elasticated strap. The total assembly weighs 29 g, including the strap, for easy wearability. A typical recording set up is shown in Fig. 4 where the electrode (on the back face of the case) is in constant contact with the skin and when a heart rate recording is wanted the user touches the front facing electrode with a finger from the other hand. This provides two points of contact, on either side of the heart, allowing a high amplitude ECG trace to be recorded (as shown in Fig. 1).

New in this system, the body contact electrodes are made using a medical grade Ag/AgCl ink which is painted on to the casing. Previously we have demonstrated Silver (Ag) coatings painted on to plastics for the recording of bio-potentials without requiring a conductive gel to be present [21]. We now make use of a Silver/Silver Chloride paint available from Creative Materials [22], which is not only bio-compatible but marketed as medical grade, and allows superior sensing performance with lower half-cell potential, less long term drift, and reduced contact noise compared to using Silver [23].

This painted electrode approach allows flexibility for different sizes and shapes of electrodes to be used with different people, and different cases for the wearable sensor node to be used and tailored for individual preference. As just one example, Fig. 5 shows a customised electrode shape which can be used instead of a flat electrode painted on the case in order to get better penetration through hair on the wrist. Similar to electrodes for recording through hair on the head which have *fingers* [21], the electrode in Fig. 5 has small bumps to push aside the hair and make a better contact with the skin. It has not been investigated in this work, but this structure can be 3D printed as in [21], and the sizes and shapes of the bumps changed on a person-by-person basis to get the best trade-off between long term comfort, body contact quality, and sensing performance.

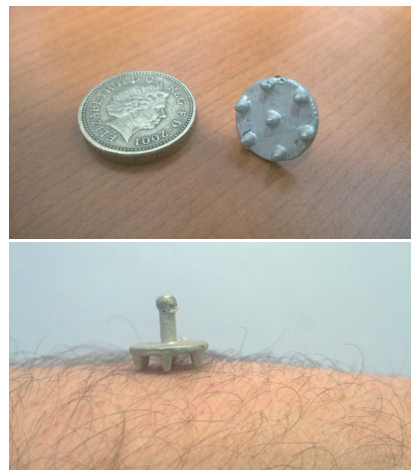


Fig. 5. Example Ag/AgCl back face electrode which can be used instead of the flat painted on electrode in Fig. 2 if better penetration through hair on the wrist is desired. Top: The electrode has small bumps to better pass through hair. Bottom: Placement on the wrist.

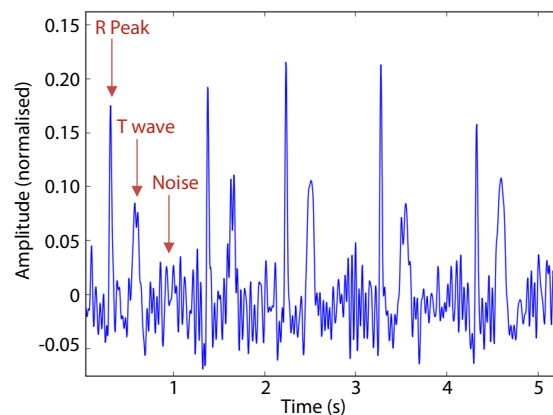


Fig. 6. Example ECG trace shows R peaks due to heart beats. These can be identified in the time domain allowing extraction of heart rate and heart rate variability measures using standard ECG processing.

## III. PERFORMANCE ANALYSIS

### A. Example ECG

An example of the raw ECG signal recorded using the new ultra low power sensor node is shown in Fig. 6, prior to software filtering. R peaks corresponding to each heart beat, and T waves, a morphological feature of the ECG, are clearly seen and marked on Fig. 6. The R peaks allow heart rate and heart rate variability analyses to be performed, extracting the wanted long term physiological measurements when the user has their hand on the front facing electrode.

In addition, in Fig. 6 a large amount of background noise is also seen. This is a deliberate design decision: the system is optimised to allow R peak identification for heart rate and heart rate variability measurements. It is not optimised for measuring ECG morphology components such as P and T waves, which even with low recording noise floors may not have the same shapes when recorded on the wrist as when (conventionally) recorded on the chest. This noise is easily

removed in the system back-end using standard ECG filtering approaches as described below.

### B. Heart rate measurements

For quantifying the performance a series of experiments were carried out with the new ECG sensor worn on one wrist, and a commercially available PPG device (Empatica E4 [24]) worn on the other wrist. Note that it is not possible to use a second simultaneous ECG recording as the gold standard comparison: as both ECG units drive the body (via a driven right leg circuit or fixed voltage in our design) the total body driving is different when both units are connected compared to using only one unit, resulting in neither device recording the same ECG signal as they would if only one unit was connected at a time.

A total of 12 five minute ECG recordings were carried out with 8 different participants, aged 22–33. Subjects were stationary at a desk while ECG data was streamed in the *connected* state. For analysis three recordings were subsequently discarded: two due to the PPG device recording a poor quality signal allowing no comparison signal to be extracted; and one due to the SPHERE board being incorrectly used such that only very small amplitude R peaks were recorded. To extract a heart rate from the SPHERE data the following ECG processing procedure was applied:

- 1) The raw ECG data was low pass, high pass and notch filtered using first order zero phase delay (`filtfilt`) Butterworth filters.
- 2) The ECG baseline wander was removed using the Discrete Wavelet Transform as described in [25].
- 3) Candidate initial R peak locations were extracted using the Pan-Tompkins algorithm [18]. Identified R peaks with amplitudes less than 1.5 times the RMS of the signal, or greater than 9 times the RMS, were discarded.
- 4) The ECG smoothed using an extended Kalman filter based around the initial R peak locations as described in [26].
- 5) Final R peak locations extracted by re-running the Pan-Tompkins algorithm on the cleaned ECG data. R peaks closer together than 0.4 s (150 beats per minute) were discarded.
- 6) A Kalman tracking filter with zero order hold state model implemented to track the heart rate in the presence of both missing R peaks and additional R peaks due to transient events.

