Implementing a Low Noise, Low Power Portable EEG Sensor System

by

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Abstract

Electroencephalograms, also known as EEGs, are useful tools for monitoring brain activity, whether they are used for diagnosing serious conditions or simply monitoring stress levels. However, these systems usually require expensive equipment to have an acceptable level of precision and are cumbersome to use. The goal of this project is to create a wireless EEG sensor at a lower cost while maintaining accuracy. In order to maintain a high signal-to-noise ratio in these systems, the impedance measured at the front end of the system must be at least in the low gigaohms range. By using a FET Buffer with an input impedance in the multi-teraohm range, the cost and quality of the electrodes that the EEG system uses can be decreased while still achieving a sufficiently high signal-to-noise ratio. Through the use of analog filtering and amplification, our system accurately processes and displays brain waves within a bandwidth of .5 to 40 Hz while retaining signal quality and minimizing noise within the system. Benefits of this system include a high level of portability due to wireless capabilities, low cost, and simplicity of use.

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Chapter 1

Introduction

The inner workings of the human brain have always been a curiosity to scientists, one that has led to the creation of studies and technologies that aim to answer some of the many questions that this interest has sparked. It is amazing to think that simple changes in the electrical potential across the many billions of synapses between neurons within the brain can create complex thoughts, feelings, and synchronize the thousands of muscle movements and processes that occur within the human body.

Different areas of the brain control different functions within the body. The outermost layer of the brain, known as the neocortex, controls functions such as sensory perception, motor commands, spatial reasoning, thought, and language. As a person undergoes changes in their level of attentiveness or state of mind, there are slight changes in the voltage levels and frequency of the signals emanating from their neocortex. A method known as electroencephalography (EEG), which uses dozens of electrodes connected across the scalp, is used by scientists to accurately measure these signals and their fluctuations as they occur across the neocortex[1]. These signals can then be used for multiple clinical purposes, including diagnosing epilepsy, recording brain death, and measuring a person's level of stress. However, there are some drawbacks to modern EEG systems. Due to high levels of noise from outside sources, accurate readings of brain waves can take anywhere from twenty to forty minutes to record. Also, expensive electrodes and adhesive materials are usually necessary to achieve a connection that will accurately represent the desired brain signals[1].

The aim of our project is to create an EEG system that will not only function as well as today's modern electroencephalographs, but also remove some of these limitations. Through using a FET buffer with an extremely high input impedance level of around 20 TΩ[2], the system will maintain an adequate signal to noise ratio, allowing for our data to be well preserved. An instrumentation amplifier will be included within the circuitry to reject any common mode signals, and different filtering methods will be used to remove DC offset, extraneous noise, and any unwanted frequencies outside of the bandwidth of the desired signal. The high quality of these components and the overall system will allow for observation of a weaker input signal, therefore permitting the use of inexpensive electrodes and a more convenient connection method. It is also expected that a shorter time and fewer electrodes will be needed in order to obtain a good reading.

Through extensive research into prior methodologies on the construction and use of electroencephalographs and analyzing each system's strengths and weaknesses[12][24][26], it was possible to create a design for our system that will be able to exceed the performance of existing devices. After the construction of our prototype, testing procedures that measured the test subjects' level of attentiveness were completed to prove the functionality of the device.

1.1 Report Overview

Each of the steps and considerations that occurred during this project are discussed at length in this report. In Chapter 2, we will discuss the beginning steps of the project decision making process, including the different criteria that were considered when choosing a project, the different project ideas that were formulated during the brainstorming process, and an initial decision on which ideas should be further pursued based on a pass or fail table of the criteria under consideration. The final decision making process is detailed in Chapter 3, where the three top project ideas are discussed, a decision matrix is shown with different weights for each criteria to narrow down the decision to two projects, and a reevaluation of these weights is made to make the final project decision.

In Chapter 4, the preliminary research that was done for the EEG system project is detailed, including the frequencies of brain waves, typical circuitry for an EEG, different errors that prior designs have faced, requirements, and specifications for our system. This chapter also includes a system block diagram based on these specifications and requirements. Chapter 5 highlights the different factors that were taken into account when creating the design of our system, such as signal filtering and different interface decisions. In Chapter 6, the testing and simulation of the system is detailed, including different altercations that were made to the system based on these tests.

Chapter 2

Preliminary Project Decision

In this chapter, each of the criteria that were considered during the project decision process will be discussed in detail. The project ideas that were created during the brainstorming process will then be explained, followed by the feasibility analysis for each of the ideas.

2.1 Decision Criteria

The criteria used to choose which idea would be developed for this project include time constraints, visibility, winnability, learning curve, innovation, ADI part integration, and cost. Following is a brief discussion of each of these factors.

Time Constraints An important factor to consider for this project is the amount of time needed to complete it. With only 14 weeks to complete the project, have a written report, and produce both an article and video for visibility purposes, the amount of work required to have a completed working prototype needed to be reasonable. Some of the other constraints were used to gauge how long each project would take, including the learning curve for the project team and visibility factors.

Visibility In addition to the final report, another outcome of the MQP project will be an article. This article will be written to highlight the architecture and design of the project for Analog Devices Inc. Also, a video will be created for posting on Youtube that will give an overview of the project, some insight to the MQP process, and demonstrate the final prototype. Both the article and the video will allow for maximum visibility to the public. It is important that the article and the YouTube video show the crucial design components of the project. They will also need to convey some sort of “wow factor” in order to get the audience interested in the project. If the project has some potential to be patented, this is also of interest.

Winnability Winning the ECE Department’s MQP Award is another priority. This also relates to the visibility aspect, seeing as a project with more of a “wow factor” may have a better chance of winning the award. At the same time, depending on the project, the time constraints may hinder any additional improvements to the design that would give the project a significant edge over the competition.

Learning Curve It is important that there be a balanced combination of using knowledge from previous courses and learning new topics to implement into the final design. The project idea should not involve topics that would be too difficult to learn in the time allotted, but should also not be so easy that the project will be easily completed in a few weeks. The project idea needs to present a challenge that can be overcome within a reasonable amount of time. A project that uses a certain type of technology that isn't very familiar to the group might be too difficult, but it is important that some risks are taken to design something new and noteworthy.

Innovation An important consideration when choosing a project is how innovative the idea is. This holds true for the project not only if it in an entirely new idea, but also how much the finished prototype might be improving on an existing product, including how novel the technology was that was used to make these improvements. It is also important that the project is interesting, in order to spark the motivation of the team and any potential future viewers of the project. However, it is important to make sure that the project isn't so innovative that it encroaches on the amount of time available to complete the project and implement everything that is desired.

ADI Part Integration Analog Devices Inc. is the primary sponsor for this

MQP project. They will be providing the parts that we choose, and hopefully the final product will be able to highlight the performance of these parts. The final decision on the project should take into account how well the ADI components can be showcased and possibly how many of their products can be integrated into the design. While most of the design will be created by the MQP team, it is unreasonable to assume that there would be enough time to design the entire system down to the transistor level. Therefore, an appropriate balance between integrating a few ADI parts and having enough original design is desirable.

Cost The final constraint to consider will be the cost of the finished product. While ADI will be covering the project costs, a final product that is cost competitive with the prior art already in the market is also desirable. Due to the fact that a finalized block diagram will not be completed until later in the project, a final list of the parts can't be used to determine the cost. When deciding how competitive each of the project ideas will be, a rough estimate based on the cost of the prior art was used depending on the type of technology.

2.2 Project Ideas

During the initial stages of formulating different ideas that might be of consideration for this project, brainstorming methods were used that allowed for the creation of novel project ideas with varying levels of innovation and feasibility. While brainstorming, it was important to keep an open mind to ideas that might not be as conventional as previous projects, as this broad perspective allows for the consideration of ideas that might end up having a higher success rate than projects that stay within a certain comfort zone. By maintaining this open point of view, a variety of different project ideas were formulated, which are listed below. Each of these different ideas has varying levels of the criteria that were explained in the previous section. While this variation is initially acceptable in terms of brainstorming, these ideas will then be narrowed down based on how well they fit each of the categories, taking into consideration the importance of certain factors over others.

2.2.1 Brain Wave Sensor

A small brain wave sensor can be created highlighting the performance of an ADI FET buffer and instrumentation amplifier. Devices known as electroencephalographs, or EEGs, have been used to record brain signals so that they can be accurately interpreted[3]. This project would try to improve on what methods and devices are already in existence, whether it be reducing the number of electrodes on the subject, reducing the size of the equipment needed in order to make it portable, or reducing noise seen at the output of the circuits so that a cleaner signal can be recorded. Not only would this project look impressive in a YouTube demo and scholarly article, it would be reasonable to complete this in the required time frame. There would be a sufficient amount of design, and it would have a chance to win an MQP award.

2.2.2 Helmet Sensor

A system would be created that integrates a sensor into a helmet design to sense risk of concussion or other head injuries. The main application of this system would be for motorcycle helmets. Often emergency responders are not properly trained in the correct method of motorcycle helmet removal when serious head injury has been sustained. This often can lead to further injury for the user. Either an accelerometer or surface mount pressure resistors (or a combination of both) would be implemented into this design. An accelerometer[4] on the front end would be triggered when rapid acceleration/deceleration occurs. Pressure sensors interfaced onto the outside of the helmet would also be able to measure the intensity of an impact to the helmet. For this system, it would be possible to incorporate a few different components from ADI, starting with the built-in accelerometer. Some drawbacks of this system would be possible problems in its accuracy, marketability, aesthetics and finding an efficient way of powering the device. This causes time constraint issues, and while it would be possible to complete a system within the time allotted, it would be difficult to get the system to the point of properly highlighting the capabilities that would actually make this project idea innovative and better options than prior art.

2.2.3 Blood Pressure Sensor

The blood pressure sensor topic is interesting due to the potential for innovation in the method of measuring the blood pressure. Blood pressure is typically measured using a sphygmomanometer. A cuff is wrapped around the patient's arm, and then air is pumped into the cuff to increase the pressure. This restricts blood flow [5], and the movement of blood past this cuff helps to make an accurate blood pressure reading. The issue with this method is that it takes a few minutes to take a reading, and often causes minor discomfort. In addition, the user must remain relatively still while measurements are taking place. By designing a less complicated method of measuring blood pressure, users who have either high or low blood pressure can more easily monitor fluctuations based on daily activities. If the device were to be as small as a watch, the user would likely get used to the additional weight and would be able to easily track their progress towards raising or lowering their blood pressure. While this project idea would allow for a high level of innovation, that same level of ingenuity could cause drawbacks in the progression of the project. It is also possible that the learning curve in regards to the biological aspects of the project might take a significant amount of time to overcome.

2.2.4 AMR Sensor

A demo board would be constructed for ADI's new AMR sensor to show off its performance and possible applications. These Anisotropic Magnetoresistance (AMR) sensors are directional; the magnetoresistance of the sensor is based upon the angle between the current and the magnetic force [7]. The greater the angle, the smaller the resistance. AMR sensors are commonly used for electronic compass applications in Cell Phones, GPS, power steering, cars with navigation capabilities, and vehicle detection [6]. ADI has created this part already, so the design would revolve around a bridge configuration with the AMR being the variable resistor and using the voltage differential to create the demo. Since the project focus is to create a demo board, there is a lot of freedom to do interesting things. In addition, the demo board would work well with the goal of creating YouTube videos for the process. However, a concern is that a board might demo multiple things and that could cause the scope of the project to exceed what can be completed within the time constraints. In addition, this technology is very expensive and it might not be beneficial for ADI to have a board for this product. The amount of design involved would end up being very little, and there would be a very low chance at winning the MQP award.

2.2.5 Radar System

Using a pre-designed ADI chip[8], a demo board would be designed and implemented to calculate the distance between itself and another object that it is pointed at. This will demonstrate how the system operates if it were integrated into a car; the sensor would detect if a person was about to crash/hit something and would stop the car automatically. This technology could also be implemented to autonomous devices such as robots, unmanned vehicles, etc. This project would specifically highlight the radar system designed by ADI. While this project might be challenging due to its learning curve, the results from the project and the demo have the potential to look really impressive and give us a good chance at winning the MQP award. One of the major drawbacks is the required RF background that is needed. This would be a very complicated system and an antenna would need to be designed in order for the system to function. There is a high likelihood that this project might not be feasible to complete in the time frame allotted. Additionally, all of the components for the system except for the antenna are already designed, so the only design work would be for the antenna. Finally, some sort of interface between the system and a computer would have to be created so that calculations can be made on the data collected.

2.2.6 Amplifier-Converter Integration

Using analog-to-digital converter specifications, it is possible to determine one amplifier that would work optimally with that specific converter. Usually in a signal chain, either the amplifier or the converter is already chosen, so all that is left is to choose the other component that fits prior specifications the best. Further analysis would be done on the pair, possibly by showing how one combination would work better than another, etc. This could be done by selecting different performance specifications and seeing which combinations work best with certain applications. The best-matched pair could then be applied into a signal chain on a demo board. The spreadsheet of matched ADI converters and amplifiers is already made, so a few pairs from this list would be chosen to analyze further. A general signal processing chain is shown in Figure 2.1. There is typically some analog front-end that includes an amplifier to amplify the signal along with any necessary filtering. The signal would then go into an ADC before any further processing. A drawback of this project idea is that it would require little innovation or freedom of design.

Since the spreadsheet of the best working pairs is already created, no analysis would be done as to why that pair works better than another combination. The only design would come from the signal chain application, and this will demonstrate the performance of the ADI parts that would be appealing in a trade show. While this might be the easiest to complete in

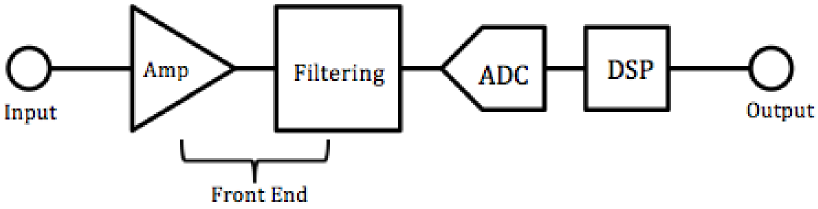


Figure 2.1: General Signal Chain Diagram

the two terms available, it is not likely to be a good candidate to win the MQP award. This project would be more analysis than design, which is not desirable.

2.2.7 Conserving Power with Fast, High Precision SAR ADCs [9]

This project would highlight some of ADI's successive-approximation-register (SAR) ADCs that have sleep mode or low power mode functions. With the drive for everything to be less energy consuming, many ADCs, and now amplifiers, are being designed so that they will power themselves down when they are not being used. A demo board could be created that runs off of very little power but will still be able to do signal processing. This project idea would be something that Analog Devices might be able to demonstrate at a trade show. Similar to the amplifier-converter project, all of the design work has been done already. A lot of time would need to be put into understanding how the components shut down when they are not in use in order to maximize their capabilities in whatever signal chain is created.

2.2.8 Stride Improvement

This device would be for runners, hikers, people with inefficient strides, or people who recently acquired a prosthetic limb and would like a more natural stride or movement. It would use multiple sensors placed on the skin that would show if the movement of the person is optimal for their body's performance. The device could be placed on just the legs, just the arms, or all four limbs. It would help improve efficiency by allowing the user to find out if the strides are too small, too large, or if they swing their limbs out too much. More interestingly, the device might be very useful for users who recently have obtained a prosthetic limb. It could help a user learn about any overcompensation done and it could help make their walking movement more natural.

The accelerometer used in this project would be a three axis accelerometer. The accelerometer would output signals to a microcontroller, where the acquired information would be stored. Measurements would have to be taken at a fast rate, because the length of time it takes for a single stride to be completed is relatively short. Around 10-20 different measurements would be taken per stride to get an idea of where in the stride inefficiencies exist. More measurements would allow for greater accuracy, but would increase the amount of storage necessary, as well as the data acquisition rate of the system, which might require faster and more expensive accelerometers. The device would later be connected to a computer, where a program would upload previously taken data and plot the points to show stride efficiency. Plotting of these points could likely be done using MATLAB or a similar program. An important aspect of this project would be creating a program that is easy to use. If possible the plotting program would launch automatically whenever the device is connected to the computer. A difficulty that this project idea would cause is creating an algorithm that would calculate how efficient the user's steps are, and discovering different anomalies in strides that should be taken into account.

This device could be difficult to design because quite a bit of time would have to be spent designing the program, as stride measurements are not already in the group's area of expertise. In addition, time would have to be spent determining a way to calculate the most efficient stride for any given person. That alone could be a major qualifying project in itself. Finally, the programming of the microcontroller would also take a lot of time.

This device would probably have to track location in a 3D space over time. If possible it would come alongside a program that would help the user visualize what kinds of movements they should reduce and what they should increase. If possible, the movements of the user could be replicated in a 3D model using Maya[17], so that they can have a visual of their strides. There would probably be a lot of programming associated with this device, but it would be useful for many people around the world. This would likely be a great project for a group with a Computer Science major or a Biomedical Engineering major on the team.

