AN ABSTRACT OF THE THESIS OF

Samuel House for the degree of Master of Science in Electrical and Computer Engineering presented on March 20, 2013.

Title: Passive Health Monitoring With Wirelessly Powered Medical Devices

Abstract approved: _______________________________________________________

Patrick Y. Chiang

The proliferation of body worn autometric devices has been enabled by advances in low-power electronics and fueled by the quantified-self movement. These devices range in complexity from pedometers to clinical vital sign measurement. They all share the same drawback, typically the most expensive and heaviest component, the battery. The future of autometric devices lies in wireless power. This work explores what is required from autometric devices and presents the results of testing both an embedded version and an application specific integrated circuit (ASIC) version of a wirelessly powered autometric device.
Passive Health Monitoring With Wirelessly Powered Medical Devices

by

Samuel House

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Major Professor, representing Electrical and Computer Engineering

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Director of the School of Electrical Engineering and Computer Science

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Dean of the Graduate School

I understand that my thesis will become part of the permanent collection of Oregon State University libraries. My signature below authorizes release of my thesis to any reader upon request.

__________________________
Samuel House, Author
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For my family, who have wondered what I’ve been doing and why I don’t have time to visit them.
“How are you today?” This is a common and often rhetorical question that rarely has anything better than a subjective answer. When asked in passing or by polite company a qualitative response is expected if not encouraged. However, when quantifiable responses are required, things change. Currently, health professionals recommend physical exams, or preventative health exams, at the most, once per year and in many cases just once every two years [1], [2]. These exams should include measurement of weight, height, and blood pressure. Complete adherence to recommendation provides five to ten samples per decade. There is no question that there would be difficulty correlating those data to lifestyle changes. Of course, people collect some of this data themselves on a more regular basis. People are most adherent to this during periods of increased interest in physical well being, for instance, when people change their diet for a few weeks they will also often track their weight. If there is one device that is synonymous with quantifying progress in fitness goals, it is the bathroom scale. The logging of this data is typically done in a single way, by remembering yesterday’s measurement. Advances in technology have provided a solution to this poor data logging method. Antiquated counterbalanced scales have been replaced by WiFi enabled smart scales. These scales are capable of not only massing a human, but also measuring body composition and calculating body mass index (BMI) while syncing all this information to a secure web server. These are important metrics, but the bathroom scale is missing out on the action, quite literally.

Activity levels are a major contributor to the overall health of an individual [3], [4]. This metric is, at best, self-reported during physical examinations and even then is usually unreliable [5]. What is needed to accurately capture this metric is a device that moves with the individual. This dilemma has spawned a plethora of body worn auto-metric (measurement of oneself) devices for recording and quantifying various aspects of our active lives. Some of the first and still most common of these devices are the actigraphs, more specifically pedometers, which simply measure the number of steps the wearer has taken. Great strides have occurred in the actigraph market in recent years, devices have become more fully featured and more complex than their simple step
counting predecessors. Modern actigraphs include microcontrollers or application specific integrated circuits (ASICs), wireless communications, rechargeable batteries, USB interfaces, six or nine axis inertial measurement units (IMUs), and a host of other sensors. Some of the notable devices on the consumer end of the spectrum include the Fitbit Tracker [6], the Nike+ Fuelband, and the Jawbone UP. These devices are non-diagnostic consumer medical devices (NCMDs), meaning they are sold to track activity and not as a means for diagnosing any medical conditions. This limitation announced by the respective manufacturers is more likely due to a litigious society rather than any technological impediments. These NCMDs are aimed at consumers, thus, usually do not provide users with the raw data they collect, typically opting towards a web-based subscription model that displays abstracted measurements, in some cases, with proprietary units [7]. Some manufactures provide an open platform for their devices, these devices typically have similar, if not identical, components [8] to the closed platform devices but without the data interpretation services. This class of device is the most attractive for researchers and clinicians because they provide raw sensor data [9], [10]. In both classes, open and closed platform, several device manufactures have focused on the accuracy and quality of their data collection and, for all intents and purposes, these devices are considered medical grade, typically have achieving Food and Drug Administration (FDA) approval [11].

The recent explosion of personal quantification devices is closely associated with what has been called the quantified-self (QS) movement [12]. The QS movement is part of an emerging patient driven care model [13]. This is the DIY of health care. The QS movement is defined by its use of electronic devices to measure, track, and compare autometric data. Many people are interested in self-improvement, a subset of those people implement tools to actually accomplish self-improvement. For the dedicated self-augmenter, data collection is an important aspect of tracking progress toward a goal; self-knowledge through numbers, this is the basis of the QS movement. Anyone who has attempted to manually keep track of physical activity the old fashioned way, by scraping carbon mineral on dried wood pulp (pencil and paper), knows it is difficult to be consistent over long periods of time. A lapse in prompt recording of activities as small as one day results in severely reduced accuracy of those reports [14]. We use autometric devices to help us be more accurate, and honest, in our measurements. However, the devices themselves still require adherence in the form of maintenance. What is really
required for these QS tasks is a technology that we do not consider technology. Passive technologies that perform their function without drawing any attention to themselves, causing us to forget they are there, sometimes causing us to forget that they are even technology. Eye glasses are an excellent example of this type of technology, but perhaps too low-tech for a proper comparison. Pacemakers have this property. Pacemakers are very advanced devices, modern devices adjust the user’s heart rate based on physical activity [15], they do this without requiring any conscious input from the user. Imagine a pacemaker that must be manually set each time a new activity level is reached or one that requires the user to remember to periodically charge it; entirely impractical. Clearly, in the medical domain this device feature, autonomy, is especially important. For fitness monitoring, the obvious presence of devices may sometimes benefit the user with a Hawthorne-like performance increase [16], similar to an observer effect [17]; basically, people work harder when they know the are being monitored. However, this can be achieved as desired in software, rather than as a side effect of the hardware. For clinicians, being able to accurately measure changes in health or activity levels that are more likely due to the treatment under study rather than the addition of some device, would be an ideal situation.

