The Berkeley Tricorder: Ambulatory Health Monitoring

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Abstract

We developed and tested the Berkeley Tricorder, a health monitoring device capable of measuring a subject's ECG, EMG, Blood Oxygenation, Respiration (via Bioimpedance), and motion – almost equivalent to the feature set of a hospital bedside patient monitor. Our focus has been a highly integrated design incorporating the radio and all associated circuitry on a single PCB. The device stores data locally on microSD flash and/or transmits via Bluetooth. We will also discuss a strap we have developed which utilizes reusable electrodes for data acquisition as well as a desktop and mobile application for real-time data telemetry. We have evaluated the efficacy of the device in recording ambulatory data from 24 subjects and found the data acquisition relatively free of motion artifacts.

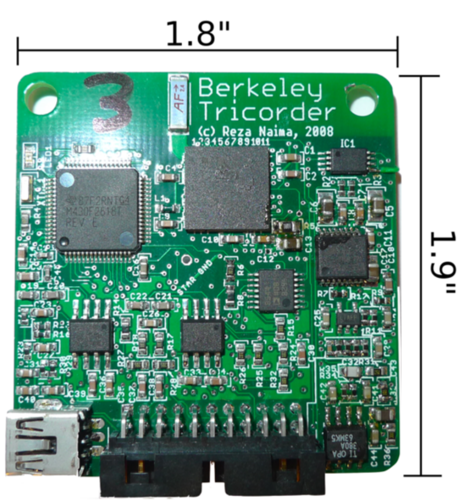

Figure 1. The Berkeley Tricorder Device

Figure 2. Subject Wearing the Monitor

1 Introduction

The advancement of precision micropower amplifiers, microcontrollers, and MEMs technologies have enabled the development of very small, low power, wireless health monitoring technologies. A number of groups have worked on this problem[5][12]. However, most devices developed only have one or two sensing modalities or require multiple networked devices attached to the subject. Our approach has been to develop a highly integrated ambulatory device which reduces cost and improves reliability by reducing the number of components.

Thousands of people die each year in developing countries due to lack of proper healthcare. This is especially true in remote rural areas where rapid transportation to a hospital is not available. But, cell phone coverage is prevalent, even in remote regions. By developing a robust, low cost device capable of remote telemetry through the cellular network, we hope to enhance telemedicine in remote locals. The feature set of our device is comparable to a hospital

Modality	Sampling Rate	Bits/Sample	Comment
ECG	256Hz	12	2-Stage HPF; RFI Filer; 2-Electrode & ground
EMG	256Hz	12	1-Stage HPF; RFI Filter
Respiration (Bioimpedance)	256Hz	12 Phase; 12 Mag.	4-Electrode configuration; 350μA current at 50kHz
Acceleration (3-Axis)	256Hz	8/Axis	
Blood Oxygenation	256Hz	15	Reflective forehead sensor; 2-Stage Amp with DC-Offset Subtraction

Table 1. Device Parameter Summary

patient monitor, lacking only blood pressure.

2 Design Goals

Our primary goal was to produce a highly integrated device which records multiple health parameters while minimizing size and cost. Our other requirements include:

- A means of data storage for at least 24 hours, and the ability to transfer the data from the device.
- A means of remote telemetry for real-time data viewing.
- A comfortable means of wearing the device, concealed under clothing.

We decided to incorporate five modalities (Table 1) based on feedback from medical practitioners.

2.1 Microprocessor

A microprocessor (MCU) is the core of most contemporary health monitoring devices. We required a MCU that provided us with multiple fast, high-precision ADC inputs for data acquisition as well as a DAC for our SpO_2 implementation. We needed multiple serial (UART, SPI, I^2C) interfaces, a DSP for digital filtering and low power consumption for extended battery life. We found the Texas Instruments MSP430 MCU a perfect fit for our needs. Our current design utilizes the MSP430F2618 which is fast (16Mhz), and has large RAM (8kb).

2.2 Data Storage

Our sensing modalities generate up to ~10kbps of data (Table 2), or 823 Megabytes of data in 24 hours. The availability of inexpensive microSD cards not only provide many gigabytes of data storage, but also allow for rapid data transfer to a personal computer by physically removing the card and connecting it to a USB reader. This process is facilitated by saving the data on a FAT filesystem, which has become the de facto standard for removable media.