2.2.9 Climbing Fall Notification

This design would use an accelerometer to detect a fall and would send a signal that would notify the belayer. When climbing outdoors, especially on longer or more difficult routes, it is possible for the lead climber to climb outside the range of vision of the belayer or out of the range where verbal communication can be made. In most cases, this type of communication is not necessary because an experienced belayer can feel when a climber is clipping in, climbing, or falling. In other cases, this device would be very useful. Due to the way they are set, some routes will have a large amount of rope drag [18]. Some of this rope drag can easily be avoided by carefully selecting gear, but not always. When this happens along with lack of visible cues or verbal communication from the climber, a more dangerous situation can occur. If a climber falls and the belayer cannot feel additional tension in the rope, and cannot see nor hear the climber, there is not really a good way for the belayer to know if the climber is climbing, has fallen, is resting, or is even conscious. Some climbing teams will use walkie talkies or even cell phones for communication, but walkie talkies can be heavy and communication can be tricky if multiple groups are using the same channel. Cell phones can work well if there is a signal and if they are not dropped.

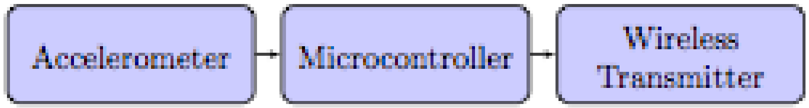


Figure 2.2: Climber Block Diagram

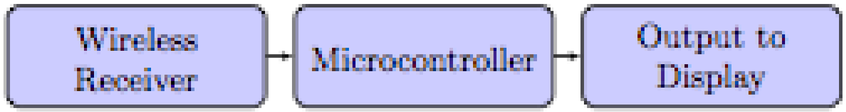


Figure 2.3: Belayer Block Diagram

Additional features would include buttons to send signals that the climber would like to be lowered or would like to continue climbing. A safety feature would be an alarm on the climber's end that will be activated in the event of a fall. If the climber does not deactivate the alarm within a certain time, the system will send a signal indicating that the climber may be unconscious which will allow the belayer to make an informed decision on what to do next. Some drawbacks include that a fall might be detected when a climber makes a dynamic move. Another drawback is that the device will not tell a belayer what to do, it will only help them make a better decision. This drawback can also be a benefit because it still requires climbers to have a knowledge of climbing rather than relying solely on technology.

2.2.10 Energy Absorption Tiles

Tiles would be created using either piezoelectric or kinetic technologies to absorb energy from the pressure exerted on the tiles by passersby. Piezoelectric materials have the ability to produce electrical charges when deformation occurs across the material[19], and kinetic tiles would have small springs underneath them that would absorb the force that is put on the tiles. These tiles would be installed around campus and would use their energy to help power academic buildings. After researching this idea, it was found that kinetic tiles have been installed in certain parts of London[20], and that piezoelectric tiles have been implemented at colleges such as Rutgers university[21]. A small area of piezoelectric tiles ended up costing over fifty thousand dollars to install on this campus, leading to the conclusion that it would be too expensive to go forward with this project idea. This project would also require more in-depth and time consuming research on possible methods to create an innovative and possibly more cost-effective tile design.

2.2.11 Ball Sensor

The incorporation of an accelerometer into a baseball or similar sized ball would allow for the user to determine how fast the ball has been thrown and how far it has traveled. The sensor would wirelessly send this data to a smart phone app or similar external interface. These balls could be used by professors to teach students about the forces of gravity, especially if the accelerometer is programmed to output the vertical and horizontal magnitude vectors over time.

One difficulty that this project poses is being able to find a wireless system that will integrate easily into the ball itself without altering the weight or aerodynamics of the ball. For example, an antenna would obviously not be a feasible option for this project, and a heavy sensor would affect the performance of balls in certain sports such as tennis or golf. Powering the system would also be an issue for this project, as a charging interface would again affect the mechanics of the ball.

2.2.12 Muscle Hyperextension Sensor

Sensors placed on the skin would register if the muscles underneath are being put under an excessive amount of strain, and would notify the user if damage is likely to occur. This sensor could be used by athletes, physical trainers, and those trying to stretch their muscles without causing any lasting damage to their body. This project idea has a large knowledge curve, and would require a great deal of time to be spent learning about the biological aspects of the project, including how far muscles can be extended without causing any damage. It would also be difficult to create a sensor that would be able to interface with muscles in the body, causing yet another feasibility issue for this project.

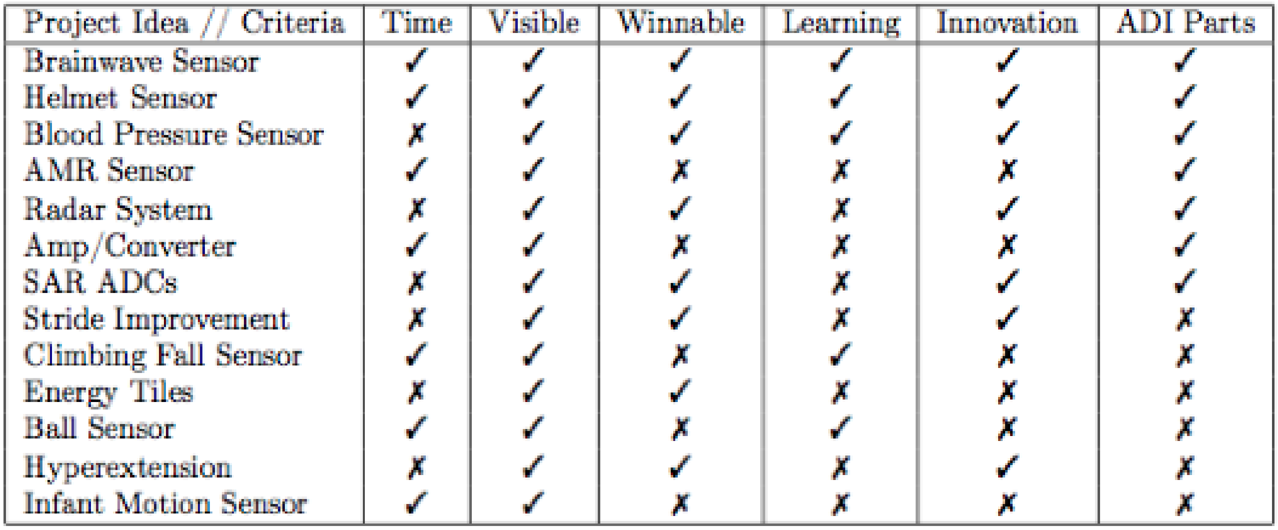
2.2.13 Infant Motion Detector

A device placed on the infant's clothing would measure if the baby was being shaken, and would then alert authorities. This project idea raises multiple issues, including the possibility of the child choking on the sensor, or the possibility that abusive parents likely would not go out of their way to purchase the item. The only way for the device to be successful would be if it were a regulatory safety device that all parents must have, which is difficult to implement. There also is not much room for innovation with this project idea.

2.3 Initial Decision

Each of these project ideas was considered in detail in regards to each of the seven criteria that were described in Section 2.1. These considerations were then translated into either a pass or fail for each criterion, which allowed for the creation of Table 2.4.

From this table, the number of project ideas that could potentially be pursued was significantly narrowed down. If an idea was not expected to have the capability of being completed within the allotted time for this project, it was almost always eliminated. Other project ideas were eliminated from the decision process because they failed multiple other criteria that are essential to the success of this project, such as winnability and learning curve. The only two potential projects that passed every criterion were the brain wave sensor and helmet sensor. The blood pressure sensor passed all but one of the criteria, with the only one that it failed being in the area of time constraint. It was decided that this factor could possibly be overcome, and that because it did indeed pass the rest of the criteria, it should be allowed to continue onto the next step of the decision process.



Project Idea // Criteria	Time	Visible	Winnable	Learning	Innovation	ADI Parts
Brainwave Sensor	✓	✓	✓	✓	✓	✓
Helmet Sensor	✓	✓	✓	✓	✓	✓
Blood Pressure Sensor	✗	✓	✓	✓	✓	✓
AMR Sensor	✓	✓	✗	✗	✗	✓
Radar System	✗	✓	✓	✗	✓	✓
Amp/Converter	✓	✓	✗	✗	✗	✓
SAR ADCs	✗	✓	✓	✗	✓	✓
Stride Improvement	✗	✓	✓	✗	✓	✗
Climbing Fall Sensor	✓	✓	✗	✓	✗	✗
Energy Tiles	✗	✓	✓	✗	✗	✗
Ball Sensor	✓	✓	✗	✓	✗	✗
Hyperextension	✗	✓	✓	✗	✓	✗
Infant Motion Sensor	✓	✓	✗	✗	✗	✗

Figure 2.4: Initial Pass or Fail Decision

Chapter 3

Final Project Decision

Out of all of the project ideas, the top three were chosen by process of elimination. Only three of the potential thirteen project ideas passed six or all of the seven criteria that were used in the first step of the decision process. These project ideas include a brain wave sensor, a helmet/concussion sensor, and a blood pressure sensor. To get a further understanding of the feasibility of each of these projects and delve into how well each project idea fulfills the given criteria, these ideas were researched and analyzed further. In the following section, more detailed descriptions of each of the project ideas, the potential issues that they pose, and the innovative capabilities of each will be explored.

3.1 Brain Wave Sensor

This system would be created to detect and read brain waves using small electrodes, ADI's FET buffer and their instrumentation amplifier (in-amp). This project would be the proof-of-concept of a system that would eventually be able to fit completely inside the electrode encasing.

There are multiple methods for recording signals from the brain that are used mostly in medical applications. An EEG, or an Electroencephalogram is used to see the electrical activity or measure electrical potentials of neuron activity inside of a brain[3]. The test consists of putting many electrodes around a person's head and having them lie still for a certain amount of time in order to get an accurate reading. The output reading of the test appears as various waves, as shown in 3.2, which would be used to try to diagnose any problems the person might have. It is commonly used to diagnose epilepsy and other serious health conditions. MEG, or Magnetoencephalography, is similar to EEG except it maps brain activity by recording small magnetic fields produced by the electrical currents in the brain[10]. Using this kind of measurement, it is easier to determine which part of the brain is active at a given time.

There are already many existing methods for this technology. Not only have these tests been available for many years, there are now several brain sensors on the market. NeuroSky[11] makes a sensor that is supposed to detect a subject's attention and relaxation levels and show the brain signals on another device through Bluetooth. Another product called Melon[12] is similar to the NeuroSky product and reports the data that the sensor reads to an app on a smartphone. This gives a numerical score for attention levels and tries to distinguish which activities make a person more alert.

It is important that this project is not just a replication of current methods, so a novel idea will need to be implemented. The major goal would be to implement a system with a single sensor or electrode pair, while also being able to get fast, accurate readings. The preferred block diagram is shown in Figure 3.1.

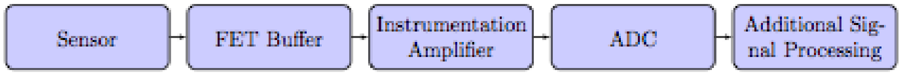


Figure 3.1: Brain Wave Sensor Block Diagram

This project would highlight one of ADI's FET buffers, which has a very high input impedance and low input bias current. This could potentially eliminate the need of an expensive skin contact since the input impedance is in the $20T\Omega$ range. In addition, the low power AD8421 in-amp would be used to amplify the signal being read. Depending on the power needs of the final system, it may be possible to create a device that is all on one small unit that will come in contact with the skin. The amplified signal from the in-amp would then go to an ADC and then onto additional processing before the final signal is displayed.

This project would be different from the products that are already on the

Figure 3.2: Raw data from an EEG reading [13]

market in that it aims to be an all-inclusive sensor and signal processing package. If the system is also rechargeable and can eventually be configured to transmit the data it reads wirelessly, this would significantly improve on previous art.

Finally, there are some concerns regarding this project. Most electronic components aren't supposed to be used as a life saving device without permission from the manufacturers. If this project is considered as a life saving device, there will be the extra hurdle to overcome. Noise may also be an issue with the measurements, but this might be mitigated with the FET buffer.

Overall, the project has enough of a learning curve that a lot of knowledge will be learned and it can be completed in the amount of time that is required. An article and YouTube video will be able to highlight the design aspects of the project very well, and there is a good chance that this project could win the MQP award.

3.2 Helmet Sensor

This would be a system that would be integrated into a helmet design in order to sense risk of concussion or other head injuries. The main application of this system in regards to our project would be for motorcycle helmets. Often emergency responders are not properly trained in the correct methods of motorcycle helmet removal when serious head injury has been sustained. This could lead to further injury for the user. Either an accelerometer or surface mount pressure resistors (or a combination of both) would be implemented in our design. This system could also be implemented in sporting helmets, and resulting data from the finished product could possibly be used in the future for the analysis of helmet damage for future strengthening of helmet designs, or for finding correlations with certain impact points and risk of head injury. Currently, there are helmet sensors on the market that are stand-alone and can attach to the helmet. The device known as ICEdot has an accelerometer that detects whether or not a cyclist has been in a crash, and then uses low power Bluetooth to send a signal to the user's cell phone that would call their emergency contacts.

The block diagram below shows the basic path that a signal would take through this system. An accelerometer on the front end would be triggered when rapid acceleration/deceleration occurs, which would cover the instance of a user having neck injuries caused by whiplash without actually hitting their head. Pressure sensors interfaced onto the outside of the helmet would also be able to measure the intensity of an impact to the helmet. One type of sensor that would be applicable in this situation are force sensitive strip resistors[22]. These strip resistors would be able to cover more lateral distance, which could be placed in a lattice pattern across the helmet for multiple data points. When a force is applied to these resistors, the resistance lowers. After reaching a threshold value, it would trigger the rest of the signal chain. Moderate impact would call for a "possible concussion" notification, while a severe impact would send a "serious head injury possible" notification. These notifications could possibly be in the form of LEDs on both the outside of the helmet for emergency responders and the inside of the helmet or wrist for the user.

Figure 3.3: Block Diagram of Helmet Sensor

It would be important for the device to continue outputting the signal in the case of the sensors receiving so much damage that they break. Once the helmet receives the signal, it would stay on unless reset. However, it is currently assumed that the amount of pressure needed to break the sensors would also mean that the helmet has sustained significant damage and should be replaced anyways. There are multiple options for powering the device. Piezoelectric scavengers could be used to absorb the shock that a motorcyclist undergoes when driving, and transfer this energy into power for the device. Small solar panels could be implemented onto the shell of the helmet, though they would have to be able to undergo a certain amount of force. Batteries could also be used, as long as they are both safe to interface into the helmet and are efficient. To save power, the clip of the chinstrap of the helmet could be the switch for the device. When the helmet is on and secured to the head, the device would turn on. When the helmet comes off, it will save power within the system to allow batteries to last longer.

While ICEdot is stand-alone, this new device would be integrated directly into the helmet, which would hopefully encourage more motorcyclists to purchase it, as they don't have to go significantly out of their way to attain the sensor. Another similar product is a football helmet designed at MIT with several accelerometers within it. However, the output of data isn't displayed on the helmet, but wirelessly sent to a computer that can only be a max of 20 yards away. These helmets are costly, at one thousand dollars apiece, where the new design aims to be significantly cheaper, and only heighten the price of a motorcycle helmet by a marginal amount, one that is worth it for the safety of the user. Another drawback of this existing helmet is that it assumes that a concussion would occur at 98gs, when it has been shown that they usually occur anywhere from 50gs upwards [23].

For this system, it would be possible to incorporate a few different components from ADI, starting with the built in accelerometer. We could also use an in-amp to measure the differential voltages of a wheatstone bridge that includes the resistances of the force sensors, and use this differential voltage to trigger either an ADC or another device depending on what threshold level it has crossed. Some drawbacks of this system would be possible problems in its accuracy, marketability, aesthetics (depending on what kind of integration method are used for the sensors), and finding an efficient way of powering the device. This causes time constraint issues, and while it would be possible to complete a system within the time allotted, it would be difficult to get the system to the point of properly highlighting the capabilities that would actually make this project idea innovative and better options than prior art.

3.3 Blood Pressure Sensor

According to the Centers for Disease Control and Prevention website, 31% of Americans have high blood pressure. This represents 67 million people. High blood pressure also puts people at risk for other life threatening problems. "69% of people who have a first heart attack, 77% of people who have a first stroke, and 74% of people with chronic heart failure have high blood pressure. High blood pressure is also a major risk factor for kidney disease" [14]. Because high blood pressure is such a large problem in the United States, it is worthwhile to create a system that can help a person better monitor their blood pressure and make better lifestyle decisions.

Most people who have been to the doctor are familiar with the machines used to measure blood pressure. The blood pressure cuff, also called a sphygmomanometer, is used by pumping air into the cuff to increase pressure and then measuring the pressure using a manometer. Blood pressure cuffs are accurate but they are bulky and require air to be pumped into them. This aspect restricts them from being portable. A pulse oximeter is commonly used to determine the amount of oxygen in blood using a bright light and a light sensor. A pulse oximeter must determine the volume of blood and in doing so, the blood pressure can be determined [16]. The final method is based on pulse transit time. This method is done by determining how long it takes for a pulse to get from one spot to another. Because the pulse wave velocity is related to the pressure, the difference between the diastolic and systolic pressure can be determined [15].