Each new body worn automatic device adds features or automation processes that bring users or researchers closer to a seamless and ubiquitous monitoring situation. However, each device still has the same drawback, these devices all run on batteries that require periodic charging. This may seem to be a small inconvenience, but only for the reason that users are so accustomed to needing to perform this maintenance. It’s clear that the future of truly continuous and unobtrusive automatic devices are ones that require the absolute minimum in user compliance. This work demonstrates the feasibility of designing and building a wireless powered automatic device prototype from commercial-off-the-shelf (COTS) components and examines a custom ASIC designed for the same purpose. The completed embedded platform, the Medical Electronic Device/Implant Communicator (MEDIC), is capable of wirelessly powering and communicating with a body worn automatic device. The MEDIC also completes the chain of communication by uploading data collected from devices to a secure server via an Android smartphone interface [18]. Truly continuous and unobtrusive automatic devices will let us collect data at unprecedented resolution and scale. They will be an invaluable tool for consumer and clinician alike.
Chapter 2: Background

A subset of the VLSI group at Oregon State University led by Dr. Patrick Chiang has been researching and building medical devices for several years. Much of this work to date has focused on building embedded platforms. These platforms have been used to determine ideal sensing methods [19], test sensor filtering [20], and application testing [18].

2.1 Previous Work

The two main embedded platforms from our previous work were the OSU Life and Activity Monitor (OLAM) [21] and the Personal Dead Reckoning Fiducial Updated Device (PDRFUD) [22]. The OLAM was designed and built to be used in a clinical trial involving the assessment of lipoic acid supplementation on circadian rhythms [23]. It featured an accelerometer, gyroscope, and full ECG analog front end. The OLAM made periodic measurements of heart rate and activity level measurements over two week periods and required a 2200 mAh battery to do so. The PDRFUD was designed to precisely track foot movement through three-dimensional space using low-cost COTS components. This ability allowed for accurate and inexpensive monitoring of gait velocity and activity. It included components to make up a nine-axis IMU, wireless communications radio in an industrial, scientific, and medical (ISM) band, and a passive radio frequency identification (RFID) reader for location updates. The PDRFUD required a 1700 mAh battery and extensive longevity tests were not performed, but the calculated and informally observed battery lifetime was less than one week.

The large batteries required by these devices not only increased their cost and weight, but also decreased their usability for long-term studies and added the complexity of user/patient compliance with device charging/battery replacement. Especially in the case of the PDRFUD, which had an application tracking the gait velocity of patients at risk of Alzheimer’s disease, requiring patient compliance would likely rule the device out entirely in such use cases.
From our previous work with medical devices, we devised some requirements for creating a passive autometric device.

THE SYSTEM SHALL BE UNOBTRUSIVE. This is not to be confused with non-invasive. Non-invasive is assumed for all of our current and near future systems; in medicine, invasiveness means entry in to the body by surgical implantation or existing cavity. Unobtrusiveness means that the body worn device falls in to the category of invisible technology. There are several aspects to this requirement spanning from the functionality of the system to the user’s perception of the body worn device. The placement of the device should be in the user’s aura-of-self, which is the small area around the user’s body that they perceive to be a part of their body, further described by proxemics [24]. In practice, this means minimizing size and mass as much as possible.

THE SYSTEM SHALL REQUIRE MINIMUM USER COMPLIANCE. Basically, the user should not be required to preform any maintenance for the system to continue operation. For non-invasive systems the typical minimum compliance would be wearing the system, but due to a lack of required maintenance the system could be embedded in clothing or shoes. In an extreme minimum user compliance situation the user would only be required to not remove an already applied system, such as in the case of an adhesive patch or bandage. In practice, this means devices on the body should be wirelessly powered and that data is transmitted off the device wirelessly.

THE SYSTEM SHALL BE FLEXIBLE AND MODULAR. It should be a platform that sensors or sensing modules can plug in to. This allows the freedom for modifying the front end of the system depending on the vital signs or environmental metrics desired. The system would be easily scaled to the task at hand. This also allows for multiple specialized nodes to be placed on different parts of the body. All vital signs and environmental metrics are not measurable from any one point on the body; a requirement of flexibility in sensing implies the ability to measure multiple vital signs and environmental metrics simultaneously. In practice, the system should make up a wireless body area network (WBAN), meaning it can consist of multiple devices.
2.2 Design Considerations

In summary, from our previous work we concluded that we needed, a small form factor wirelessly powered device with wireless communication abilities. It was clear that this new system would, initially, be at least two parts, one part consisting of the body worn device and one part consisting of the base station that transmitted power and communicated with the body worn device. A small form factor is a built-in feature of an ASIC design, which was therefore an obvious choice for the body worn portion of this system. An embedded system would make up the base station, for ease of development and highly customizable nature.