[1]DC offset subtraction value are only transmitted when they change

Modality	Bytes/Second
ECG	768
EMG	768
Acceleration	2304
SpO_2	2304-4608[1]
Bioimpedance	1536
Timestamp	6
Sync-Block	3
Total	**7689-9993 Bytes/Second**

Table 2. Bandwidth Requirements

2.3 Bluetooth

Telemetry is an important requirement for a medical device deployed in a remote region as it provides a means for the diagnostic data to be viewed in real-time, or to be transmitted to a trained physician. We chose Bluetooth to satisfy our telemetry needs for a number of reasons.

- High level of penetration in consumer devices such as cell phones and laptops.
- Standardized profiles (Serial, Audio, Object Transfer, Dial-Up Networking, etc) requiring no additional development effort.
- High-speed wireless (v1.2 Bluetooth is rated at 1Mbit/s).
- Lower power consumption per bit than competing devices. (BlueCore3 Bluetooth: 62.5nC/bit @ 720kbps; XBee ZigBee: 180nC/bit @ 250kbps; Telos Mote: 78nC/bit @ 250kbps[6])
- Single IC implementation.
- Multiple built-in interfaces peripherals to help simplify device design such as a ADC/DAC/USB interface

Rather than using a monolithic serial to Bluetooth module, we have incorporated the CSR plug'n'go BlueCore3 chipset directly. Not only does this give us full access to many peripheral interfaces, but it also significantly reduces the device cost. The BlueCore3 IC provides audio input/output, a USB interface, 16 general-purpose I/O lines which can be used for I^2C or SPI communication, ADC, DAC and a serial interface. The firmware can be configured to provide many different profiles including Object Exchange (OBEX) to facilitate bulk data transfer, and more importantly, dial-up networking (DUN) to allow telemetry without the need for any custom phone software. Our current implementation utilizes Bluetooth as a serial device and interfaces with the MSP430 at up to 1Mbps.

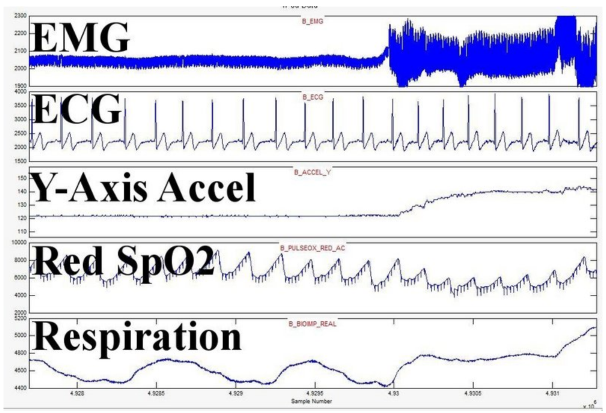

Figure 3. Sample data from device

2.4 Blood Oxygenation (SpO_2)

Blood oxygenation is a critical parameter which can help diagnose conditions of pulmonary distress, hypovolemia[8], vascular perfusion, among other. Blood oxygenation has a secondary benefit in that it can be used to compute pulse transit time (PTT), or the amount of time between the ejection of blood from the heart to the pulse arrival time at a peripheral location. It has been shown that there is a direct correlation between PTT and arterial blood pressure[2]. In order to derive blood oxygen saturation, determine pulse arrival time, and keep costs low, we needed to record the SpO_2 waveform rather than to use an OEM device which only computes the oxygen saturation percentage. Being able to visually inspect the SpO_2 waveform is also important to the physician as a figure of merit for the SpO_2 calculation. Poor perfusion or poor SpO_2 placement can lead to erroneous oxygenation figures, and is easy noticed by examining the waveform.

Blood oxygenation is computed by passing light at two different frequencies through the subject's skin, and measuring the amount of light absorbed. Given that oxygenated and de-oxygenated blood absorb different frequencies of light, one can compute the blood oxygen saturation percentage[3].

Our design called for a circuit capable of illuminated one of two LEDs with a constant current and measuring the amount of light picked up from a nearby PIN photodiode. We decided to do our initial development and testing using a commercial reflective SpO_2 sensor manufactured by Nellcor (Max-Fast, Nellcor/Tyco, Pleasanton, CA). Nellcor produces a series of disposable and reusable sensors which utilize a DB-9 connector with identical pinouts allowing for the ability to quickly test different devices and configurations.