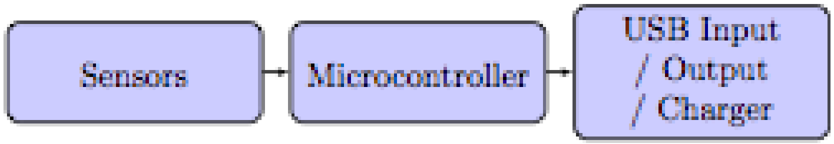


Figure 3.4: Blood Pressure Sensor System Diagram

The sensor used would be a pulse oximeter designed to be placed on the wrist. This would monitor blood pressure and heart rate. The measurements would be sent to a microcontroller that will output the display in real time and also store the measurements taken at regular time intervals so that changes in blood pressure could be monitored over a long period of time. In order to view the progress, users would connect the device to their computer and a program would show them how they have progressed over time.

With both potential methods, it is uncertain that a reasonable device could be designed. Using a method of pulse oximetry relies on the anatomy of a finger and the locations of the arteries within the fingers. If the device were to be comfortably worn on a finger on a daily basis, it probably could be no larger than a class ring or a fancy engagement ring. If the device were to be worn on the wrist, the methods used with pulse oximetry might not work nearly as well with the location depending on where the arteries in the wrist occur. Another possible location to wear the device could be on the back of the hand since it may not move nearly as much as other parts of the lower arm area. There is a known relation between pulse transit time and the difference between systolic and diastolic pressure, but the actual pressures may not be able to be determined.

3.4 Decision Matrix

After conducting research on each of the potential projects, eliminating some of the less feasible project ideas, and deciding that the brain, helmet, and blood pressure sensors adequately met the most essential requirements for the project, a decision matrix was created to further narrow down the selection in a more structured and in depth way[25]. This decision matrix rates the three leading project ideas on their strength in each of the criteria that were deemed important for the project. In order to make this matrix as accurate as possible, different factors such as weights for each criterion and rating scales were established.

3.4.1 Explanation of Weights and Scoring System

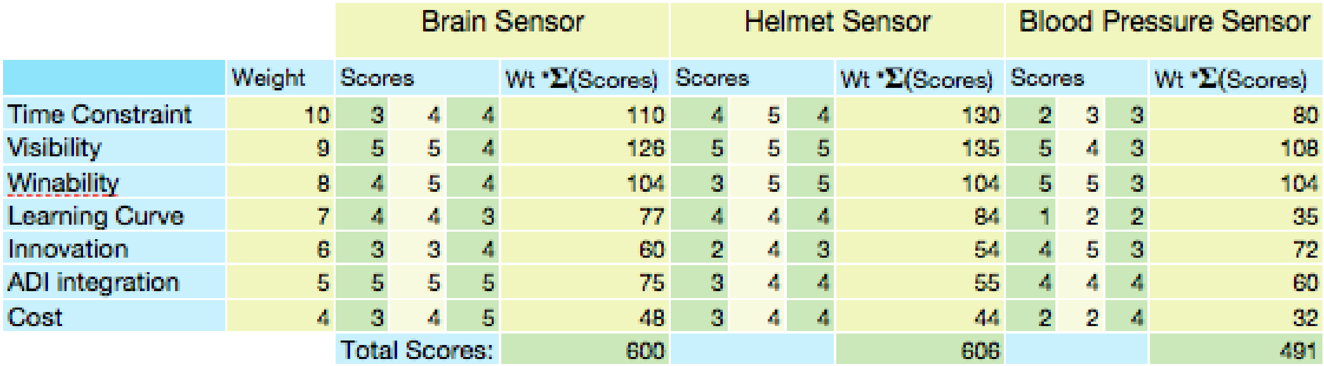
While there are seven different criteria that are important to consider for each project idea, some of the criteria are considered more crucial than others in terms of choosing a project. To account for this, multiplying weights were given to each of the criteria in order to properly skew the total scores. It was decided that time constraint is the foremost concern for the projects, as it is essential to have the project completed by the end of the semester. Because of this, time constraint was given the highest multiplying factor, with a weight of 10. After this came visibility with a weight of 9, because having a YouTube video, article, or patent by the end of the project that others can see and recognize the team for is also generally more beneficial than the rest of the criteria, though less important than being able to complete the project. The weights of the criteria continue to lessen with winnability weighted as an 8, learning curve as a 7, innovation as a 6, and integration as a 5, with the same structure of having the higher weighted criteria more important than lesser weighted criteria. The lowest weight factor was given to cost, with a weight of 4, which was not considered to have as great of an effect on the overall project when compared to the other criteria.

The scoring system for each of the criteria was on a scale from one to five. Some of these representative values are easy to define, such as a one for cost representing the most expensive project idea and a five being the most affordable option. This linear nature also holds true for time constraint, with least to most likely to be completed by the end of the semester, as well as winnability with least to most likely to win an MQP award, and innovation from least to most innovative project idea. For visibility, a 3 would represent the ability to either produce a youtube video or article, while a 1 would represent neither being possible to complete and a 5 would mean that both could be completed. For learning curve, a 5 would be the optimal situation of having an intellectual challenge and learning new things while not going overboard with what the knowledge base for the project is expected to be. On the other hand, a rating of 1 could represent either having way too much of a challenge or barely any challenge or learning component required. Lastly, the integration of some ADI parts would receive a score of a 3, while the chance of highlighting and integrating multiple parts from ADI would receive one of the higher values.

3.4.2 Matrix Evaluation Process

After defining how the scoring system and weights for the decision matrix would work, each group member scored the three project ideas in the categories mentioned. These values were then discussed to eliminate any major discrepancies in scoring from one person to the next. It was determined that the scores for each criterion would be more accurately represented if the opinion of all three group members were to be reflected within the overall scores. These three values were then summed and multiplied by the corresponding weight factors. The values for the seven criteria were then summed together, resulting in totals for each project idea. The results of the first pass of the decision matrix are shown in Figure 3.5.

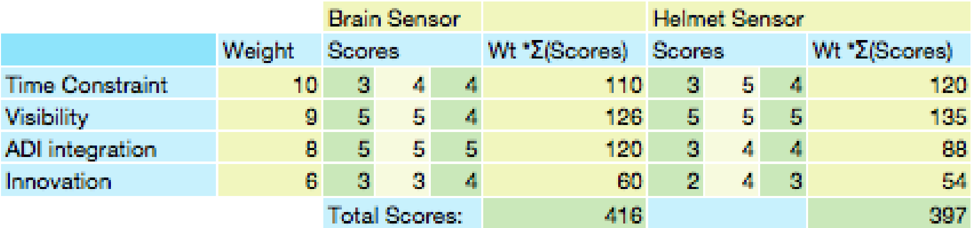
After completing the first decision matrix, it could easily be seen that the brain and helmet sensors were the two forerunners of the three projects, while the blood pressure sensor had a considerably lower score. Because of this, the blood pressure sensor was eliminated from the decision process when completing the second pass of the decision matrix. Due to the very similar scores of the two remaining project ideas, it was necessary to change some of the factors within the decision matrix to allow for a more obvious discrepancy when deciding what would be the final project idea. After discussion amongst the group, it was concluded that the weighting factor of ADI integration should be reconsidered to be much more significant than was originally planned, as this integration of ADI parts would lead to a higher



		Brain Sensor				Helmet Sensor				Blood Pressure Sensor			
	Weight	Scores			Wt *Σ(Scores)	Scores			Wt *Σ(Scores)	Scores			Wt *Σ(Scores)
Time Constraint	10	3	4	4	110	4	5	4	130	2	3	3	80
Visibility	9	5	5	4	126	5	5	5	135	5	4	3	108
Winability	8	4	5	4	104	3	5	5	104	5	5	3	104
Learning Curve	7	4	4	3	77	4	4	4	84	1	2	2	35
Innovation	6	3	3	4	60	2	4	3	54	4	5	3	72
ADI integration	5	5	5	5	75	3	4	4	55	4	4	4	60
Cost	4	3	4	5	48	3	4	4	44	2	2	4	32
		Total Scores:			600				606				491

Figure 3.5: First Version of Decision Matrix

success rate within other criteria such as visibility. Cost, learning curve, and winnability were also removed from the second decision matrix. Both project ideas had scored alike in terms of winnability, while the learning curve and cost factors were reevaluated to be less important in the overall selection of the project idea. This tweaking of weights and scores resulted in the decision matrix shown in Figure 3.6.



		Brain Sensor				Helmet Sensor			
	Weight	Scores			Wt *Σ(Scores)	Scores			Wt *Σ(Scores)
Time Constraint	10	3	4	4	110	3	5	4	120
Visibility	9	5	5	4	126	5	5	5	135
ADI integration	8	5	5	5	120	3	4	4	88
Innovation	6	3	3	4	60	2	4	3	54
		Total Scores:			416				397

Figure 3.6: Second Version of Decision Matrix

It can be seen from the decision matrix in Figure 3.6 that the brain sensor took a moderate lead in comparison to the helmet sensor after making these adjustments. While it scored slightly lower in terms of time constraint, it is still a viable option for a project that can be completed within the two term time limit. The brain sensor would allow for a high visibility, would be interesting to demo, and sufficiently highlight some of ADI's technologies, including an instrumentation amplifier and high impedance FET buffer. The level of innovation that this project idea poses is also high, without being too much of a learning curve for the members of the project team. On the other hand, the helmet sensor project idea was not nearly as innovative and did not have the capability of integrating many ADI components. Because of these factors, it was decided that the brain sensor project would be the most beneficial to move forward with.

Chapter 4

Initial Research and Specifications

In order to finalize the design for a brain wave sensor, some more information was needed in order to make the correct specifications. To achieve a fuller understanding of the measurement, research was done on the different frequencies of brain waves, the different circuits that are currently in use to measure these waves and finally, what the major issues and error sources are for these circuits.

4.1 Brain Wave Frequencies

Figure 4.1 shows the different types of brain waves that can be detected and what state of mind they correspond to. Beta waves range from about 13-30Hz, and occur when a person is in an attentive and active state of mind. Waves known as high beta waves exist from approximately 30-35Hz, and are triggered when a person is in a stressed or anxious state. Alpha waves range 8-14Hz, and correspond to a person who is relaxed. Theta waves occur between 4-8Hz, and can represent either a light sleep or meditative state if the subject is awake. Finally, delta waves occur from 0-4Hz, and are seen when a person is in a deep sleep. Gamma waves, which are from 30-100Hz, correspond to sensory processing of sound and sight [24]. However, these signals occur deep within the brain, and are not typically seen by an electroencephalograph system. Knowing the frequencies of these different kinds of brain waves will lead to a bandwidth specification for the sensor. This knowledge will also allow for focus on certain types of waves when conducting tests with our system, and will help to test the accuracy of our final prototype by seeing if our test subjects are in the same state of mind as the corresponding frequency at the output of the sensor.

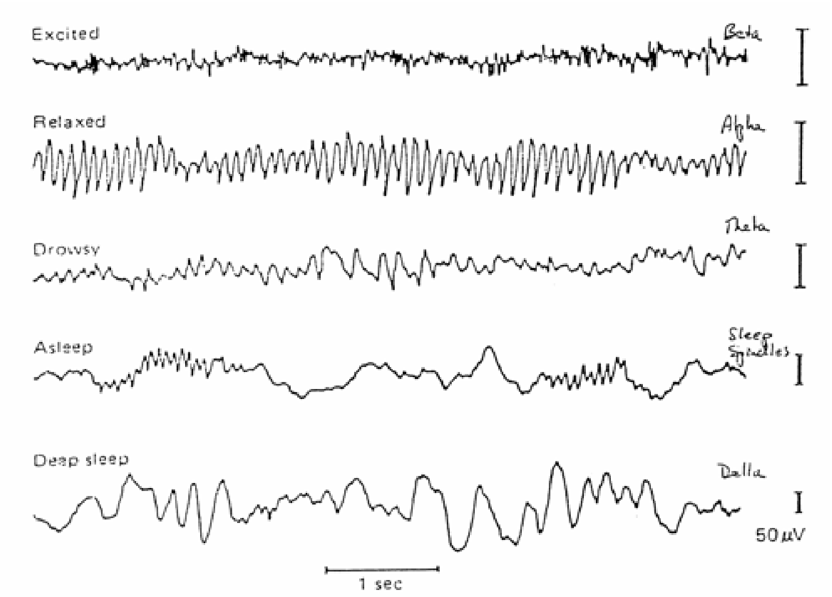


Figure 4.1: Samples of Different Brain Waves [26]

4.2 Typical Circuitry

The circuits that are currently being used consist of some kind of Analog front-end, whether it be a filter or an in-amp, then more filtering is done or there is a gain stage, and finally an ADC or microprocessor to digitize the data to be displayed on a screen. Figure 4.2 shows the circuit from a do-it-yourself EEG site[26]. Figure 4.3 shows the block diagram for the EEG system that would interface with a computer. Most of the systems found tried to make everything portable, so small PCBs were used for the hardware, but these were kept separate from the electrode.

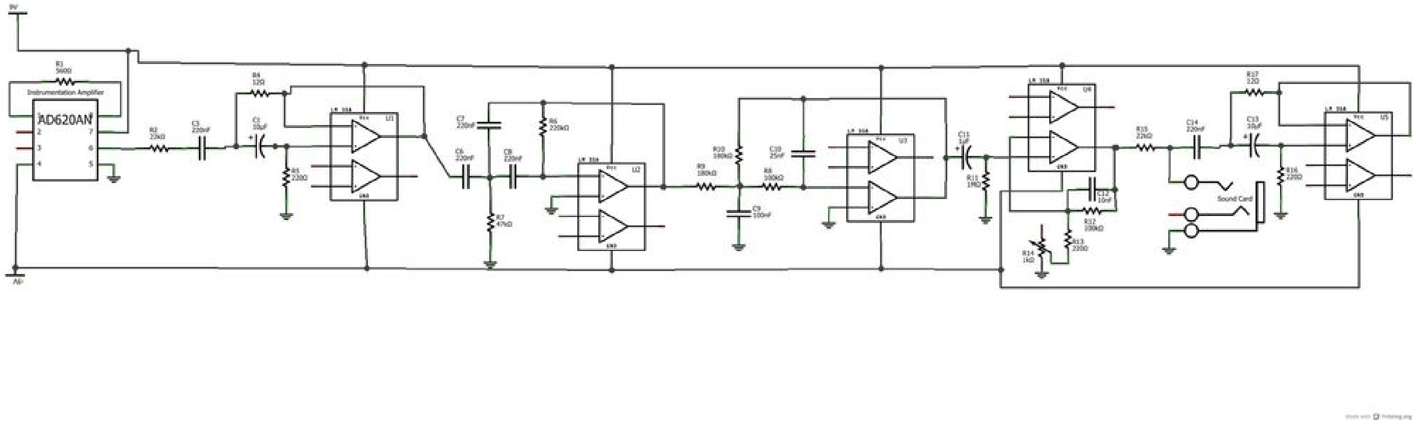


Figure 4.2: DIY EEG Circuit[26]

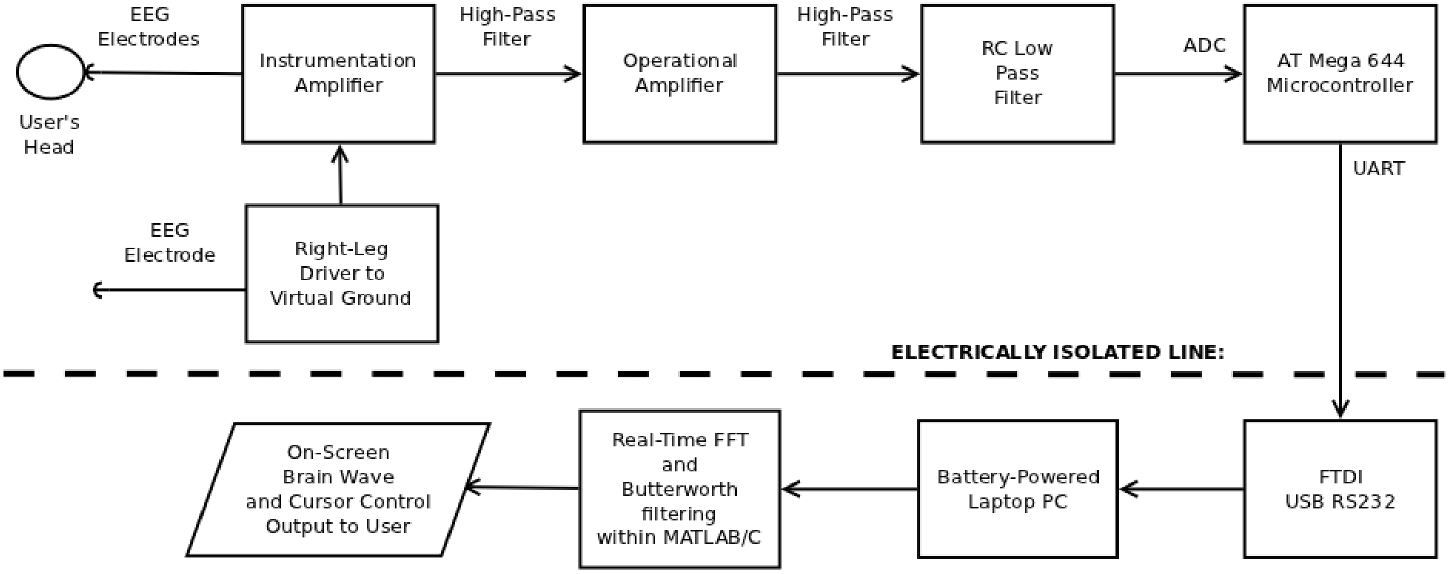


Figure 4.3: EEG Circuit Block Diagram[24]

4.3 Issues and Errors

There are many problems and error sources when it comes to measuring brain waves. Environmental issues like AC power lines, nearby technical devices such as phones or computers, and even lighting will add noise that can be seen at the output of the system. Cardiac signals, muscle contractions and even eyeball movement can also contribute to system noise [28]. It is important to remove as much of this noise from the system as possible, especially due to the very low amplitude of the original signal. Using DC sources in the testing area and instructing the subject to relax and not move their eyes for certain parts of the test can eliminate most of these errors. Figure 4.4 shows what happens to a reading when the subject blinks. Cardiac signals do not contribute very much to errors, as they will be seen at both inputs of the system and will be rejected.