2.2.1 Wireless Power

Wirelessly powered devices have been around for over a century in the form of crystal radios. Crystal radios do not store energy to power devices, they simply rectify and filter AM radio and play the result though a speaker. When we discuss wireless power, we mean it as a sort of power cable replacement for devices. The concepts of wireless power are simple; the practical implementation is another matter entirely. All wireless power transmission occurs through the electro-magnetic (EM) spectrum using either near-field effects or far-field effects. For radiators that are less than one half the wavelength being used, near-field effects typically dominate to a distance of one wavelength from that radiator and, with some overlapping space, far-field effects dominate after one-two wavelengths. These distances can be adjusted, using a larger radiator will produce longer range near-field effects. The practical difference lies in the frequency, and therefore distance over which, the transmission occurs. Higher frequencies, like microwaves, visible light, and the higher end of the radio-frequencies (RFs), have short wavelengths, to the point that getting a radiator and receiver physically close enough to leverage near field effects is impractical. There are two primary methods used and, with a great deal of grey area, they are separated by the energy carrier frequency.

**Inductive Energy Transfer**

This is usually in the low frequency range and is a type of near-field energy transfer, and is the most common method of wireless energy transfer. It is used in transformers and in recent wireless charging products for charging consumer de-
VICES. When the transfer coils are closely coupled, induction offers high efficiency, sometimes greater than 98% in larger, closely coupled, transformers. It can be implemented using relatively simple circuits with small component counts, for instance, a class E amplifier and a printed spiral coil. Class E amplifiers have well researched and documented design and feedback methods. The spiral coil shown in Fig. 2.1 was calculated, designed, and drawn in CAD software by a Matlab script [25], the measured inductance was within 3% of the simulated value. While inductive wireless power systems are simple to design, inductive energy transfer is typically only effective at very close range and, in most cases, only to one coupled coil at a time. This is useful for certain devices, namely implanted medical devices. This is indeed the method used by most transcutaneous energy transfer (TET) systems. The benefit of near-field energy transfer in this case is that the energy transfer coil can double as a communication coil using load shift keying (LSK) through the reflected impedance of the implanted coil [26]. However, for use in WBANS, inductive energy transfer systems are not ideal.

Figure 2.1: A printed planar spiral coil radiator.
Radiated Energy Transfer

Typically uses higher frequencies, including frequencies from the higher RF up through microwaves, and is a type of far-field energy transfer. For very low-power devices, ISM band RF are the preferred method for far-field energy transfer, primarily because of the ability to scavenge from the already highly saturated bands. This is the method we are going to use because it works well for multiple devices at a distance and has better immunity if the devices exhibit motion relative to one another.

There are many factors that play into how much of the transmitted power is received by the remote antenna. Specifically for far-field situations, where distance from the antenna is much greater than the wavelength, the free space loss equation can be used to loosely estimate the received power efficiency,

\[
L_F = \frac{P_t}{P_r} = \left(\frac{4\pi R^2}{\lambda^2}\right)^2
\]  

(2.1)

Where \(P_t\) is the transmitted power, \(P_r\) is the received power, \(4\pi R^2\) is the surface area of the sphere centered around the radiator, \(\lambda^2\) and is the wavelength squared. We’re usually dealing with frequency rather than wavelength, so we can substitute \(\lambda\) for \(\frac{\nu}{f}\) where \(f\) is the transmission frequency and \(\nu\) is the speed of the wave in the medium. For us, the medium is air, the speed of radio propagation in air is approximately equal to \(c\), the speed of light in a vacuum. This equation also assumes antennas with zero gain.

Many antennas are rated in terms of how they differ from one that radiates its power in a perfect sphere, isotropically; in all directions equally, which is where the spherical term comes from in the free space loss equation. The assumed antennas in this basic free space loss equation have a rating of 0 dB, that is, a gain of 0 dB with respect to what one would expect to see from a perfect isotropic antenna. It’s clear from this equation that efficiency is not great for wireless energy transfer, even in this simplified estimate. For instance, if we start by transmitting 0 dBm (1 mW) at distance of one meter the power received would be \(-31.6\) dBm (0.68 µW).

This is not great. That is 1500 times less power. It gets worse, but it also gets a little better. In practice, the free space loss equation is a bit too simple. There are many losses and a few gains we should include in order to make a better estimate of
what we can call our link budget. For many RF systems, values are often expressed in
$\text{dBm}$, which are an expression of power ratio in decibels referenced to 1 $\text{mW}$. Having
an equation expressed in power will be useful for future calculations. Thus, our slightly
more complex link budget equation in terms of power is,

$$P_r \approx P_t + G_t + G_r + 10\log_{10}(L_F) - 10\log_{10}(1 - |\Gamma_t|^2) - 10\log_{10}(1 - |\Gamma_r|^2)$$  \hspace{1cm} (2.2)

As before $P_r$ is the received power, $P_t$ is the transmitted power, both in $\text{dBm}$, $G_t$
and $G_r$ are the gains for the transmitting and receiving antennas respectively in $\text{dB}$,
the $L_F$ term the free space loss term from Eq.2.1, and finally $10\log_{10}(1 - |\Gamma_t|^2) \hspace{1cm}$
and $10\log_{10}(1 - |\Gamma_r|^2)$ are the losses (in $\text{dB}$) due to impedance mismatch, for the transmitting and receiving antennas respectively. The $10\log_{10}(1 - |\Gamma_r|^2)$ term especially can be
quite large if the antenna being used near the body is not matched for that environment.
We will still be assuming line of sight (LOS) with this new equation. These terms can be
seen where they occur in the simplified block diagram in Fig. 2.3.