The output stage of our device consists of a constant-current H-Bridge LED driver. The LEDs on the Nellcor sensor have their anode and cathodes connected. By driving one side high and the other low, one can choose which LED to illuminate. It is critical to drive the LEDs at a constant current to avoid coupling supply noise onto the received signal.

The input stage consists of a transimpedance amplifier connected to a PIN photodiode in the sensor followed by a differential amplifier used to subtract out the dc-offset of the signal and add additional gain. The offset subtraction signal is generated from the MSP430's on-board DAC. A feedback loop keeps the SpO_2 signal from saturating by adjusting the DAC output. The DAC value is also stored, and used for the blood oxygenation computation.

The system also measures background levels by repeating the measurements with both LEDs are turned off. These values are also used in the blood oxygenation calculation.

2.5 ECG

The ECG is perhaps the most important diagnostic signal for a health monitor, hence why almost all implementations include it. It's utility in diagnostic medicine is well established, and it has deep penetration in the consumer market in the form of wireless heart rate monitors by manufacturers such as Polar.

Although the ECG should be easy to measure given it's strength (a few mA), it does present some technical challenges. The ECG requires high-pass filtering to keep it's signal mid-rail. Our first implementation utilized a signal HPF (-3dB @ .79Hz), however we found the signal would frequency saturate. Increasing the -3dB cutoff was not an option as we would be losing information, given that the bandwidth of the ECG signal is in the range of 0.15Hz-150Hz[11]. Our second implementation added a second HPF stage, which proved to be sufficient to keep the the signal mid-rail. However, further testing revealed that Radio Frequency Interference (RFI) sporadically corrupted our signal, especially during motion. This is an often overlooked problem – one group reported that their wireless ECG monitor worked flawlessly, as long as the radio transmitter was not enabled. To solve this problem, we added an RFI filter prescribed by Analog Devices[4]. We have since seen amplifiers enter the market with built-in RFI filters, which would be a good choice for new designs. Our final implementation has been found to be robust and relatively immune to motion artifacts. Figure 4 demonstrates the quality of the ECG signal while the subject was undergoing significant motion.

2.6 EMG

Our initial interest in an EMG sensor was to measure back tension by measuring muscle activity as a proxy for stress. For this, we did not require a very fast input stage and reused the bulk of the ECG design. Given that the bandwidth of the EMG signal is in the range of 20Hz-8kHz, we could be more aggressive with the HPF. A single stage with a a -3dB of 7.9Hz was chosen and found to be effective. The amplifier we chose demonstrates a gain roll-off starting at 300Hz, and allows us to sample the signal at a slower rate. This is sufficient to measure a general level of muscle activity (Figure 3, top plot). However, for a more diagnostically relevant EMG implementation, we are considering migrating to a faster amplifier and higher sample rates.

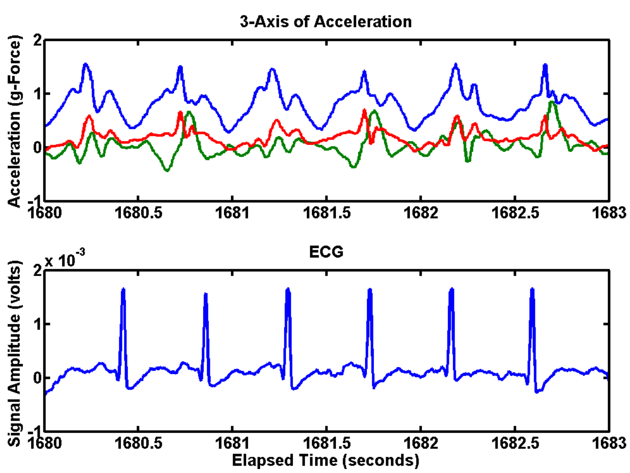

Figure 4. ECG signal (lower plot) captured while subject in motion (upper plot)

2.7 Respiration (Bioimpedance)

Monitoring respiration is important in a large number of cases. It's critical to monitor respiration in patients with cardiopulmonary issues such as congestive heart failure and patients who are on medication which suppresses breathing. It's also useful in diagnosing a number of conditions such as systemic inflammatory response syndrome (SIRS).