Figure 4.4: Eye Blink Artifacts [24]

One way to remove 60Hz noise from the power supply would be to use a battery to power the system. The circuit from Figure 4.3 shows how a rightleg drive circuit was implemented in order to eliminate the 60Hz frequency by making it common-mode. Shielding the electrode cables may also help, and removing the noise by filtering could also be an option. The analog front-end of the system should also have high input impedance in order to counteract the skin's high impedance. Most of the currently existing EEG circuits use in-amps with about $100M\Omega$ of input impedance, while our system will have a $20T\Omega$ input impedance. This high impedance value will help to maintain as much of our input signal as possible.

DC offset at the output of the sensor may also be an issue if it is too large. This DC offset can be an indicator of how good of a connection the electrode is making with the subject, but only if the reading is below 25mV [27]. Typically, DC offset would be a problem if it caused the AC signal of interest to be close to or cut off by the rails of the amplifier's power supplies. Not only could this affect the quality of some of the data, but it is also not a good setup for an EEG system, as the power supply would be contaminating the reading too. Depending on what voltage supplies are chosen for the system, certain measures will need to be taken to remove this offset as much as possible.

The final issue presented by this project will be working around combining analog and digital parts on the same PCB. One of the issues seen in a previous EEG system was a large amount of noise coming from the ADC's output line in between bit transitions. The clock signal was contaminating the ground, power, and reference lines of the entire system, and in addition there were other parasitics affecting the measurement [27]. Not all noise will be able to be removed by the circuitry itself, so it is possible that some filtering will need to be done in Matlab at the output of the system to allow for an optimal readout.

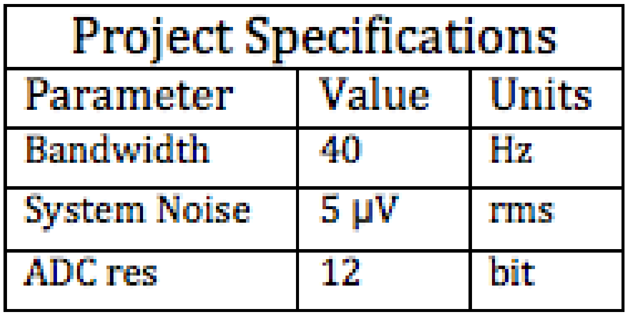
4.4 Requirements

After making the decision to go forward with the brain wave sensor project idea, and through taking the previously found data and research on EEG systems into account, the specific requirements for the system had to be determined in order to proceed with the creation of a list of system specifications. These specifications will lead to the completion of a basic block diagram of our system, as well as helping to decide which testing methods will be used for each component of our system.

Aside from the criteria listed in Section 2.1, there are also a few other requirements that the sensor must meet. The capability of portable power is an important aspect of the system, as well as being as compact of a system as possible for ease of use purposes. The brain wave sensor was chosen partially because it provides an opportunity to include a number of components from ADI. It is important not only that these parts be integrated into the system, but that the system properly highlights the technical capabilities of each of these components. By pushing each of the components to their functional limits within reason, it will be clear what makes each device stand apart from similar existing technologies. Also, taking advantage of the capabilities of each of the ADI components will lead to more of a divide between prior brain sensing systems and the system that will be created. It is important that this brain sensor system stand apart from others as much as possible in terms of design and functional capabilities, such as need for fewer electrodes and a shorter sampling time, in order to demonstrate sufficient innovation within the project.

4.5 Specifications

By taking each of these requirements into consideration, as well as through additional research on existing brain sensors that are in use and the technology and circuitry that drives them, it was possible to obtain a more solid grasp on what the initial specifications of the system should be. It was found that most normal brain waves occur between 0-30 Hz, with some beta waves that read slightly higher than 30 Hz when under a significant amount of anxiety or stress. From these values, it was decided that the bandwidth of our system should be at least 40 Hz, which will be obtained using a mix of several filtering systems. Research on the noise of current brain sensors showed that the noise of these systems is usually pretty high, with a low value of 5μV. Our system aims to have less than $5\mu V_{RMS}$ of noise, and will potentially have around 2μV at most. It was also found that the lower impedance value on the electrode end of the system, the less noise would be in the signal. However, the FET buffer at the front end of the system will have such a high impedance that it should mitigate most noise caused by impedance from the electrode inputs. Even though this high impedance will help reduce noise, it will still be important to make the electrodes as efficient as possible, preferably without the use of any pastes or abrasion methods. For the ADC of the system, at least 12 bits are required for a more accurate readout to the microprocessor. A microprocessor with a built in ADC will be chosen through comparing the capabilities of different microprocessors, comparing the specifications for each on board ADC, and deciding which will be best based off of the desired functionality and performance of our system. It is also important for the system to be easy to use and portable. This list of specifications can be seen in Figure 4.5.



Project Specifications

Parameter	Value	Units
Bandwidth	40	Hz
System Noise	5 µV	rms
ADC res	12	bit

Figure 4.5: Table of System Specifications

4.6 System Block Diagram

Based on continued research of brain sensors as well as the determined requirements and specifications for the system, the block diagram in Figure 4.6 was created.

The output of the electrodes that are connected to the subject is fed into the FET buffers. The high input impedance of these buffers will help maintain the integrity of the signal and a higher SNR ratio, which will be shown in Section 5.1.2. After the buffers, the signal will be filtered to remove the frequencies outside of the 0Hz to 40Hz range, and any offset from the buffers and then electrode-skin connection. The instrumentation amplifier will apply gain to the signal and remove any common mode noise (such as 60Hz). Another op-amp will then level-shift the signal and amplify it further so that it fits within the ADC range. The ADC within the micro controller will convert the signal as it comes in so that it can be sent wirelessly. One Xbee module will receive the data from the micro controller, and the other will receive the data wirelessly. The computer will be running LabView with the serial port connected so the data coming in can be viewed as a waveform.

4.6.1 Test Approach

Basic functionality testing will begin by analyzing each block individually. The input bias current for the buffer and amplifier combination will be

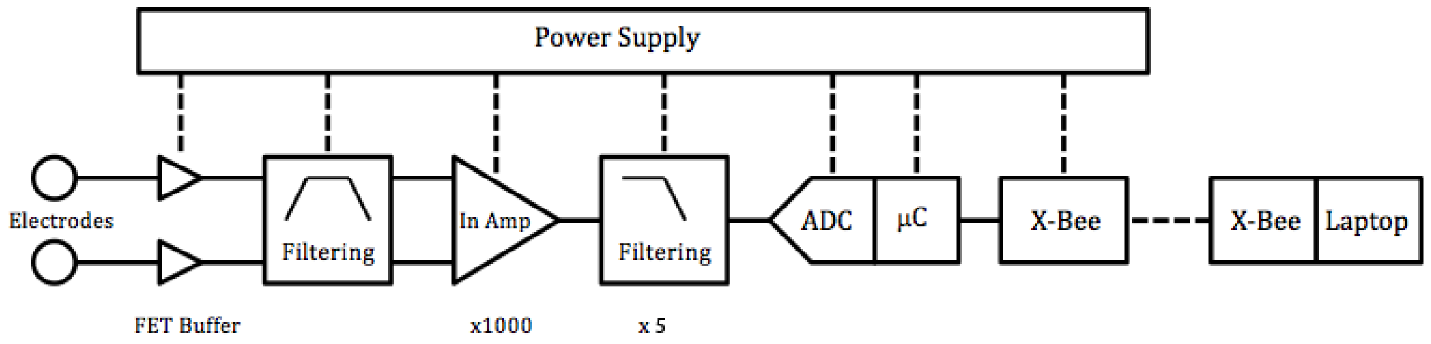


Figure 4.6: Intial System Block Diagram

measured to find out what the impedance of the front end of the system is. The common mode rejection ratio and gain of the instrumentation amplifier will be measured, to determine if the part is working as expected. Initial testing will also consist of testing the voltage inputs from the connected electrodes to the FET Buffer/In Amp board, and measuring the output of the instrumentation amplifier while connected to the electrodes and dual supply 9V battery supply source. This will determine if the in amp is rejecting the common mode signal as expected, as well as if the inputs are giving a typical readout value, and how the voltage levels of the two signals relate to each other. Filter testing will include frequency sweep Bode plots, and possibly a voltage sweep to test the functionality of the ADC. The noise for each block will also be measured. By testing each of the parts individually at first, it will be easier to pinpoint any issues when testing to whole system as a single entity.

Chapter 5

Detailed Design

Before deciding on the final design, several calculations and considerations needed to be made. In Section 5.1.1, an equivalent model of the skin and electrode contact shows that the large input impedance will allow for non-invasive electrodes, which is a huge advantage. Analysis was done on the signal-to-noise ratio from this model, considering the thermal noise of the skin-electrode contact resistance, which is detailed in Section 5.1.2. A signal map was also created as shown in Section 5.1.3 to model what signals were to be expected in each block of the system. Concerns of managing DC offset are discussed in Section 5.1.3, and the factors that were considered for the high-pass/RFI filtering are included in Section 5.1.4. The ADC and wireless choices are detailed in Sections 5.1.5 and 5.1.6, respectively. Finally, the Wireless XBee interface is discussed in Section 5.1.7, and the current consideration of the batteries of this interface are discussed in Section 5.1.8.

5.1 Miscellaneous Design Factors

5.1.1 Brain Wave Sensor Equivalent Circuit Model

In order to get an idea of what kind of noise will be seen at the front end of the system, an equivalent model of the brain wave to FET buffer interaction was created. This circuit, shown in Figure 5.1, models the brain wave as an AC voltage source V_S, the skin-electrode impedance as a resistor R_s and finally the input of the FET buffer as a very large input resistance R_{in}. The resistance of the skin and the electrode were combined in R_s to simplify the circuit.

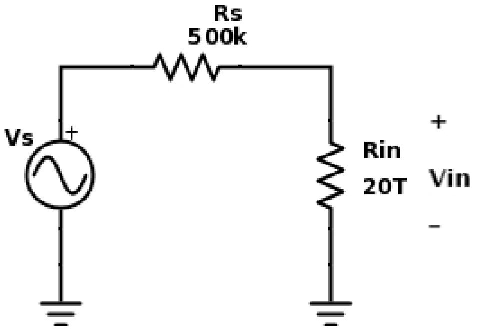


Figure 5.1: Brain to Input FET Buffer Equivalent Circuit

The network creates a simple voltage divider, so the actual signal being seen by the FET buffer would be calculated using Equation 5.1.

$$V_{in} = \left[\frac{R_{in}}{R_{in} + R_s}\right] V_s \tag{5.1}$$

If R_{in} is much larger than R_s, then there will be negligible loss due to the voltage divider network. The 12-bit ADC that we are using has a full scale range of 2V, and an LSB voltage of 488μV. This translates to a full scale voltage of 200μV at the input and 48nV of change. If we say that we had a 200μV input to the system, a drop of 48nV across R_S would be the maximum drop that it could handle before we would see an error at the output of the ADC. The maximum allowable R_S value for less than 1LSB error can then be calculated by seeing that:

$$\frac{R_S}{R_S + R_{in}} \approx \frac{R_S}{R_{in}} < \underbrace{\frac{V_{LSB}}{V_{FSR}}}_{\frac{1}{2^N}} \tag{5.2}$$

This is just the voltage divider relationship of this LSB drop across the resistor R_S. The voltage ratio can be simplified to 2^N where N is the number of ADC bits. The resistor ratio can be simplified, as shown in Equation 5.2, since we are assuming that $R_{in} \gg R_S$. Solving this equation for R_S gives a maximum resistance value of $4.88G\Omega$. This shows that the impedance at the skin needs to be extremely large in order for any loss or error in the output signal to be visible, and that the very large R_{in} contributes to this advantage.

5.1.2 Signal to Noise Ratio (SNR)

Even with the very high input impedance from the FET buffer, the amount of noise due to this configuration needs to be calculated. The R_{in} from the equivalent circuit model in Figure 5.1 does not need to be accounted for in terms of thermal noise. Its effect is captured by the noise sources in the analysis completed in Appendix D. Therefore, thermal noise only needs to be completed on R_s which is the resistance created by the electrode and skin contact. The thermal noise can be calculated using Equation 5.3.

$$V_n = \sqrt{4 \cdot k \cdot t \cdot R_s \cdot BW} \tag{5.3}$$

Determining the signal to noise ratio (SNR) from Equation 5.1 and Equation 5.3 will be an important performance factor for the circuit. The higher the SNR, the better the performance since this means we would be seeing more actual signal than noise. Equation 5.4 shows the SNR calculation, separating the variables of interest.

$$SNR = \underbrace{\frac{1}{\sqrt{4 \cdot k \cdot t \cdot BW \cdot R_s}}}_{Noise} \cdot \underbrace{\frac{R_{in}}{(R_{in} + R_s)} \cdot V_s}_{Signal} \tag{5.4}$$

Equation 5.4 shows that increasing R_{in} will always make the SNR better. When comparing our system to others, the input impedance from the FET buffer ($20T\Omega$) is much better than the input impedance from these other systems (usually around $10M\Omega$ to$100M\Omega$). Using Equation 5.4, Figure 5.2 was created to show the relationship between the SNR and the skin-electrode resistance R_s.

We want the highest SNR possible without the electrodes being invasive.

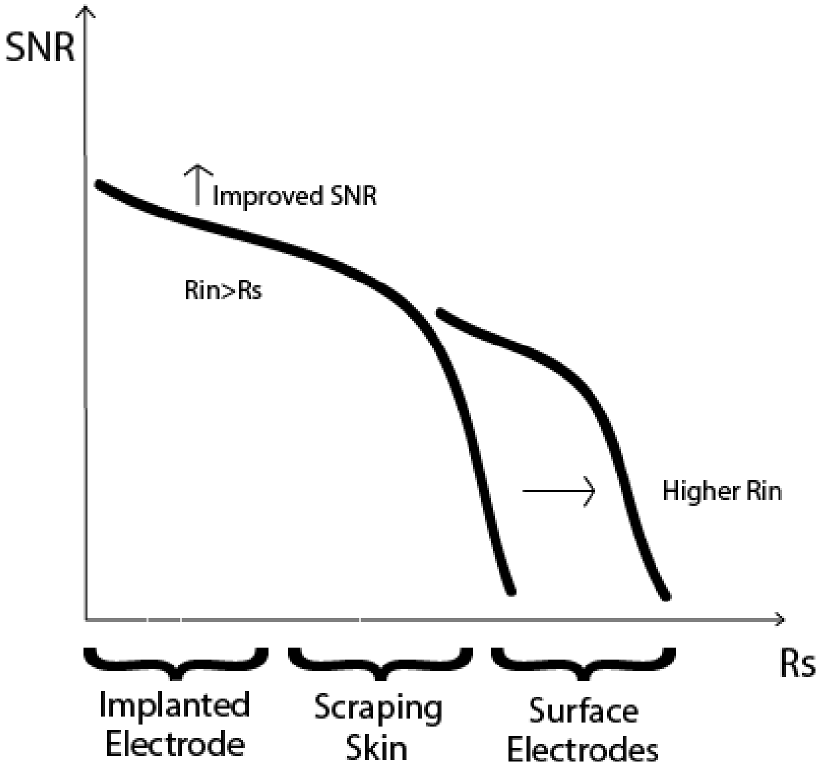


Figure 5.2: SNR Plot Depiction Equation 5.4

At the beginning of the curve when R_{in} is much larger than R_s, the SNR would be degrading by 30dB per decade, typical of an electrode that was implanted into the head to measure waves. Moving down the curve as R_{in} approaches R_s, the SNR degrades by 10dB per decade. Scraping the skin to achieve a better skin-electrode resistance would result in this SNR. Finally, if an electrode will just be mounted to the skin, this would be depicted by the end of the curve when R_{in} is closest to R_s. This is the least invasive method, and if we are able to "push the curve" out enough by using a very large R_{in}, the SNR for our system will be much better than those who are using this minimally invasive method.