Now that we’ve made our equation more complete, we can do some engineering
simplifications to it. As can be seen in Eq. 2.2, now that we’re working with log scale,
Figure 2.3: Point source losses of power lost in transmission to body worn sensor node.

all the terms besides the free space loss simply shift the free space loss curve up or down. Thus, if we assume a one meter distance for our estimates, we can find the loss at that distance and simply sum it with the other elements. As seen in Fig. 2.2 loss at one meter is between 31 dB and 32 dB, we’ll estimate on the safe side and assign a free space constant loss of 32 dB assuming one meter distance. Now we have,

\[ P_r \approx P_t + Gains - Losses - 32 \text{ dB} \quad (2.3) \]

In the use of this equation, we get to set how much power we’re transmitting, the distance to the receiving antenna is also known, the return loss from impedance mismatch can be measured using a network analyzer, and the gains of the respective antennas are listed in their datasheets. Components with pressing design constraints can be selected first and then this equation can be used to help select the less constrained components based on the properties of the more constrained components. For instance, if we wish to deliver 0.5 mW (−3 dBm) to the body worn device, we know our free space loss for one meter is 32 dB, we assume matching losses can be lowered to 2 dB, and the maximum power output for a COTS RF amplifier is 27.5 dBm. We still need to select antennas, our body worn device antenna needs to be small, and the best fit found has average gain of −1.27 dB, so to meet the power requirements of the body worn device and to allow some margin for error, our transmitting antenna needs about 6 dB of gain to provide 0.74 mW to the device.
Table 2.1: Total Task Energy Calculation

<table>
<thead>
<tr>
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<th>Current</th>
<th>Duration</th>
<th>Energy Cost</th>
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<tr>
<td>Step 1: Wake Up</td>
<td>1 mA</td>
<td>120 µs</td>
<td>1 mA * 3.3 V * 120 µs ≈ 4 µJ</td>
</tr>
<tr>
<td>Step 2: Sample UV Sensor</td>
<td>100 µA</td>
<td>20 µs</td>
<td>100 µA * 3.3 V * 20 µs ≈ 7 nJ</td>
</tr>
<tr>
<td>Step 3: Initialize the Radio</td>
<td>9.4 mA</td>
<td>5 ms</td>
<td>9.4 mA * 3.3 V * 6 ms ≈ 186 µJ</td>
</tr>
<tr>
<td>Step 4: Send Sensor Data Packet</td>
<td>13.6 mA</td>
<td>1 ms</td>
<td>13.6 mA * 3.3 V * 1 ms ≈ 46 µJ</td>
</tr>
<tr>
<td>Step 5: Sleep and Wait</td>
<td>1.5 µA</td>
<td>60 s</td>
<td>1.5 µA * 3.3 V * 60 s ≈ 300 µJ</td>
</tr>
<tr>
<td>Total</td>
<td>N/A</td>
<td>60.01 s</td>
<td>≈ 536 µJ</td>
</tr>
</tbody>
</table>

This is a good time to quickly reiterate the difference between power and energy. Metaphorically, energy is money, power is how fast it is spent or acquired. We’ve been discussing the delivery of power, which is delivery of energy per unit time. If we consider that the body worn sensor has sufficient energy storage then what we care about is the average power for the time it takes a given task to complete, or to phrase it differently, the total energy required by a task. For tasks with known duration that have a higher average power than we can receive wirelessly, we can store energy until we can complete that task; to go back to the metaphor, if we want to spend money for a month faster than our monthly income rate, we need to save money until we have enough to spend it at the higher rate. For instance, in a particular sensing task called UV-Sample-and-Report, the body worn device needs to wake up, sample an ultraviolet (uv) sensor, transmit the sampled data, and then go back to sleep for one minute before repeating. This process takes a specific and regular amount of time, though different steps in the process require different amounts of power. This would be like taking a trip to the beach one day, text messaging a friend about it, and then going back to a normal, lower than income, spending level for the rest of the month. Electronic devices are required by thermodynamics to live at or below their means, so we must plan these expenditures carefully. To calculate the average power in the task UV-Sample-and-Report we first find the energy required for each step and then divide by the total time the task takes. We will quantify this example. First we will say our system voltage is 3.3 V, this is a common COTS device supply voltage. Second, we will calculate the energy requirements of the steps in the example task, as seen in Table 2.1. Third, calculate the average power. For this example task, which uses realistic values, the average power is 8.93 µW. This means that after the other transfer losses we would need to deliver about −20.49 dBm to the body worn device to keep this task active, again assuming that the body worn device
can buffer the required energy for the short periods of higher power. For our previous example we would be able to lower our the transmission power to about 9 dBm.