Our original attempt at respiration rate detection relied on acoustic pickup with a microphone, but we found bioimpedance to be a much more reliable indicator of respiration rate. Bioimpedance also allows us to measure relative breathing tidal volumes and to detect coughs. Bioimpedance can also be used for a number of other diagnostic measurements beyond respiration such as measuring cardiac output[1] and body fat[9], although we have not investigated these applications.

A high frequency sinusoidal signal (50kHz) is used so that the effects of the skin impedance are minimized through capacitative coupling. We chose a 350μA current at 50kHz in order to provide us with a good signal, while being significantly lower than the maximum safe current as defined by IEC60601-1-2005. Using a pair of electrodes connected to a high impedance differential amplifier, we can accurately measure the voltage difference between the electrodes, which is directly related to the impedance between those points.

We originally planned on building a fully analog impedance measurement system, however, we were able to significantly reduce the system complexity by utilizing Analog Device's AD5933 impedance analysis IC. The AD5933 has been shown effective in bioimpedance measurements[7]. Our output stage consisted of a voltage to current circuit followed by a 3.3nF DC blocking capacitor. The input stage utilizes the ECG electrodes, passes the signal through an RFI filter to a programmable differential amplifier which feeds the amplified signal back to the AD5933. Although the AD5933 should be very effective at filtering out any high-frequency noise we felt it would be prudent to add an RFI filter to the input stage – given it's importance in ECG measurements. The IC then computes the real and imaginary components of the signal which we convert to a magnitude and phase. The magnitude of the impedance is proportional to the chest volume, and breathing can easily be seen as a variation of the impedance (Figure 3, bottom trace). From this signal, we can determine respiration rate and approximate tidal volume.

2.8 Motion (Acceleration)

Most health monitors incorporate an Accelerometer. With this, one can determine body orientation, activity levels and perform fall detection. Furthermore, it is very easy to implement – hence it's prevalence.

Our first implementation utilized a 3-Axis analog accelerometer that was connected to the MSP430 by means of three analog input lines. Although this approach worked well, it used too many ADC channels of the MSP430. We later switched to the LIS302DL manufactured by STmicroelectronics. Not only does it provide a digital serial interface, but it also has built-in hardware fall detection. Although initially designed to detect the fall of a portable electronic device, the IC can be used to detect if the patient falls down without taxing the microcontroller.

3 The Harness & Electrodes

The harness (Figure 2) was initially prototyped in house and then replicated in a small, medium, and large sizes through a local manufacturing company. The harness is made of fabric that is designed to only stretch lengthwise. It has been sewn into a tube, and Velcro attached at either end allows the strap to be wrapped around the subject's torso and fastened.

The Berkeley Tricorder device, a Lithium-Polymer Ion battery, and a wiring harness were placed inside the strap through an opening that can be Velcroed shut. The electrode leads run through the inside of the strap, exit through button holes placed at regular intervals, and connect to the electrodes (Figure 5).

The electrodes are manufactured by Respironics (Model 16510-1, Murrysville, PA), and consist of conductive rubber, an affixed Velcro backing, and a slot for the electrode to attach. The electrodes have no adhesive or gel, and are held in place by the harness. The electrodes attach to small squares of Velcro on the inside of the harness.

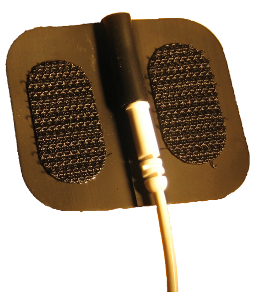

Figure 5. Non-Adhesive electrode with Velcro backing

4 The Applications

A C# application was written to connect to the device via Bluetooth and display real time data. The data is transmitted in 3-byte packets. The first byte contains the packet type, and the next two packets contain the payload. Longer messages, such as debugging message, are transmitted over several consecutive packets. Every second, a 3-byte sync block is transmitted (0x55, 0x55, 0x55) to allow for stream alignment between the Tricorder and the application. Between 7689 and 9993 bytes/second (Table 2) are required to transmit the data. The C# application has also been ported to a pocketPC mobile phone as a proof of concept and is able to display live streamed data.

5 Power Consumption

A good deal of effort was spent minimizing the current consumption of the device. The lower the power consumption, the longer the device can operate without a recharge. Most of components were chosen to optimize this parameter, however, several required components were unavailable in low power versions. Current measurements were made by measuring the voltage drop (V_{rms}) across a 1Ω resistor in series with the battery over 10 seconds. With the Bluetooth powered on, the device uses 55mA. Connecting to the Bluetooth device and streaming data resulted in an additional ˜10mA of power consumption, and writing to the SD card resulted in an additional ˜15mA.