5.1.3 Signal Map

When working with any sort of system, it is important that the designer know how the signal will change at each step of the system. Figure 5.3 depicts the predicted signal map through this system. The smallest waves (in amplitude) that we will encounter are Beta waves that can be anywhere from $5\mu V$ to $10\mu V$ peak to peak. These waves will pass through the FET buffer and will remain unchanged except for any thermal noise that is added by the front-end network. After the FET buffer, the signal will be filtered to remove as much noise and DC offset as possible. By removing the DC offset from the front end of the system before the in-amp, it will assure that this unwanted offset is not amplified through the signal chain. The signal will then go into the AD8422 in-amp, where it will be amplified by a gain of around 1000. Another gain stage will be needed at the output of the in-amp, because the Beta waves will only be around 5mV peak-to-peak at this point. A gain of 10 amplifier will be used so that even with the largest signal, alpha and delta waves, the max peak-to-peak output of the system will be approximately 2V. This will be the final signal that is fed into the ADC for quantization. During testing, the Analog Discovery kit will take the output of the ADC and display it on the computer it is attached to. As our prototype progresses, we want to be able to integrate the functionality of a wireless system that will transmit the data from the ADC to a laptop.

DC Offset Management

One of the main concerns regarding our system is how much DC offset would be seen, whether or not it should be removed from the circuit, and if so, at what point within the signal chain it should be removed. It was found from prior studies that the DC offset caused by the front end of the system is usually magnitudes higher than that of the signal itself, usually a measure of mV as opposed to μV. This DC offset value can be affected by factors such as the material of the electrodes in use or the quality of the contact that they make with the skin. In most EEGs, there is a DC offset component in the system. However, this offset from the front end will be amplified when it goes through the instrumentation amplifier, which could cause issues with headroom within the circuit, clipping the signal of interest, especially if the voltage supplies that are used by the system are relatively small. It will be possible to almost completely remove this offset using the AC coupling of a high pass filter to allow for higher readability of the output signal. Most EEG systems have anywhere from 25mV to 300mV of DC offset from various sources, while this system aims to have a lesser value than these current devices.

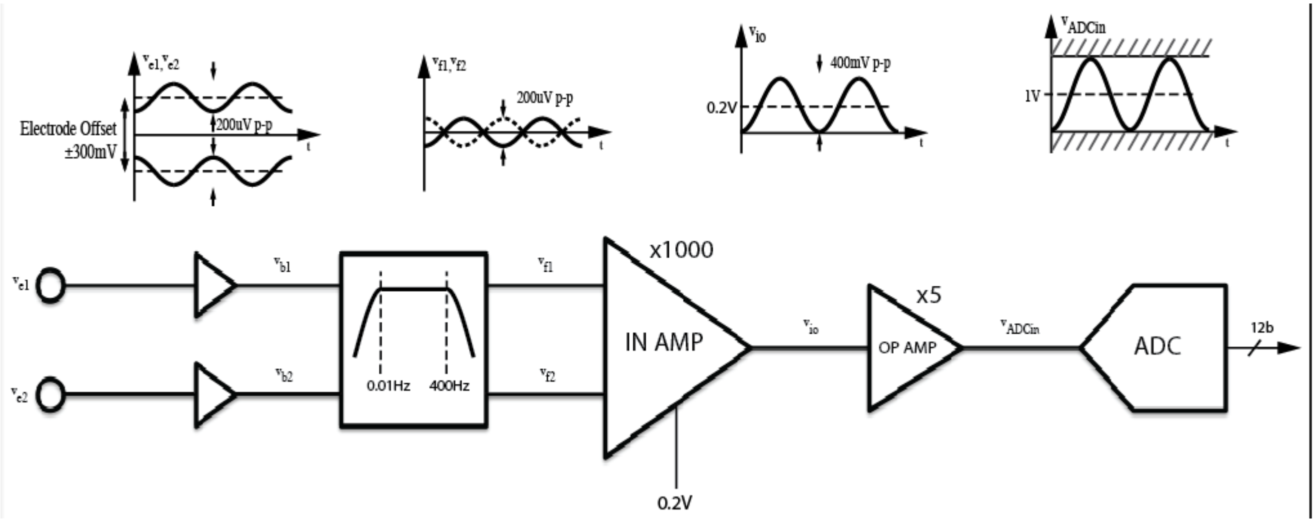


Figure 5.3: Signal Map of Brain Wave System

During early stages of testing, 9V batteries were used as the positive and negative voltage rails of the system. Because these values are relatively high in comparison to the amplitude of the entire system, approximately 2V peak to peak, the DC offset did not distort the output signal. However, this setup did give a good sense as to what the DC offset of the system would be before filtering, and showed that there was indeed a large offset (23μV) from the front end of the system that would need to be filtered out, even before the electrodes were added. The measured offset with the inputs to the system grounded ranged in the hundreds of microvolts to about 1mV. Even more offset (up to 300mV) was seen when the electrodes were attached and this initial offset was amplified through the high gain of the instrumentation amplifier, which lead to the decision that DC offset filtering would need to be done before the in-amp in order to achieve the desired results. Because the power rails for the system are planned to be between positive and negative 3V, the DC offset will need to be removed as much as possible, because even small values might cause headroom issues. Assuming a worst case of only 2.8V rails and expecting a signal of 2V pk-pk, this leaves only 800mV of head room at the output, which translates to only 80μV at the input that can cause issues. If some discrepancy within the system causes the amplitude of the signal to be a bit larger than the expected 2V pk-pk voltage range, due to differences in signal input or gain, then even small DC offset values could cause issues at the output, and will need to be removed.

When removing DC offset, it is important to put the circuitry that causes it directly after the source of the offset, which would be after the FET buffer in the system, as it is the first place within the circuit that there will be freedom to add in additional circuitry. One method of DC offset that was explored was AC coupling. If AC coupling is to be used, then the high pass filter created from the circuitry will need to have an extremely low corner frequency, so as to not distort the data that is needed from the electrodes. This filter will effectively remove DC components, as DC is considered "infinitely far away" from the AC frequency response. Another method of removing DC offset is through an active nulling loop, which would be designed based off of specific aspects of the existing circuitry. This loop can be done digitally or with analog methods. Because a microcontroller with a built in DAC will be present in our system, the digital method of filtering out this DC bias might be more accurate and efficient, but would require extensive coding. An analog feedback circuit could also be created based off of the architecture of the rest of the circuit and the voltage levels that are present.

It was decided that the most efficient DC offset management method in terms of time constraints for our project would be the AC coupling method. These simple, first order filters will be implemented directly after the FET buffer, and will also be equivalent on both inputs to the instrumentation amplifier in order to avoid any matching issues. The two high pass filters will have to have a very low cutoff frequency, of around 0.05Hz, so as to not ruin any important data from the system. These filters will then integrate into the RFI reduction filter before the signals enter the instrumentation amplifier. Because there is no buffer between each of the filtering methods, the circuitry of one filter will slightly affect the performance of the other. The transfer functions for the summation of these two filters can be seen in Appendix C. Through these equations, values of $23k\Omega$ and $100k\Omega$ were found for the resistors and $384\mu F$ and 3.98nF for capacitor values.

5.1.4 Signal Filtering

After finding that the waveforms that are of interest for this system lie between 0 to approximately 35 Hz, it was decided that the general range for the bandwidth of the system would be 40Hz. This allows the system to acquire the necessary data without compromising the signal to noise ratio of the system. In order to achieve this bandwidth, it is important that we choose a cutoff frequency that is not too close to the band of interest so that the signal does not get attenuated. A cutoff frequency of 400Hz was chosen to avoid this signal attenuation. To remove the high frequency noise of this system, a low-pass filter will be used.

RFI Filter One of the considerations that was made in regards to the filtering process was where in the signal chain it should be located. The two main locations that were considered for this filter were before or after the instrumentation amplifier in the circuit. It was found that when high frequency signals go into the in-amp, they are going to become distorted, and radio frequency interference (RFI) rectification can possibly create lower frequency products within the band of interest, which will negatively effect the SNR at the system's output. To mitigate this effect, a simple RFI filter will be placed before the instrumentation amplifier. The design of this filter is shown in Figure 5.4. The values for R1 and R2 should be as close to equal in value as possible in order to keep the inputs of the instrumentation amplifier matched. To accomplish this, resistors with a 1% tolerance will be used. Any significant mismatching of these values will cause an increase in the CMRR, which is detrimental for the signal to noise ratio.

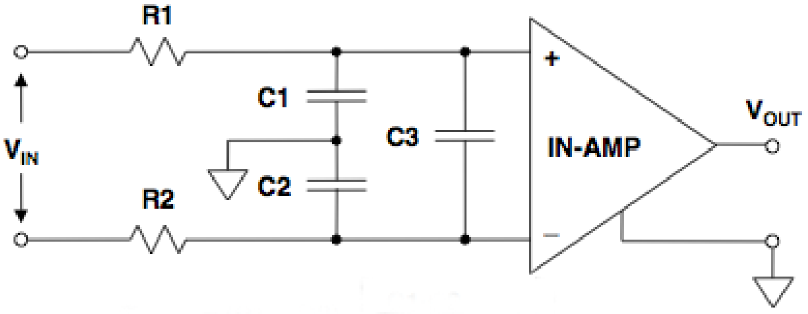


Figure 5.4: RFI filter circuit

The design for this filter originally included two capacitors to ground (C1 and C2 in Figure 5.4) to allow for a more exact calculation of the cutoff frequency of the filter. These capacitors also filter the RFI of the common mode voltage. However, because capacitors usually have a tolerance value that is higher than that of resistors, typically 20%, potential mismatching of the values is highly likely. Because capacitors with very low tolerances are considerably more expensive, these capacitors will be eliminated from the design to protect from any negative effects to the CMRR. This means that there won't be RFI filtering of the common mode voltage, but because the CMRR of the in-amp at the frequencies within our bandpass filter is very high, and because the bandpass filter eliminates any frequencies with a lower CMRR level, this common mode voltage will still be successfully removed during the In-Amp stage. This omission of the capacitors to ground is acceptable for our system because the exact cutoff frequency of this filter is not important to the performance of the system, it is simply to remove any RFI artifacts from the system. The desired bandwidth of the RFI filter should be around 100x that of the desired final bandwidth of the system, or 4kHz. However, this ratio comes from the assumption that other high frequencies are important at the output signal, while our system is entirely dependent on very low frequencies. Because of this, the cutoff frequency for the RFI filter within our system can be anywhere in the low kilohertz range. The process to find the cutoff frequency for this filter can be found in Appendix B.

Common Mode to Differential Mode Conversion While designing for the filter components, we realized that there could be problems if there were mismatched components in the high-pass filter. The RFI filter components would have no effect on the output since the capacitor is across both of the outputs and is modeled as an open circuit for common mode and DC signals. The resistors are in series with the outputs, so they will not affect the analysis either. To figure out how this mismatch would affect the output of the filter, it is only necessary to find the transfer function of each of the high-pass filters with a common mode input. The circuit used for the analysis is shown in Figure 5.6.

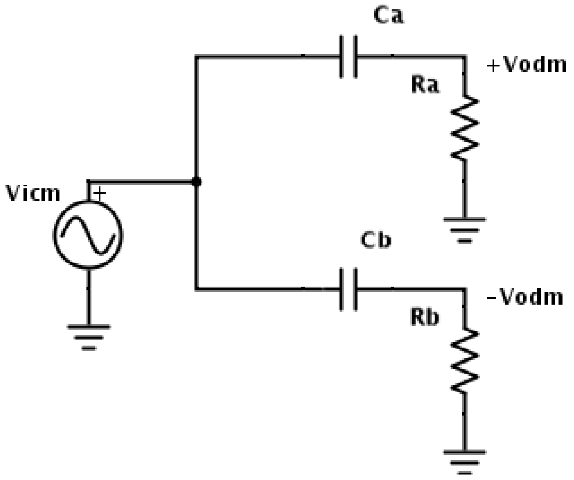


Figure 5.5: Common Mode to Differential Mode Conversion Analysis Circuit

The transfer function of the high pass filter is:

$$\frac{V_o}{V_{icm}} = \frac{s\tau}{1 + s\tau} \tag{5.5}$$

where τ_x is $R_x \times C_x$. Assuming that the values of the high-pass filters are slightly different from each other, we can use τ_a and τ_b to distinguish them. To find the difference between the two outputs, it's simply the difference between the two transfer functions:

$$\frac{V_{oDM}}{V_{iCM}} = \frac{s\tau_1}{1+s\tau_1} - \frac{s\tau_2}{1+s\tau_2} \tag{5.6}$$

To get a common denominator, we multiply the numerator and the denominator of the first fraction by $1 + s\tau_2$, and the second by $1 + s\tau_1$. After simplifying this expression, we get:

$$\frac{V_{oDM}}{V_{iCM}} = \frac{s(\tau_1 - \tau_2)}{(1+s\tau_1)(1+s\tau_2)} \tag{5.7}$$

We can see from the transfer function that we will have a zero at $\frac{1}{\tau_1 - \tau_2}$ and poles at $\frac{1}{\tau_1}$ and $\frac{1}{\tau_2}$. To get a better understanding of what the percentage difference is between the components, τ_1 can be replaced with $\tau(1 + \frac{\epsilon}{2})$ and τ_2 can be replaced with $\tau(1 - \frac{\epsilon}{2})$. Substituting these in and assuming that the error is very small, we find:

$$\frac{V_{oDM}}{V_{iCM}} = \frac{s\tau\epsilon}{1+2s\tau+s^2\tau^2} \tag{5.8}$$

The circuit was simulated in PSpice (results shown in Figure 5.6), and the magnitude of Equation 5.8 was plotted using Matlab with a 1% error, shown in Figure 5.7. The design for the error is determined by how big the gain will be at 60Hz. The larger the gain, the worse the SNR. We are expecting a maximum 150mVpk-pk to 200mVpk-pk of 60Hz common mode noise at the input. Using Matlab, we found that we need a .06% error for a 200mVpk-pk signal in order to see 1mV of noise at the output. If we were to assume 150mVpk-pk, we would need a .08% error to achieve the same result. This is very difficult due to the tolerance of the capacitors, so they were measured to find the closest possible pairs.

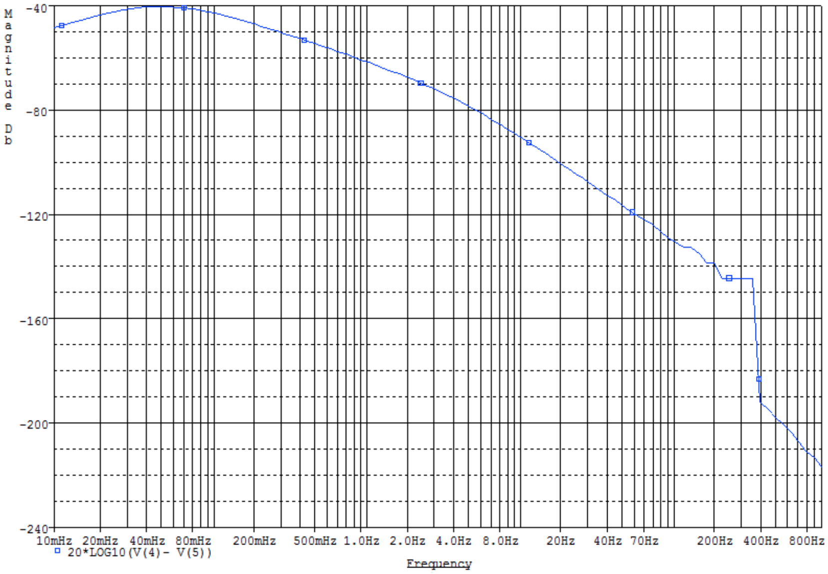


Figure 5.6: PSpice Common Mode to Differential Mode Conversion Analysis

Op-Amp Low-pass Filter The capabilities for a second filter will be added after the instrumentation amplifier, to filter out any remaining frequencies below the cutoff frequency of the RFI filter that are not in our band of interest. Because the filters within our system are sufficient enough and do not need any additional lowpass filtering, the operational amplifier will simply be configured to have a gain of 5. This op amp will have to add minimal noise and voltage offset into the circuitry, while

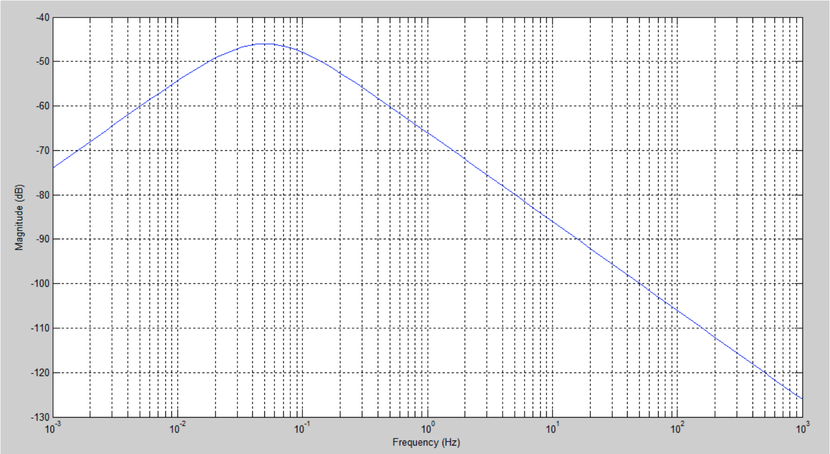


Figure 5.7: Matlab Common Mode to Differential Mode Conversion Analysis

utilizing the voltage rails of the system. These necessary requirements led to choosing the OP97 op amp, which has a maximum voltage offset of 20μV, an incredibly low value that will guarantee for minimal effects to the system signal (up to 500mV of offset is allowed before going outside the voltage range of the system). The op amp also has a lower noise level compared to other amplifiers, with an input voltage noise of .5μV/$\sqrt{Hz}$, and an operating voltage range of +/-2V to +/-20V, which allows for the same voltage supply as the rest of the system of +/-3V.