2.2.2 Wireless Communications

Low-power wireless communications are ubiquitous. Some of our previous work already included low-power wireless communications. However, these were pre-built solutions, typically Bluetooth, which may not lend themselves to the flexibility required by our system. Radio systems have a few tiers involved in their design or selection, the lowest level is modulation, on top of that is encoding, and finally there is protocol. Working from the top, protocol defines things like encryption, handshakes, preamble, and endianness. Encoding, or line coding, is one of the ways we ensure the data in our protocol doesn’t get corrupted, this is usually done by adding information that allows the receiver to better receive the data. Finally, modulation is how we’re altering the RF wave to convey the information from the previous layers. We can only alter RF waves in so many ways, primarily this takes the form of shifting the frequency of the carrier wave or changing the amplitude of the carrier wave. A good way to think about how these layers work is to relate them to human communication, the protocol can be thought of as the language used, encoding like word inflections and pauses between words, and modulation like the use of vocal cords to vibrate the air. In our system we will be using relatively low data rates and will not be transmitting very far, so we will keep this in mind and start by examining two of the simplest modulation methods and, from there, we can add look in to encoding and building more complex protocols. Finally we’ll cover a few basic parameters to look at when dealing with wireless communication systems and what they mean in our system.

On-off keying (OOK)

OOK modulation is achieved by simply turning the carrier wave on or off, signifying either a digital one or digital zero. OOK modulation is the simplest form of amplitude shift keying (ASK) which is the discretized version of amplitude modulation, better known as AM radio. This is a very simple modulation technique to implement for both transmitter and receiver. An example of OOK RF modulation is shown in Fig. 2.4. For the simple case of OOK modulation the data rate must
be known on the receiving end, else any sequence of bits, like the repeated ones in Fig. 2.4, will not be differentiable.

Frequency shift keying (FSK)

FSK modulation alters the other aspect we control on a RF wave. By shifting the frequency away from the carrier wave frequency, we signify either a digital one or a digital zero. In the case shown for Fig. 2.5, a decrease in frequency denotes a digital zero while the carrier frequency alone denotes a digital one. The same problem occurs in simple FSK modulation as with simple OOK modulation. Strings of ones or zeros can easily blend together.

Manchester coding

The simple modulation schemes discussed here share the similar problem of recovering clock information from the data, especially in the case of many successive ones
or zeros. We can make things simpler by adding some encoding to our transmitted data. A very common and simple to implement line coding scheme is called Manchester coding. To implement Manchester coding an exclusive-OR (xor) operation is performed on the clock and the data from the transmitting device, the output of that operation produces a new data stream for modulation. For the Manchester coding simple OOK modulation, the results are seen in Fig. 2.6. The xor operation ensures that there is at least one transition per bit. Although Manchester coding may be more difficult to read for a human, it is quite a bit easier for a computer or microcontroller to read correctly. The downside of Manchester coding is that the bit rate is doubled while the data rate remains the same. This can been seen in Fig. 2.6 where the shortest bit time is half that of the same non-coded data seen in Fig. 2.4.

![Figure 2.6: OOK RF modulation encoded by IEEE 802.3 Manchester Code.](image)

**Receiver sensitivity**

In simple terms, the sensitivity of a receiver is how large a signal needs to be for that signal to be correctly received with a given error rate. A typical plot of this parameter will be for a specific bit error rate (BER). For instance, at 1% BER, a typical plot might look something like Fig. 2.7. For any fixed data rate and BER, a lower sensitivity is better. Different modulation schemes will usually have different sensitivity plots, but they are all typically the same shape, just shifted up or down. The plot does not continue indefinitely, eventually the receiver sensitivity crosses over a maximum, sometimes called saturation. As the maximum is approached the range decreases, in some cases additional attenuation is required to shift the received signal to this range, to avoid swamping the receiver.
2.3 Wirelessly Powered BAN

A wirelessly-powered body-area-network (WPBAN) was designed by Jiao Cheng, Lingli Xia, Chao Ma, Patrick Y. Chiang, et al [27]. This WPBAN harvested energy from a supplied 431.16 MHz RF energy carrier and included a medical implant communication service (MICS) band radio for communication. A second revision, the WPBAN-2, was made that harvested energy at 904.5 MHz and used the same MICS radio to communicate at 402 MHz. Both WPBAN devices used global clock scheme where the devices generated their local clock signals based on division of the energy carrier wave frequency. The communications radio on the WPBAN-2 uses OOK modulation at 250 kbps with no encoding. The protocol allows for four programmable start bits, followed by ten data bits, and completed with four stop bits. The datum frame can be seen in Fig. 2.8 with a sample of the data and RF modulation. The protocol does not include preamble or synchronization bytes. The lack of these standard reliability items has to do with how the clock is generated, because the data clock is a division of the energy carrier frequency, the two will theoretically be in phase.