6 Human Trials & Discussion

Two sets of human trials have been performed with a total of 24 participants ranging from 18 to 62 years of age. The subjects were fitted with the device, and asked to wear the strap for 24 hours while keeping a log of their daily activity. Fitting involved determining the ideal electrode placements along the strap and routing wires from the device to the electrodes. After the subject was fitted, a PC application was used to connect to the device via Bluetooth to validate that all the signals were being recorded. The firmware was programmed to shut down the Bluetooth connection 30 minutes after being powered on.

Our initial goal was to place a reflective sensor on the chest along with the other electrodes. Unfortunately, we found dermal vascular perfusion on the chest to be insufficient for SpO_2 measurement. As a result, we used a reflective forehead sensor (Figure 2) which provided an excellent SNR and is minimally influenced by motion artifacts. It has been shown that the upper arm is also a viable site for a reflective SpO_2 sensor[10], and we will investigate this as a much more convenient, and concealed, location for ambulatory monitoring applications.

The Bluetooth telemetry worked well under most cases, however, the device was placed under the left arm which caused significant signal degradation while the arm was covering the device. In future trials we plan on placing the device on the center of the chest where it will not be obstructed. We can achieve telemetry distances of up to 10m with a clear line of sight. In addition, the Bluetooth implementation worked quite well in automatically reestablishing broken connections.

Our SpO_2 circuit is currently the most problematic, but we feel that we can solve the problems in the next hardware revision. The microSD card write cycle consumes considerable current, causing noise on the supply place which coupled to the SpO_2 LED and produced regular spiky noise in the signal (Figure 3). A more aggressive constant current circuit should solve this problem. We also had problems with the signal saturating the transimpedance amplifier. We plan to solve this problem by reducing the first stage gain and utilizing a programmable gain second stage.

The memory card was ejected prematurely from the first subject so an enclosure was built to keep the memory card secure before being placed in the strap. The reported comfort of the system varied greatly based on the individual. Most people stated that the device was comfortable, except the electronics package bothered them while they were sleeping. The strap, although made of elastic material, was still a bit too restrictive for some subjects. The worst complaint received was from an overweight subject who resorted to putting additional bits of cloth under the strap for added comfort.

Some subjects exercised at the gym, and reported that their sweat made the electrodes, especially the forehead sensor, somewhat irritating. The forehead sensor has two plastic bulges where the LEDs and photodiode are encapsulated. Most subjects reported mild topical irritations after wearing the headband overnight, with an underweight subject reporting pain.

As this was only the first iteration of the strap, there is significant room for improvement. The location of the electronics package will be moved to the center of the chest where we expect it should minimize irritation while sleeping. The enclosure was fabricated from laser cut acrylic which greatly increased it's thickness. Fabricating a thinner enclosure will also help minimize the effects on the subjects while sleeping. The strap will also be redesigned so that it will have sections of elastic material along the length to provide more give and be less restrictive. Electrode irritation can be reduced if the subjects take the strap off occasionally. We also aim to reduce power consumption by simplifying portions of the circuit and by writing to the SD card in larger blocks to reduce the number of SD write operations.

The device performed very well under most conditions, allowing for rich parameter extraction such as heart rate, respiration rate, respiration tidal volume, and pulse transit time.

7 Conclusion

We have presented the Berkeley Tricorder, and ambulatory health monitoring device with Bluetooth-enabled telemetry. The device has been designed to provide a rich feature set in a small form-factor. This, combined with reusable electrodes and any contemporary cell phone is capable of filling an important need in rural communities in developing countries. Our initial trials have validated the platform as robust with data quality comparable to most modern monitoring devices. Moving forward, we are looking at adding additional sensors and optimizing the existing sensors with regards to current consumption and cost. We are also working on leveraging our Bluetooth platform further, allowing for interoperability for existing Bluetooth devices as well as developing the dial up networking features for telemetry without the need to develop different applications for different cell phones.

8 Acknowledgments

I would like to thank the Department of Bioengineering at Berkeley for funding through a NIH training grant, VG Bioinformatics for collaborating with us and providing funding for development, and to Patrick Goodwill and Steve Conolly for invaluable help and support.

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