5.1.5 Analog/Digital Converter Decision

For this system, it's important to have a high enough sampling rate so a clean signal can be seen once it reaches the computer, but also to try and limit the quantization noise in our signal from the ADC. It has been found that EEGs typically have sampling rates of 250-2000Hz**??**. It is necessary to have a quick data transfer while maintaining signal integrity. The cost to performance tradeoff is also important. The minimum sampling rate of an EEG signal was found to be between 2.5 to 3 times the size of the highest frequency of interest, which gives this system a minimum sampling rate of 120Hz. Because it is not difficult to obtain a sampling rate above this value without compromising system speed, the sampling rate should be at least 250Hz, as is with typical EEG systems. It is also important to find an ADC that has a similar bandwidth to the system and is relatively close to the voltage levels that we will be using, as this would mean each bit would be representing a larger voltage than it would have if the full scale range was smaller, making the output less precise. Another important factor in choosing which analog to digital interface would be used in this sensor is whether or not there is a benefit from other capabilities of the device. A microprocessor would not only allow for data transfer from the analog to digital world, but would also more easily integrate into a data transfer device such as a USB. A microprocessor that also uses a DAC would be useful in sending the DC voltage levels of the signal back to the reference pin of the in amp, allowing for an effective digital nulling loop that would be one method of managing the DC offset levels.

5.1.6 Data Communication Options

There were a few different options for finding a way for the sensor system to send data to the computer. An FTDI cable would take the data from an ADC and send it to a computer through a USB port [29]. Another option was a Bluetooth module that had a processor onboard and a UART [30]. Finally, two XBees could be configured, one with the sensor system and one connected to a computer to receive the data coming from the other XBee [31].

The FTDI cable implementation was eliminated first because it was more desirable to have a wireless interaction between the sensor and the computer. While this might be the easiest data communication method, we would not be able to highlight any more ADI components.

The Bluetooth module seemed to be very complicated and we anticipated that a lot of time would have to be put into figuring out how the configuration software would work. If we were to interface it with Apple products, we would need to register as a developer for a fee in order for the bluetooth to work. While the module already had a microcontroller onboard, it wasn't clear whether it had an ADC, which would be a hassle to also configure if it didn't. For these reasons, the Bluetooth module idea was dismissed.

Finally, the XBee module interaction seemed to be the best option. While the module has an ADC onboard, a technique called "line passing" [31] would need to be used so that it would not have to interact with a microcontroller.

This technique involves programming the XBee so that any inputs to one module appear on the same pin of the other module. However, we want the data being given to one to go to the computer attached to the other, and a better ADC than the 10-bit one on the XBee. There are several different breakout boards available that would allow the XBee to easily interface with a microprocessor so that we can use the ADC within the microprocessor and not have to worry about programming the XBee. It seemed like there was enough time to figure out how to configure these modules correctly and be able to use even more ADI parts, so the XBee was chosen for the data communication system.

5.1.7 XBee Interface

The XBee modules will talk to each other immediately when they are powered up, no additional programming or configuration is necessary. There is a dongle available on SparkFun that will allow us to plug in the receiving XBee right into the USB port of a computer[32]. The sending XBee that will be attached to the sensor circuit, sending the digitized brain waves over to the computer will have to be mounted on a breakout board so that it can interact with the microcontroller. The XBee is defaulted to communicate to another XBee through the UART, meaning that when powered, the data coming into the standard UART pins will be sent wirelessly to the other XBee. The microcontroller chosen has an ADC onboard to take care of the digitization of the brainwave signal, and has a UART that will be able to talk to the XBee.

5.1.8 Battery Current Constraints- Capacitor Selection for XBee Interface

A concern arose when the battery chosen for the system had a 10mA max current specification when it needed to supply the 45mA transmit current demanded by the XBee. We determined that a capacitor would need to be placed in parallel between the battery and the XBee to prevent the battery from supplying its max current and to protect it from the current spike when the XBee transmits. Figure 5.8 shows the equivalent circuit model for the battery, Xbee, and capacitor configuration.

Most of the power consumption in this device comes from the transmitting XBee. We attempted to reduce the transmit power with one of the programmable settings, but this did not change the transmit current. An alternative method to sending the data that uses less power or has a lower programmable output power could be implemented in a future prototype. Figure 5.9 outlines the current and voltage needed for each component. The XBee runs at a higher current when receiving. From the table, it is clear that the Xbee uses the most amount of current and that even a 5 percent decrease in current used by the XBee would be significant.

Because the device runs on batteries and such a significant amount of

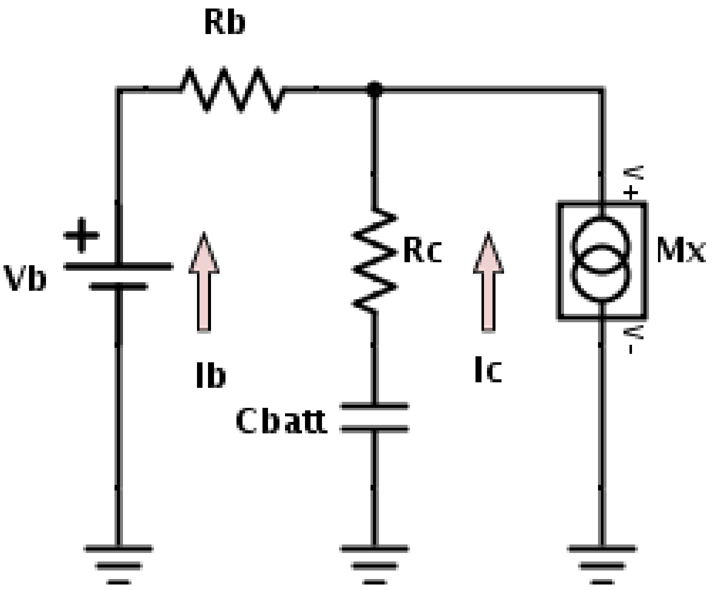
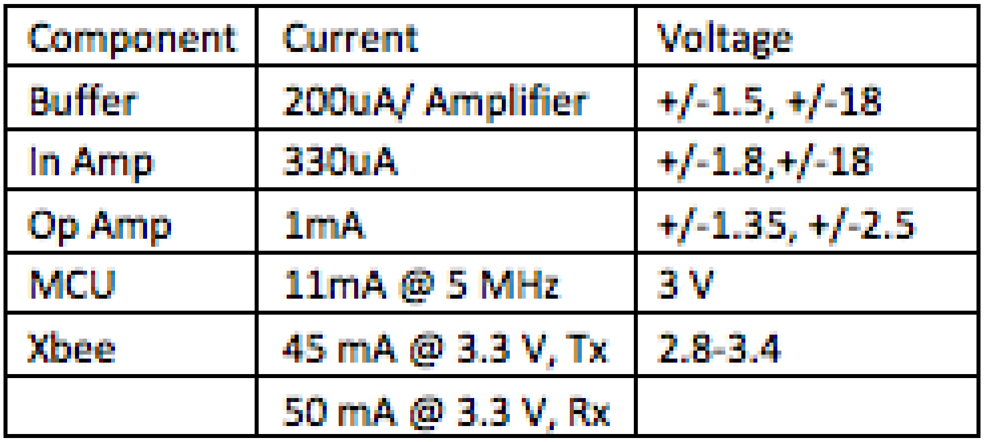


Figure 5.8: Battery to XBee Interface with Capacitor Equivalent Circuit



Component	Current	Voltage
Buffer	200uA/ Amplifier	+/-1.5, +/-18
In Amp	330uA	+/-1.8,+/-18
Op Amp	1mA	+/-1.35, +/-2.5
MCU	11mA @ 5 MHz	3 V
Xbee	45 mA @ 3.3 V, Tx	2.8-3.4
	50 mA @ 3.3 V, Rx	

Figure 5.9: Currents and Voltages Required by System Components

current is needed in order to successfully transmit the data, it is important to use capacitors in parallel with the batteries and the voltage regulators to reduce the spikes in current being pulled from the battery. For the positive 3V rail, a capacitance of 22 uF is needed for that purpose. The negative 3V rail did not require such a large capacitance, because it supplied current to only the buffer, the in-amp and then the op-amp. To determine the capacitance needed in order to protect the batteries from current spikes, it was necessary to first calculate how often the XBee needed to transmit, then calculate the capacitance needed to reduce the spike to 10 mA or less, the rated current value for short bursts.

To determine the size of the capacitor needed, we first needed to measure the current being drawn by the XBee. The output power was tuned down to .316mW, but the transmit current still measured 44.58mA. Next, we needed to determine the amount of time the XBee was actually transmitting to the other XBee, and find the internal resistance of the battery. The microcontroller is programmed to send the data coming into the XBee every 10ms, and each time it transmits it sends 5 bytes. Each of these bytes includes 8 data bits, a start bit and a stop bit. The RF data rate is 250,000bps, so with 50 total bits, this translates to a transmit time of $200\mu s$. The waveform timing diagram in shown in Figure 5.10. From the battery data sheet[33], we extracted two data points from the V-I characteristic to find an internal resistor value of 175Ω.

Knowing the current demanded by the XBee, the value of internal resistor, the maximum current that can be provided by the battery, and the amount of time the XBee transmits, we can design for C_1 and R_C. At any time,

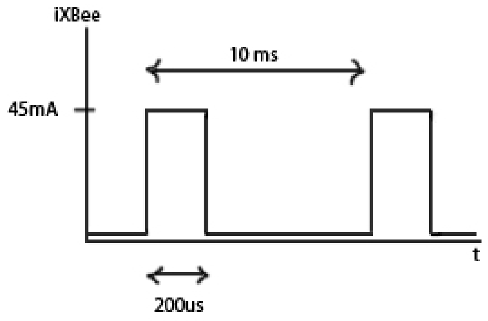


Figure 5.10: Xbee Current Waveform Timing Diagram

the current from the battery combined with the capacitor current must sum to the current demanded by XBee. The capacitor voltage and current will follow an exponential path, defined by Equation 5.9.

$$V_t = V_F - (V_F - V_I)e^{-t/\tau} \tag{5.9}$$

In which V_t is the voltage at time t, V_F is the final voltage the capacitor will charge to, V_I is the starting voltage, t is the time at which we are evaluating the voltage of the capacitor, and τ is the rise time constant for the circuit. At time 0, the XBee will see a voltage that is equal to $V_B - I_{PK}(R_B||R_C)$, where I_{PK} is the maximum current being drawn by the XBee, 44.58mA. This can easily be seen by redrawing the circuit from Figure 5.8 to show its Thevenin Equivalent, shown in Figure 5.11.

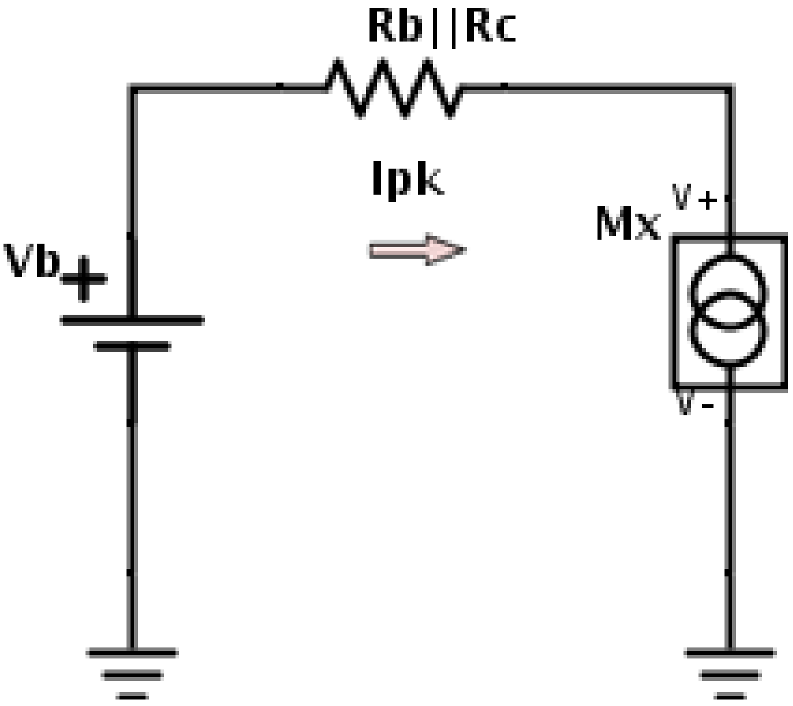


Figure 5.11: Battery Interface Thevenin Equivalent

Since there was no current flowing, the battery voltage will appear across the capacitor, making the XBee see the two resistor in parallel. Eventually, the current through the capacitor will reach a steady state (no current will flow) and the voltage across the XBee will just be $V_B - (I_{PK} * R_B)$. More importantly, the current characteristic is similar to the voltage characteristic. Figure 5.12 shows the currents throughout the circuit in Figure 5.8.

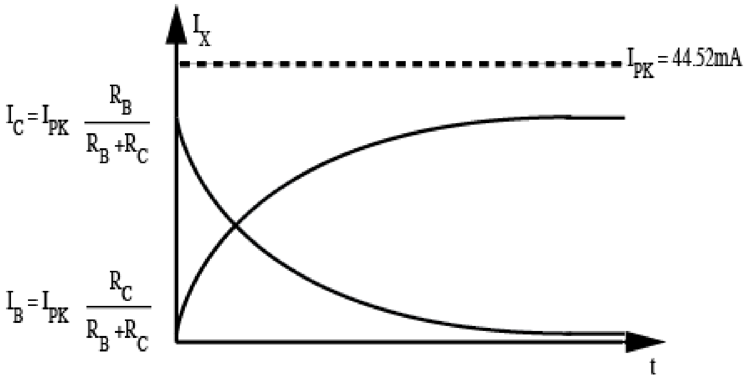


Figure 5.12: Currents in the Circuit of Figure 5.8

The same reasoning used to find the starting voltage for the XBee is used to find the starting currents throughout the circuit. The battery and capacitor currents both start as the peak Xbee current multiplied by a resistor ratio, then follow the characteristic described by Equation 5.9. We need to make sure that the battery current does not exceed 10mA before the XBee finishes transmitting, so we know that at a time of $200\mu s$ we want the battery current to be smaller than 10mA. We have a $\tau = (R_B + R_C)C_{Batt}$, which can be seen from the filter created by the capacitor and resistors in the circuit. We first chose the current that we want the battery to be supplying at the $200\mu s$ to be 5mA, which is reasonably under 10mA. Then, we chose an τ of 4ms for the rise time of the circuit, just to make sure it was well below the time of the full pulse period. Equation 5.9 was changed to model the current, which can be seen in Equation 5.10.

$$I_t = I_{PK} - (I_{PK} - I_I)e^{\frac{-t}{\tau}} \tag{5.10}$$

I_I is simply the starting battery current, which we found to be $I_{PK}\left(\frac{R_C}{R_B+R_C}\right)$. Since we are looking at a very small portion of the waveform between 0s and $200\mu s$, to make the analysis easier we assume a linear approximation for the curve up until the $200\mu s$. This is done by approximating $e^{-x} = 1 - x$. Substituting this back into Equation 5.10, and assuming that the ESR is much smaller than the series resistance of the battery we see that:

$$I_t = I_{PK}\frac{t}{R_B C} \rightarrow C > \frac{I_{PK}}{I_{BMAX}}\frac{t}{R_B} \tag{5.11}$$

where I_{BMAX} is the maximum output current of the battery, 10mA. Substituting in all of the values we have found for the peak XBee current, the time, and the battery series resistance, we find that we need:

$$C > \frac{44.58m}{10m}\frac{200\mu}{175.4} \rightarrow C > 5.08\mu F \tag{5.12}$$

This means that any capacitance value above $5.08\mu F$ could be used, as long as the ESR resistor is much less than the series resistance of the battery. Most data sheets do not have the ESR in the specs, but they do provide a testing frequency and a dissipation factor (DF) that can be used to calculate the ESR, as shown in Equation 5.13.

$$ESR = \frac{DF}{2\pi fc} \tag{5.13}$$

We chose a $22\mu F$ capacitor that had a DF of 0.03 and was tested at 120Hz. This makes the ESR value 1.8Ω, which is much smaller than the battery series resistance of 175.4Ω, so it was safe to eliminate this value from the τ equation.

Chapter 6

Design Testing and Simulation

While finalizing the design, testing and simulations were done to see what kind of noise to expect from the system and to see the frequency response for the filters we need to use. This chapter will review in detail the results of the simulations, and will explain the procedures and results of the bench tests.

6.1 Initial Testing

Before the final system was assembled, each of the blocks were tested separately for functionality. The FET buffer and in-amp configuration, the High-pass and RFI filter combination and the microcontroller/wireless block were tested.