The WPBAN-2 has a sensitivity down to $-20 \, dBm$ for receiving power. This will be an important metric to use in our wireless power calculations.
With these considerations in mind a new embedded platform was developed to test the feasibility of wirelessly powered sensors, wireless power capabilities, and applications related to invisible autometric devices. The embedded platform would also be able to perform the task of powering and communicating with the WPBAN-2.
Chapter 3: Overview of the MEDIC

The MEDIC is a prototype system consisting of at least two separate boards, a base station and a sensor node. The MEDIC Base supplies wireless power and has two way communication ability with both the nodes and a connected Android smartphone. The MEDIC Node is a wirelessly powered body worn sensor package, there can be multiple nodes that communicate with a single base in a time multiplexed scheme. The MEDIC Base is also capable of periodically listening for other RF power sources and turning off its own RF power when sufficient external supplies exist. Thes external supplies, the MEDIC Power Stations, were not part of this work. The conceptual daily use scenario for this system involves a subject carrying a MEDIC Base on their person and wearing one or more wirelessly powered nodes to measure one or more biosignal or environmental conditions. For instance a subject can wear three MEDIC Nodes, one to collect UV index and barometric pressure, one to collect accelerometer data, and one to collect blood oxygenation. Initially the MEDIC Base will power the nodes and communicate their sampling and reporting frequencies. For example, perhaps every five minutes UV index and barometric pressure should be sampled and reported, accelerometer data should be sampled every ten seconds and reported every minute, while blood oxygenation should be sampled twice a second and reported every minute. These data would be collected by the MEDIC Base and sent to the Android smartphone for further processing and display or transmission to a cloud platform for further processing and storage. The programmability of these tasks also allows for tapering based on conditions such as available wireless power, like increasing the sampling/reporting frequency based on being in the presence of a MEDIC Power Station or not. With a small amount of wireless power infrastructure in a building, the MEDIC Base wouldn’t be required to supply RF power at all, acting only as a communications and programmer for the MEDIC Nodes.

The MEDIC Node was designed in order to compare the WPBAN-2 to another wirelessly powered autometric device. The MEDIC Base was designed to work with both the MEDIC Node and the WPBAN-2. The two separate embedded designs, the MEDIC Base and the MEDIC Node, share some of their primary components, namely the microcontroller
and communications radio. This was done for simplicity of design, code reuse, and the unmatched quality of the components. A discussion of the hardware choices and firmware designs are found below.

3.1 Hardware

The hardware of the system will define our limitations for power output and communication abilities. Flexibility and options are desirable when selecting hardware.

3.1.0.1 Microcontroller

A Texas Instruments (TI) MSP430FR5739 ultra-low power microcontroller was selected for use in the MEDIC system. This microcontroller provides several advantages with its integrated Ferroelectric RAM (FRAM), enhanced universal serial communications interface (eusci), 10-bit analog-to-digital converter (adc), and integrated real-time clock (RTC). The FRAM allows for ultra-low power non-volatile memory used for storing data or microcontroller states before entering low-power modes. The eusci will allow us to communicate with any other integrated circuits (ics) on board. The adc enables us to add sensors with analog outputs. Finally the rtc lets the microcontroller interrupt from ultra-low power sleep modes in increments of minutes rather than the microseconds allowed by using a traditional timer-counter.

3.1.0.2 Communications

Each board of the MEDIC system includes a TI CC110L low-power radio. This radio was one of the few radios specified as capable of achieving the data rate and modulation used by the current WPBAN-2. This radio is finely programmable within three bands in the lower third of the ultra-high frequency (UHF) spectrum, [300 MHz-3480 MHz], [387 MHz - 464 MHz], or [779 MHz-928 MHz]. This made it ideal for use on both of the MEDIC boards as it covered both the MICS band and the 904.5 MHz required for the WPBAN-2 energy carrier.
3.1.1 MEDIC Base

![Figure 3.1: PCB layout for MEDIC Base.](image)

3.1.1.1 Energy Transfer

The MEDIC Base uses the TI CC110L radio and a TI CC1190 front-end amplifier to achieve the desired output power. The specified maximum power output for the CC1190 is $27 \text{ dBm}$ (500 mW). In application we were able to achieve a 24.75 dBm output, as seen in Fig. 3.2. This is sufficient power output for most applications of the wirelessly powered devices.

3.1.1.2 Power Management

Two power management chips are on the MEDIC Base. The first is a TI BQ24190, which accepts a USB connection and charges a single cell Li-Ion battery. The BQ24190 has a switching architecture and can source 4.5 A. The second power management chip is a
20

Figure 3.2: Maximum output power for the MEDIC Base at 904.5 MHz.

TI TPS76733, a linear regulator capable of sourcing 1 A.

3.1.1.3 Bluetooth Communications Radio

A RN-41 Bluetooth module serves as the communication method between the MEDIC base and its associated Android smartphone. The MSP430FR5739 on board the MEDIC Base communicates with the RN-41 via a universal asynchronous receiver/transmitter (UART) protocol.
3.1.2 MEDIC Node

3.1.2.1 Energy Harvest

Only recently have COTS RF energy harvesting ICs been available. The IC used in this design is a P2110 from Powercast. It harvests energy from a 915 MHz RF source generated by the MEDIC Base. The P2110 is capable harvesting power down to -11.5 dBm input and contains a 3.3 V boost regulator for supplying power to the MEDIC Node.