From this set of R peak locations a single heart rate value was extracted from every 10 s window of data, with 8 s overlap between estimates. The Empatica E4 gives an estimated heart rate value every 2 s and a single mean value for each 10 s window was similarly calculated. Results are given in Table I which quantifies the mean and standard deviation of the heart rate estimation difference across all of the heart rate calculations in a record:

$$\text{Difference} = \text{abs}(\text{ECG HR} - \text{PPG HR}). \quad (1)$$

TABLE I  
DIFFERENCE BETWEEN THE NEW SPHERE ECG DEVICE AND A WRIST WORN PPG REFERENCE. ALL UNITS ARE BEATS PER MINUTE.

Record	Mean difference	Standard deviation
1	5.61	4.02
2	2.87	1.80
3	6.77	6.08
4	11.3	2.59
5	2.66	2.87
6	5.96	6.17
7	1.83	1.85
8	2.19	2.29
9	1.88	1.36
Mean	4.56	3.23

Table I shows that all of the extracted heart rates are within a few beats per minute of the comparison PPG recording, with an average difference of 4.56 beats per minute. Note that the PPG gold standard device itself is a wearable unit, subject to motion interference and similar, giving some uncertainty in the actual underlying heart rate. We have estimated the difference between the reported heart rate from the algorithmic output of the Empatica E4 and the actual number of peaks in the PPG trace to be a mean of 2.7 beats per minute across all records, with a standard deviation of 2.1 beats per minute. The new ECG sensor accuracy compares well with this.

When the SPHERE board misreports the heart rate it is generally due to the presence of additional R peaks from movements of the finger/wrist. These incorrect peaks can be discarded by eye, particularly after the Kalman filtering, but are still identified by the Pan-Tompkins in automated processing. The tracking filter removes many of the remaining erroneous peaks, but not all. We anticipate that by replacing the current Kalman tracking filter with a particle filter [27], and the Pan-Tompkins algorithm with a more robust alternative (many methods, such as [28], have been reported), that the heart rate extraction can be improved further.

### C. System energy usage

The energy usage of the system was determined by using the Texas Instruments INA226 Power Monitor. The system was powered using a 3.7 V battery with 100 mAh capacity and the monitor configured to take 16 bit current samples at approximately 22247 samples/second. The ECG system was set up to take 128 samples per second and transmit approximately 40 BLE messages per second in the *connected* state. Fig. 7 shows the results of the experiment for a typical 0.5 second period during transmission.

In Fig. 7 the average current is approximately 9 mA. Note that this is the energy usage of the entire system, including the ECG front-end and IoT back-end. In practical use we assume that a heart rate measurement will be performed once an hour each day, and otherwise the system will be placed in a low power idle state. In the idle state, the SPHERE wearable device has previously been shown to operate at 3.3  $\mu\text{A}$  [17]. Including the front-end idle current, the total idle current is 11.3  $\mu\text{A}$ .

When performing a heart rate reading it takes between 20 and 30 seconds for sufficient heart beats to be present to allow



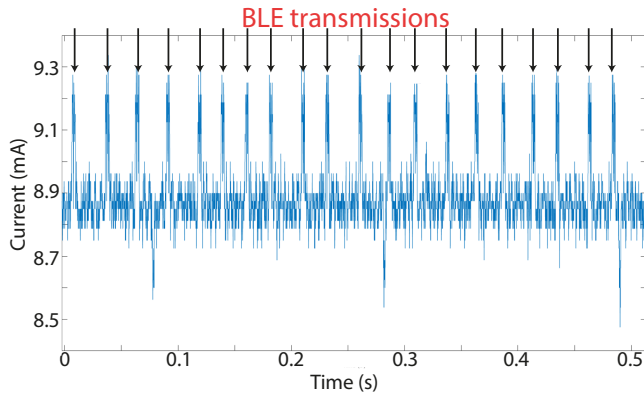


Fig. 7. Example energy usage during ECG collection in the *connected* state.

accurate estimation and the heart rate detection algorithm to stabilise and provide a measurement. The average current in one hour, where a 30 s ECG measurement is taken, is thus 86  $\mu\text{A}$  (30 seconds at 9 mA and 3570 seconds at 11.3  $\mu\text{A}$ ). Based on these parameters the system can last for an estimated 48 days (100 mAh / 0.086 mA), a step change in battery lifetime for a heart sensing wearable device weighing only 29 g including the battery and strap.

#### IV. CONCLUSIONS

In this paper we proposed a new wearable heart monitoring device with ultra low power consumption and customisable electrodes via the use of a Silver/Silver Chloride ink painted on to the device casing. The design employs few active components to minimise power and trades more system noise to further reduce power consumption while still producing an acceptable signal. The ECG front-end is integrated with an existing IoT back-end to be worn on the wrist and we compared the performance of the device against other comparable commercial devices, showing its accuracy to within a few beats per minute. Assuming moderate user activated sampling periods, the system can last for over a month using the supplied battery, and weighs less than 30 g including the battery and strap. In future work we will further use the painted ECG electrodes with the capacitive sensing inputs on the CC2650 chip to allow a button-free, customisable to the user, interface to the SPHERE board, and to allow the ECG circuit to automatically turn on when it detects that the user has touched the front facing electrode.

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