6.1.1 FET Buffer and In-Amp testing

Analog Devices provided two PCBs for testing the FET buffer and the in-amp together. To see what kind of noise is expected due to the FET buffer and the in-amp, the inputs to both were grounded, and the output of the in-amp was observed on an oscilloscope. The maximum noise seen was 20mV peak-to-peak and the offset ranged between microvolts to at most 1mV. Then, the inputs of the FET buffer were hooked up the the electrodes that were configured to measure ECG, since this signal is expected to be in the mV range. The second op-amp gain stage could not be put on the PCB, so there isn't enough gain or filtering to resolve the microvolts that we are expecting for the size of the brainwave signal. The ECG signal was recorded using both PCB boards, and we saw immediately that there was a lot of 60Hz noise. To try and remove some of it, the electrode cables were replaced with a shielded cable. Once this was done, the screen shot and data from the scope was saved and imported into Matlab where an FIR bandstop filter around 60Hz was applied. The original signal is shown in Figure 6.1. The glitch in the signal is due to the oscilloscope not being able to save all of the data. The frequency response is shown in Figure 6.2 and filtered signal results are shown in Figure 6.3.

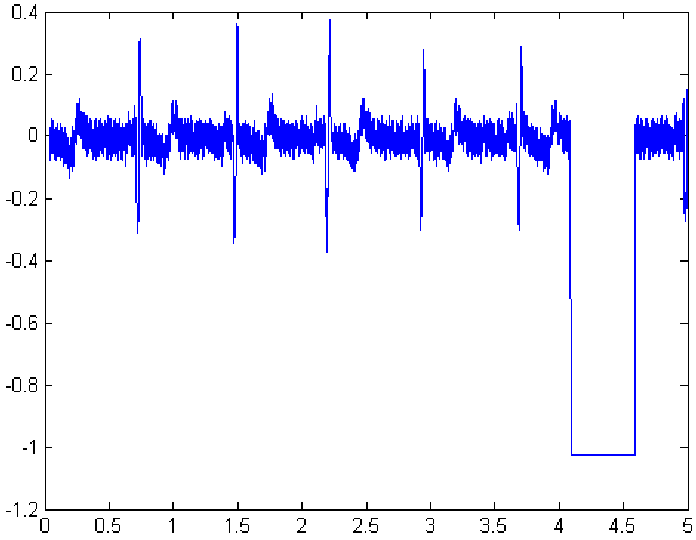


Figure 6.1: ECG Original Signal

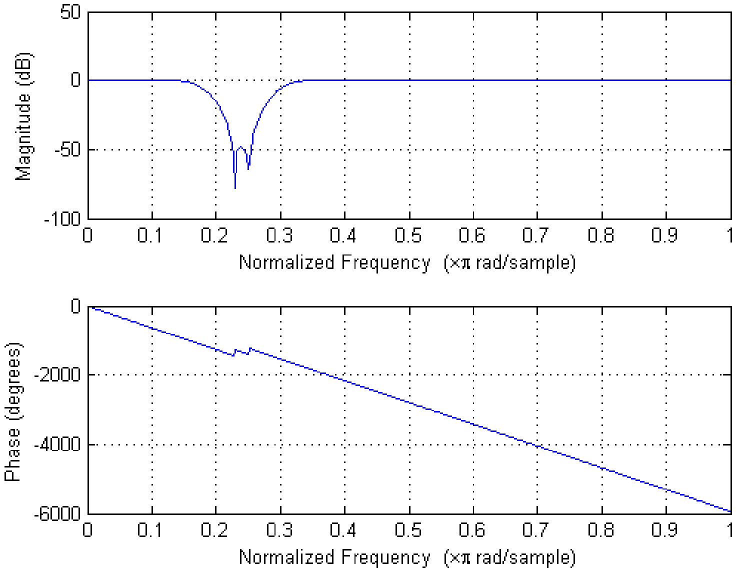


Figure 6.2: FIR Filter Frequency Response

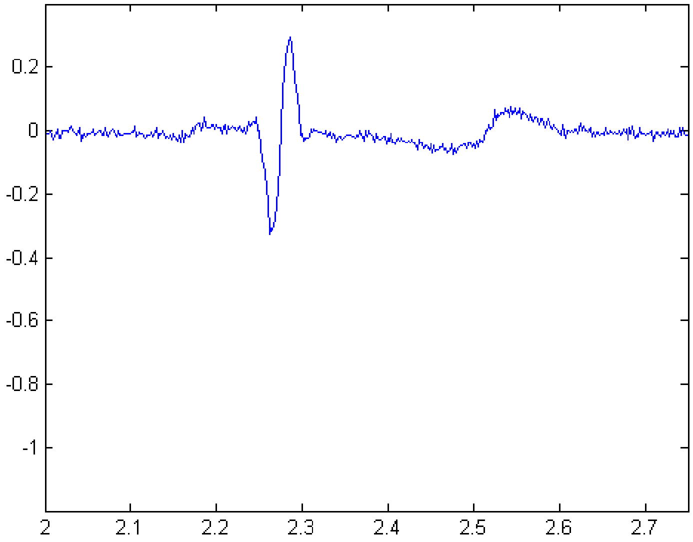


Figure 6.3: ECG Filtered Signal

6.1.2 Microcontroller and Wireless Block

The wireless communication part of the system was also tested. One of the XBees was connected to the computer through the USB dongle[32], and the other was connected to an Arduino Uno[34]. Two of the Arduino pins were configured as the send/receive pins to talk to the XBee. Another was configured as an analog input, where it took the output voltage of a voltage divider and converted it to a digital signal using its 10-bit ADC. The voltage divider was created using a potentiometer so that the output voltage could be changed easily. After the XBees were configured to talk to each other and have a specific baud rate, the code for the Arduino was modified so that the samples were taken every 10ms. Once the data was being transmitted from the voltage divider to the computer, LabVIEW was used to pull in the data from the serial port and plot the voltage vs. time as the data was being received. The Arduino code can be found in Appendix E.1. Once the ADμC7021 arrived, the code was rewritten to be compatible with this microprocessor. The final results of this test with the correct micro controller can be found in 7.2.2.

6.1.3 High-Pass/RFI filter testing

The cascaded high-pass and RFI filters were built on a breadboard to be tested. Since both inputs of the system need to be filtered, both halves of the circuit were built so that they could easily be tested as half circuits and then joined together for testing of the differential output of the signal. This filter is shown in Figure 6.4, where the 390pF capacitor is two 780pF capacitors in series.

Using a 100mV sine wave input, the bode plot of the half circuit filter was created by sweeping the frequency of the input sine wave from .01 Hz to 5kHz and measuring the change in amplitude and phase at the output. The resulting magnitude and phase bode plots of the half circuit can be seen in Figures 6.5 and 6.6 respectively.

After testing the functionalities of each of the half circuit filters, the capacitive bridge was connected between the two to create a single filter system designed for a differential input and output. In order to test the filter, a 100

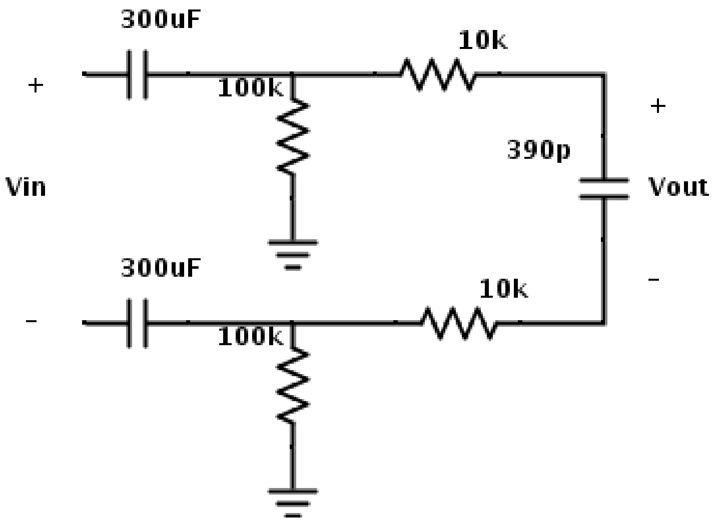


Figure 6.4: Filter Test Circuit

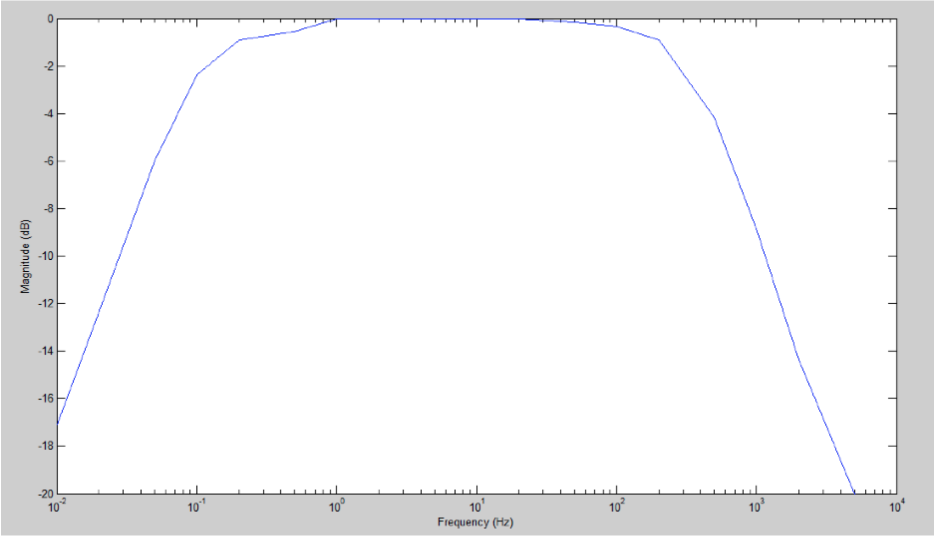


Figure 6.5: Half Circuit Bode Plot

mV sine wave was sent to one input, while an inverted 100mV sine wave that was created with an inverting amplifier configuration was sent to the other input. The differential signal was read at the outputs by subtracting the inverted signal form the non inverted signal, and measuring the amplitude and phase of the resulting waveform. The differential magnitude and phase bode plots are shown in Figures 6.7 and 6.8.

A simple test was done to check the common mode to differential mode conversion. This involved changing one of the $100k\Omega$ resistors to include a potentiometer in order to vary the error. A large common mode signal was injected into the filter, and the output was observed on the oscilloscope. We verified that when the potentiometer was turned so that the resistance values were very close, almost nothing appeared at the output. However, noise and unwanted signal was observed when the error between the two resistors was much larger.

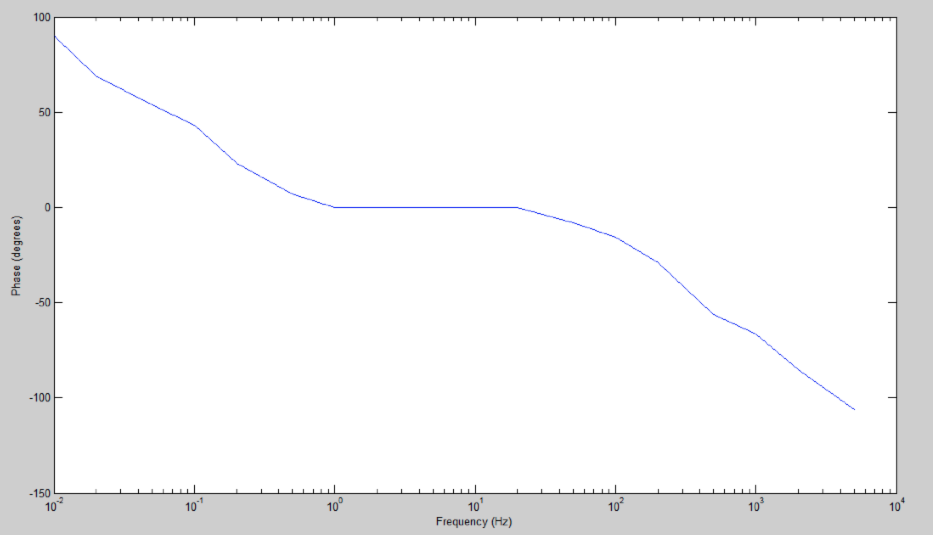


Figure 6.6: Half Circuit Phase Bode Plot

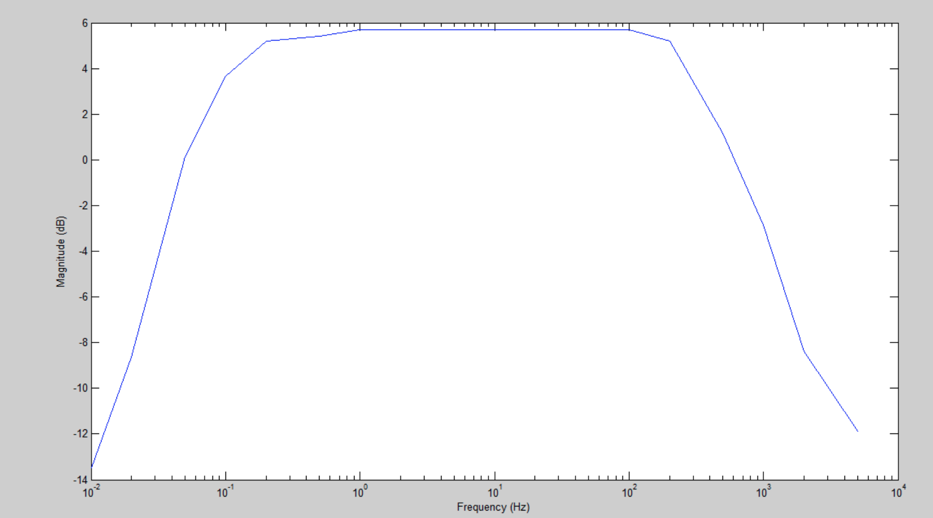


Figure 6.7: Whole Filter Circuit Differential Bode Plot

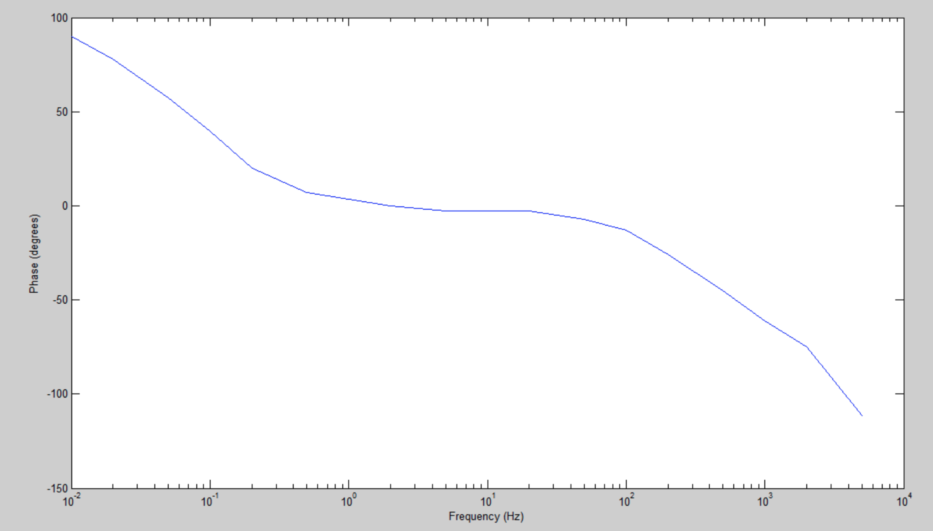


Figure 6.8: Differential Phase Bode Plot

Chapter 7

Results

After designing and constructing each of the different components within our system, along with testing and simulating each of these system segments to check for expected functionalities, the system was assembled into a final prototype. This chapter will detail the results from testing the system as a whole, and explain any expected or unexpected results.

7.1 Printed Circuit Board

The PCB had a few constraints. In order to increase the portability of the board, the board needed to be designed to be as small as possible. The minimum size ended up being dependent on the size of the largest components, the batteries. The number of layers used had to be kept to a minimum as well since the cost of the board would increase as the number of layers increased. Finally, the use of differential signals meant that the board had to be symmetrical in some aspects.

The battery size restricted the ability to minimize the size of the board. The batteries were .945” in diameter, so the board ended up being 2” x 1.35” in size. Because the batteries were so large, they took up the bottom of the board. No other components could be placed on the bottom of the board, but signal traces could still be placed on the bottom. In order to save more space, the XBee was attached to the board using header pins to raise the module off of the board. This saved space and allowed access to the microcontroller by allowing the Xbee to be easily removed.

For the board made for testing, the number of possible layers was limited to four layers in order to be able to order boards using a cheaper student special offered through the PCB manufacturer[35]. Figure 7.1 shows the layer setup for the PCB. For the testing board, the two internal planes were further split. The first internal plane was split into analog ground and digital ground. The second internal plane was split into the + 3V and -3V power planes. While this method served its purpose and allowed for split grounds and power planes, the necessary layout of the components restricted the split planes from being used in an optimal way. The -3V split plane did not end up being used.

For the final revision of the PCB, the number of layers was increased to six layers. The six layers would consist of the following: signal plane, analog

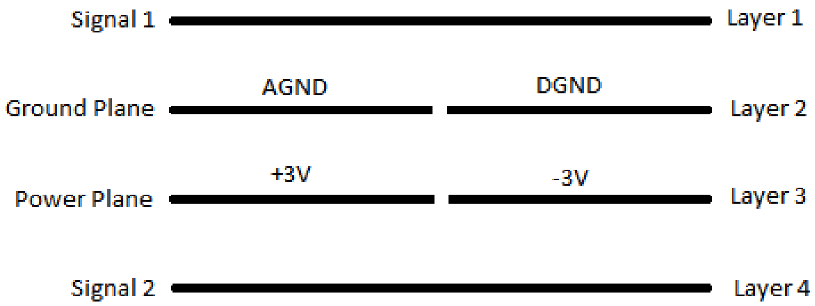


Figure 7.1: Four Layer PCB Depiction

ground plane, + 3V plane, -3V plane, digital ground plane, and a signal plane, shown in Figure 7.2.