3.1.2.2 Sensors

This platform is, for the most part, agnostic to the sensors contained on it. For testing purposes, sensors of various complexity and power requirements were included in this design. The MEDIC Node includes a six axis IMU, a barometric pressure sensor, a UV light sensor, and a set of temperature sensors.
3.2 Software

The software for both boards combined, as they shared a great deal of code, was several thousand lines. Great care was taken to highly abstract the operation of the MSP430FR5739, the CC110L, and other ICs used in the MEDIC system. To this end, code for each of the sub-modules in this MSP430 family were written from scratch. This enabled fully abstracted operation of the EUSCI, including the full duplex serial peripheral interface (SPI) operation required by the CC110L as well as inter-integrated circuit (I2C) and UART protocols, the FRAM, the timer counter modules, ultra-low power modes, real-time clock module, and the ADC.

3.2.1 MEDIC Base

The code flow for the MEDIC Base is a interrupt driven flag based loop. The microcontroller waits for an interrupt from either communications radio, the MICS band radio, or the Bluetooth radio. The microcontroller then services the flag, whether it is to turn on the power transmission or relay data from the Node to the Android, and goes back to sleep. The power savings from sleep mode is lost in the magnitude of the RF power requirements, but is beneficial at times when the RF power is not used.

3.2.2 MEDIC Node

Currently the MEDIC Node is set up to wake up from deep sleep every minute, sample the temperature, store the measured temperature and the current time to the FRAM, and then transmit the real time clock information and temperature back to the MEDIC Base. The MEDIC Node then sleeps for another minute. This is close to the planned code flow diagram shown in Fig. 3.4, except that with only one MEDIC Node being tested currently, a clear-to-send check is not required.
Figure 3.4: Code flow diagram for the MEDIC Node.
Chapter 4: Results

4.1 MEDIC

The MEDIC Base performed as expected. The communications radio was able to send and receive data packets in both MICS and ISM bands. When communicating with another COTS radio, the MEDIC Base could communicate using both OOK and FSK with data rates from 1.2 kbps to 250 kbps. The RF power output was measured for various input powers to the amplifier and the current consumption of the board was simultaneously measured. The tests were performed with a carrier frequency of 915 MHz. The results of the RF power testing can be seen in Table 4.1.

The MEDIC Node was able to communicate sensor information with time stamps back to the MEDIC Base which then relayed it to an Android tablet. While the MEDIC Base was able to reliably charge the storage capacitor fully on the MEDIC Node, the boost regulator did not reliably engage. Successful wireless powering occurred infrequently, but the amount of available energy when the boost regulator did engage was too small for wireless data transfer. Simultaneous data transfer and wireless power testing was not possible. Further refinement of the MEDIC Node software may allow for transmission with the available energy in addition to an increased storage capacitor size. Power of the MEDIC Node was measured for each high level function called from the main code. The power profile for the MEDIC Node can be seen in Fig. 4.1. It should be noted that the Sleep function has been truncated in the plot. The other functions, the majority of space on the plot, only represent 0.038% of the minute long cycle the MEDIC Node was operating on for these tests.

4.2 WPBAN 2

The WPBAN-2 was successfully wirelessly powered using the MEDIC Base. The communications were poor quality and are discussed more in Chapter 5.1.
Table 4.1: RF Power Output from the MEDIC Base

<table>
<thead>
<tr>
<th>Amp Setting</th>
<th>Amp In [dBm]</th>
<th>RF Out [dBm]</th>
<th>Total Board Current [mA]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Low Gain</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>00</td>
<td>23.2</td>
<td></td>
<td>295</td>
</tr>
<tr>
<td>05</td>
<td>23.9</td>
<td></td>
<td>319</td>
</tr>
<tr>
<td>07</td>
<td>23.94</td>
<td></td>
<td>320</td>
</tr>
<tr>
<td>10</td>
<td>24.23</td>
<td></td>
<td>352</td>
</tr>
<tr>
<td>11</td>
<td>24.24</td>
<td></td>
<td>344</td>
</tr>
<tr>
<td>High Gain</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>00</td>
<td>23.66</td>
<td></td>
<td>315</td>
</tr>
<tr>
<td>11</td>
<td>24.44</td>
<td></td>
<td>385</td>
</tr>
<tr>
<td>Power Down</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>00</td>
<td>-34.6</td>
<td></td>
<td>86</td>
</tr>
<tr>
<td>11</td>
<td>-20.56</td>
<td></td>
<td>91</td>
</tr>
<tr>
<td>Radio Off</td>
<td>N/A</td>
<td>N/A</td>
<td>74</td>
</tr>
</tbody>
</table>

Figure 4.1: Average power consumption by function of the MEDIC Node.
Chapter 5: Discussion

In order to better understand the results and to provide recommendations for the future work to be done in this area, we provide a discussion on the two main components of this work, the digital communications and energy harvesting.

5.1 Communications

The communications on the WPBAN-2 perform well in laboratory conditions and modulated waveforms can be measured and manually bit aligned. However, the WPBAN-2 has a significant data rate clock drift related closely to its input power. To the point that the presence or body position of a nearby human would alter the data rate. As explained in 2.2.2, it is critical in unencoded OOK modulation that the clock rate be known. This can sometimes be accounted for in systems with preamble bytes, a string of alternating zeros and ones. The data rate changes more slowly than the time it takes to transmit and receive one packet. So, the data rate can be measured per packet during the transmission of a known sequence; the preamble bytes. However, the WPBAN-2 includes only four start bits, half a byte, which was intended for use as a device address. Additionally the maximum data rate that the CC110L is capable of receiving for OOK modulated signals is 250 kbps; the WPBAN-2 exhibited a data rate that would vary from 200 kbps at its minimum to over 350 kbps. Clearly, the WPBAN-2 was not designed to communicate with a standard COTS radio. An exhaustive search of current COTS RF receivers or transceivers with typical OOK data rates from 1.2 kbps to 115 kbps, the radio used on the MEDIC was one of two devices offering 250 kbps OOK modulation data rates available at the time of design. This maximum available data rate has to do with the sensitivity of the receivers. The simulated sensitivity plot seen in Fig. 2.7 shows that as the data rate increases, so does the power requirement for successful detection of data. The sensitivity required for detection of the WPBAN-2 communications radio was not possible through a wireless link. Thus, the radios were connected with a SMA cable to test if the MEDIC Base communications radio was capable of receiving data from...
the WPBAN-2 given proper input power. Although the waveforms were recoverable in an asynchronous mode as seen in Fig. 5.1, the data could not be successfully extracted with any reasonable BER.

![Figure 5.1: Recovered data, the middle yellow trace, sent from the wirelessly powered WPBAN 2 in green.](image)

Concern was expressed that the RF power output was swamping the communications radio of the MEDIC Base, as the antennas are physically close together. However, the receive filter on the CC110L is capable of 37 dBm adjacent channel rejection and 50 dBm blocking for ± 2 MHz. As seen in Fig. 5.2 with a full power being transmitted on a 915 MHz carrier there is no significant signal received in the channel bandwidth for the communications radio.

There are a few considerations to make when attempting to solve this problem of communication quality. If the WPBAN is to go through another revision there are three things that could be done: a) Change the modulation scheme. Switching to another simple modulation scheme like FSK or phase-shift keying (PSK) may not be necessary and would likely only marginally improve communication performance. b) Reduce the
Figure 5.2: Cross talk from RF power on MEDIC communications radio.

WPBAN data rate. This is likely a good augmentation to the WPBAN in any case. The ability to set the device to have a lower data rate allows the WPBAN to also use a lower transmit power. The data rate should remain adjustable, scaling the data rate to the particular measurement or task that the WPBAN has will allow for task based power scaling. c) Augment the datum frame. Adding preamble bytes will mean the WPBAN will have to transmit for slightly longer, but will provide increased reception reliability. The number of preamble bytes should increase for higher data rates, this provides the receiver with more points to adjust gain and sample timing of the incoming bits. A sync byte will act as the end marker of the preamble, as by its nature, some of the preamble bits may be improperly received. To allow for maximum flexibility, the number of data bytes should not be fixed. It is likely the number of bytes will not need to exceed sixteen
(4 bits) for a single packet nor is it likely for the address to exceed the need for sixteen devices, thus packet length information can be included with the address information in a single byte. Finally, the WPBAN should report its RF power received signal strength indication (rssi) back to the MEDIC Base, so that the RF power or sampling/reporting rates can be adjusted as required. An example of this datum frame can be seen in Fig. 5.3 along with sample data and RF modulation with Manchester coding.

![Figure 5.3: Proposed improvement of the datum frame of the WPBAN, with sample data and RF modulation.](image)

For the next revision of the MEDIC Base, there is a new RF IC that is recently available and may have superior performance over the CC110L. The Silicon Labs Si4362 Receiver has much better sensitivity than previous COTS devices, allowing for even faster data rate, up to 500 kbps. The receiver sensitivity plots for the Si4362 can be seen in Fig. 5.4. The sensitivity for the WPBAN-2 data rate, 250 kbps, is $-100 \text{ dBm}$. This is more than three times less power required than by the CC110L, which has a 250 kbps sensitivity of $-95 \text{ dBm}$

5.2 Energy Harvesting

The output power of the MEDIC Base lacks fine increments for adjustment, this was seen in Table 4.1. This would be a desirable feature for future versions of the MEDIC Base as any power that it transmitted above the required link budget is wasted.

The only improvements that could be made to the energy harvest capabilities of the WPBAN are to further decouple it from the device’s digital communications data rate. The drift observed on the data rate from a change in received power was significant.
The MEDIC Node needs a new energy harvest system. The extreme cost ($35 per module in low volume) and subsequent failure of the Powercast P2110 module makes them very unattractive for future use. Given proper time for design and testing, there are several attractive discrete implementations such as in [28]. The MEDIC Node should also be able to accept future WPBAN ASIC prototypes on board. Using a microcontroller to program the scan chain through the SPI would greatly increase the mobility of the test set-up, which is currently programmed with LabVIEW.
Chapter 6: Conclusion

In this work we have demonstrated the feasibility of building a wirelessly powered autometric sensor. The MEDIC system is the most recent embedded platform to come from the VLSI group at Oregon State University and will allow us to easily test sensing abilities and applications for wirelessly powered sensors. Through its design, many improvements for future ASICs have been provided. Both the consumer market and clinical research arena would benefit greatly from the flexibility and adaptability of wirelessly powered autometric devices.
Bibliography


APPENDICES
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