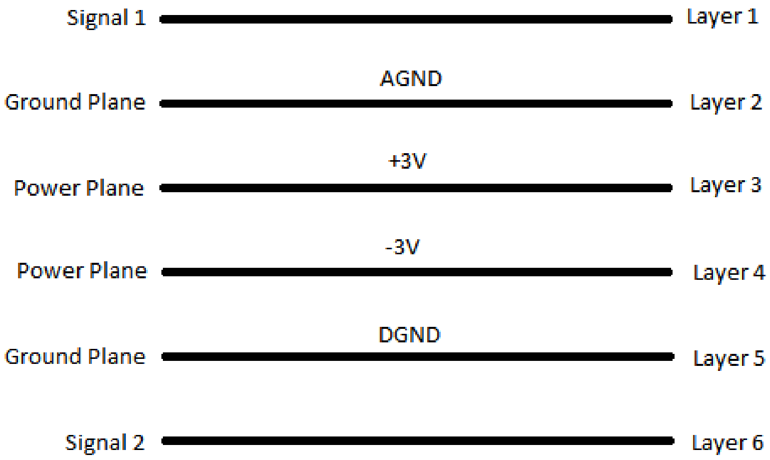


Figure 7.2: Six Layer PCB Depiction

Using six layers would allow each plane to span across the entire board and would get rid of roundabout connections made on the top and bottom signal planes of the board. Unfortunately, the cost of the board would increase compared to the four layer board, but the increase in effectiveness of the board outweighs the increase in costs.

7.2 Prototype Testing Process

The first and most basic test that was completed was a functionality test of the signal chain of the system. Through testing the analog front end of the circuit and wireless test separately, we were able to isolate any issues within the system, allowing for easier debugging. After testing the individual blocks of the system, the entire system was tested as a whole.

After testing the functionality of the system as a whole, it was necessary to do further testing to assure that the specifications of our system were all met. These include the bandwidth of the system, the amount of noise within the system, wireless capabilities, and portability. As the system was already tested for wireless functionality and we chose to use a 12 bit ADC, both of these specifications were already successfully met. As for the portability of the system, the prototype is self contained and takes a matter of seconds to set up before testing, which is considerably more simple to use than existing EEG units. The only specifications that still needed to be verified were the system noise and bandwidth. It is important to know the noise of the system because our unit aims to have as accurate of a system as prior EEGs without the use of expensive electrodes or excessive application methods of these electrodes such as pastes and adhesion. If our system maintains a low level of noise with the inexpensive electrodes, it will mean that we have successfully achieved the goal of this project.

In order to test for the noise within the system, the inputs can just be grounded and the output data collected and observed in Matlab or Excel. The sigma of the output codes will indicate the amount of noise present in the system. This test was not performed, but in a general demo of the prototype, there seemed to be no visually noticeable noise.

7.2.1 Analog Front-End Test

Before the micro controller and the XBee were added to the PCB, the front-end of the system was tested. First, a function generator that was set to output a small sine wave with an amplitude of $20\mu V$. The signal was then scaled down to 200 μV using a voltage divider to attenuate the signal to 1/100 of the function generator signal. An op amp configured for an inverting gain of 1 was used to allow the use of differential signals. An oscilloscope was used to determine if the output of the signal was amplified by the expected gain of 10,000. Because a sine wave is more easily identifiable than an actual EEG signal, testing was continued with a sine wave. Due to tolerances in parts, additional signal loss due to capacitances and inductances created on the PCB, and slight variations in phase between one differential signal and then other, the outputted signal had a peak to peak amplitude of 1.54 V instead of 2 V.

Figure 7.3 is a screen capture of the signal at various stages of the analog front end. The first waveform shows the output of the function generator. The second waveform shows the signal after the instrumentation amplifier set to a gain of approximately 1010 V/V. Finally the third waveform shows the output of the op amp to the ADC.

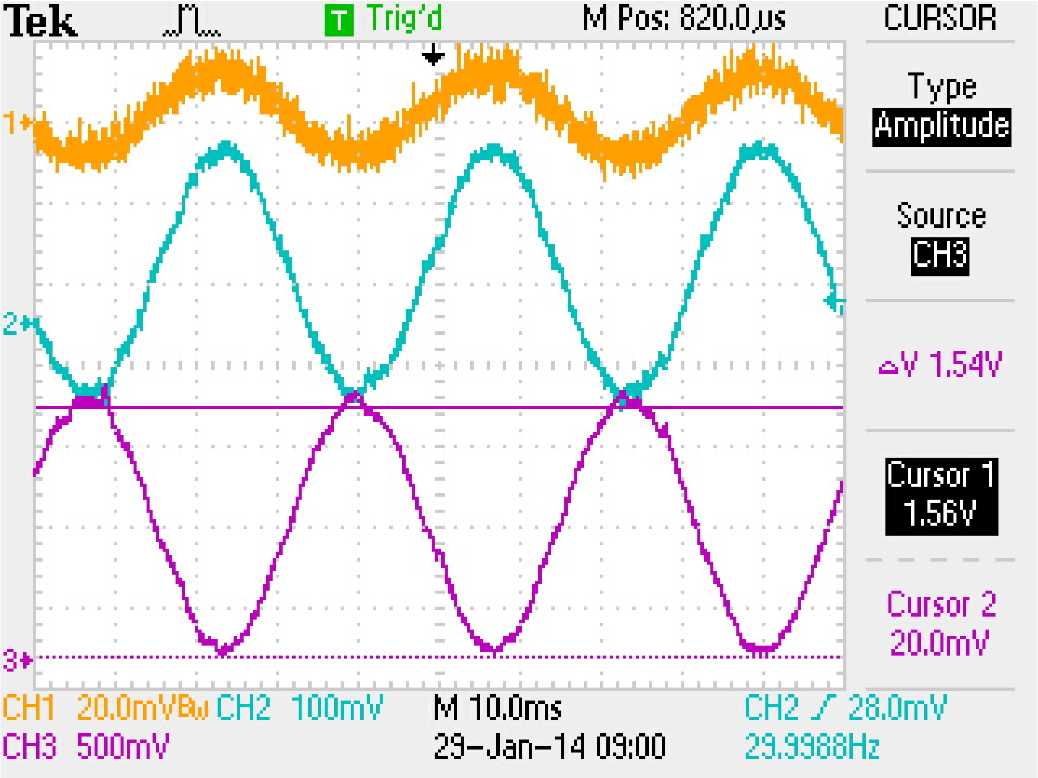


Figure 7.3: Oscillogram of Analog Front End

7.2.2 Digital and Wireless Test

To make sure that the data was being correctly transmitted to the computer, the micro controller was programmed similar to the Arduino as an initial test. A counter was created so that the transmitting XBee was sending 0 to 4096 to the receiving XBee, and a clear triangle wave was visible in LabView. Once this was functional, the full signal chain was ready to be completed. The code for this test is shown in Appendix E.2. The final code only needed to be modified slightly from this code, so the lines for this test are commented out and labeled. Figure 7.4 shows the LabView output, where the triangle wave clearly goes from 0 to 2.5V, the expected range of the ADC.

7.2.3 Full System Test

Both sections of the signal chain were assembled on the PCB for the full system test. Again, the function generator was setup first to verify the output in LabView and that the micro controller code was working properly. Then, the function generator was attached to the PCB so that signal would be displayed in LabView.

Figure 7.5 shows the first set of full results. Having already demonstrated that a clean signal can be obtained at the output of the op amp, a clean signal at the receiving end of the wireless block needed to be observed.

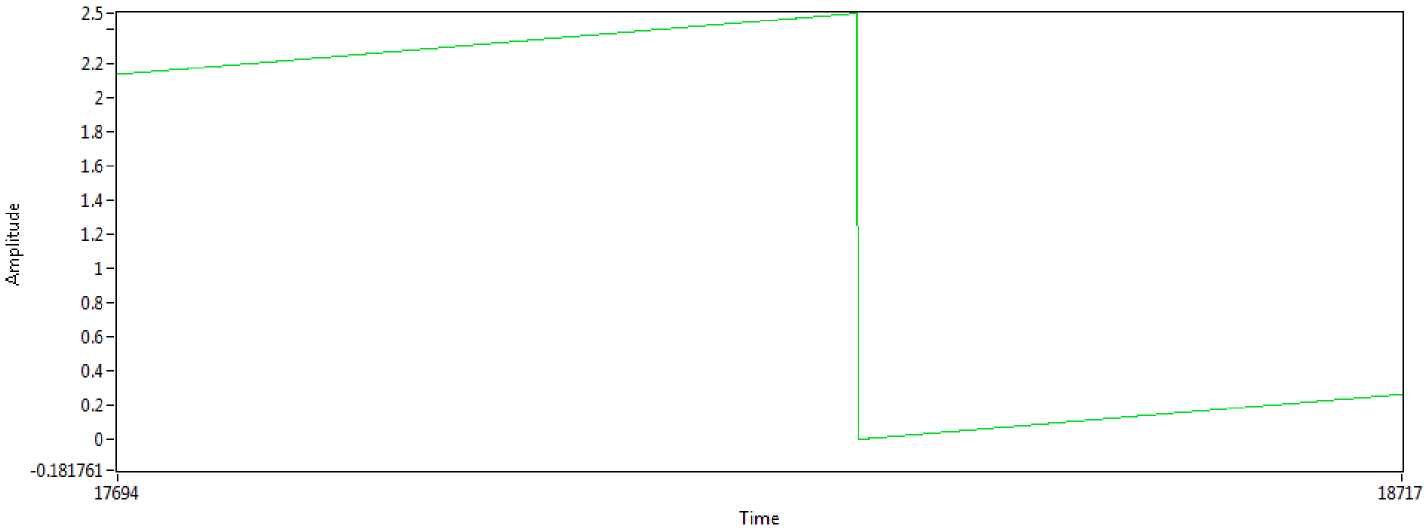


Figure 7.4: LabView Output Results for the Digital and Wireless Test

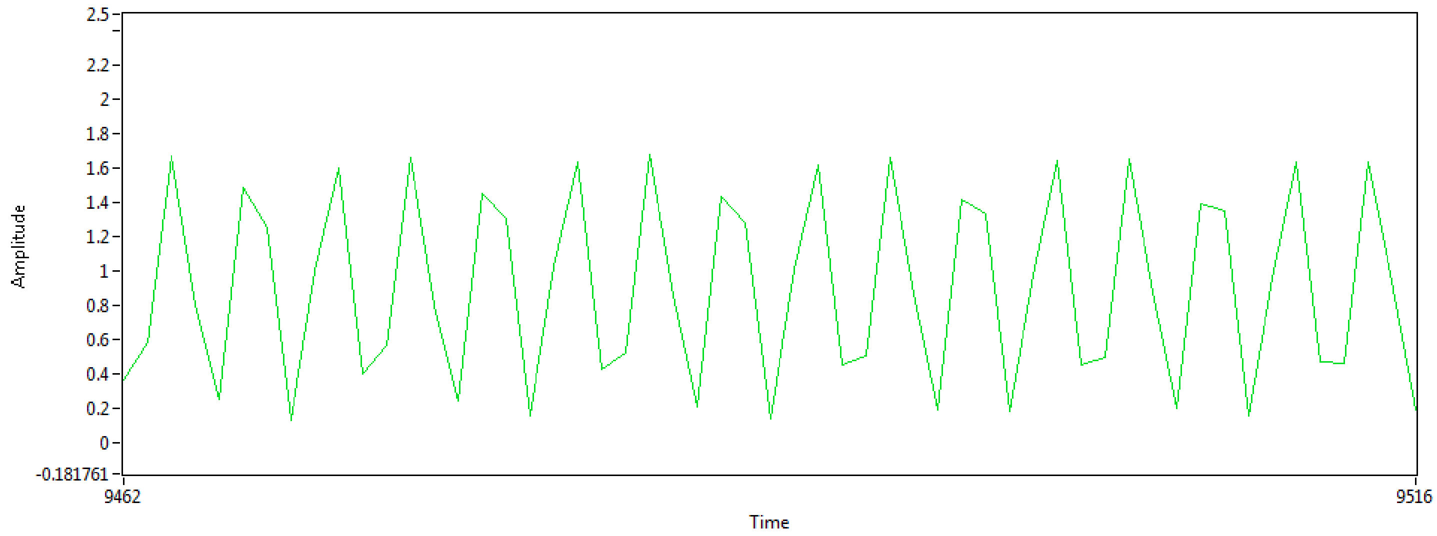


Figure 7.5: LabView Output Results for Signal

It is clear from the initial LabView results that some of the signal is lost through the wireless transmission of the signal. The final micro controller code can be found in Appendix E.2.

7.2.4 Unexpected Results

Although the prototype worked well, some results were either not as expected or were not to our standards. These results led to some changes for the next revision of the design. The design was originally made to allow for flexibility in testing. Since noise was found to be greater than planned, yet still less than what was needed in order to observe a reliable output signal, there were multiple options for the next revision. The design did not have to be changed, but could be if noise were still a big concern. If noise was a concern or we wanted to attempt to further decrease the noise, the range of the corner frequencies could be adjusted to limit noise from outside sources.

Another unexpected result was due to the internal reference of the microcontroller being difficult to change and manipulate. Instead of 2.0 V, it was 2.5 V, and an external reference would need to be able to overdrive this internal reference. This difference in the reference voltage made the quantized ADC signal have a different scale factor. Instead of 1 bit representing $488\mu V$, it now represented $610\mu V$.

Finally a few revisions were suggested that aim to make the design easier to use. One simple aspect that would improve the board is the addition of an on and off switch. This would allow the device to be turned on and off in between use, maximizing the life of the battery. Another change to be made would simplify the production of the device. The previous revision did not allow the microcontroller to easily be programmed on the board. An updated revision would change this by adding four pins for a UART connection. Then, the cable provided with the other evaluation board for the micro controller could be used with our PCB.

Chapter 8

Future Work and Recommendations

In this section we will review what further steps could be taken with this project, specifically addressing any of the issues that were encountered and how they could be solved with more time and resources.

8.1 Wireless Data Transmission

There were some issues with the XBee serial interface and LabView. When looking at the data being received by the XBee in a normal COM port window, it seemed to be coming in as expected. However, there would be missing data packets or incorrect packets when plotting the data real time in

LabView. Since digital and wireless communications are not our expertise, it was very difficult to try and locate the source of the problem. Future implementations will hopefully be able to overcome this problem, or find an even better solution of plotting the data in real time. Additionally, some sort of flow control could be implemented as well as a way of error checking to know if a data packet that was sent was not received.

8.2 Additional Filtering

The band-pass filtering before the in-amp was used to remove DC offset before it saw the gain of the in-amp and op-amp. While this method seemed to be the best option, those trying to make a similar device might try making the first gain stage much smaller and do most of the analog filtering at the output of the instrumentation amplifier so that there is no way to ruin the CMR of the in-amp. In addition, digital filtering might be a consideration to obtain more precise filters and to have a filtering option if needed after the signal reaches the ADC.

8.3 Power Consumption and Prototype Expansion

Most of the power consumption in this device comes from the transmitting XBee. We attempted to reduce the transmit power with one of the programmable settings, but this seemed to not change the transmit current at all. An alternative method to sending the data that uses less power or has a lower programmable output power could be implemented in a future prototype.

Additionally, if desired, this device could be expanded so that more electrodes and front-ends could be added, using different channels of the ADC. There are several ADC channels still available on the microcontroller, and the non-invasive electrodes could still be used. The design could take advantage of the quad buffer design in order to save space. Space could also be saved on the microcontroller, the XBee and then batteries. With a slight increase in size, mostly due to the filtering of the analog front end, the prototype could become more accurate.

While the three electrodes our device uses are very convenient, someone who still wanted a wider range of readings around the skull would also be able to use this device. In addition, an increase from 3 electrodes to 5 electrodes does not cause a significantly increased level of convenience. Eventually, a single IC that has all components in one package could be built.

Chapter 9

Conclusions

The goal of this project was to create a low-cost, portable EEG monitoring device that took advantage of the $20T\Omega$ input impedance of the AD8244 FET buffer for a high signal-to-noise ratio while being able to still use cheap, non-invasive electrodes. During the design process, the major issues encountered were noise removal, particularly 60Hz noise, power consumption due to the wireless and digital components of the prototype, and finally the wireless interface with the computer. While the filter components were matched as much as possible in order to maintain the common mode rejection of the instrumentation amplifier, noise was still able to pass through this filter and be amplified through the signal chain. An expertise in the digital area may have helped to reduce power consumption by programming the micro controller and XBee transmitter to sleep when not processing data. In any case, these devices were still very power hungry, so an alternative low power solution might be considered for a future prototype. Finally, the problem of the missing data packets at the receiving XBee could not be fixed with LabView, so an alternative program was found to read the incoming data and plot it real-time.

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Appendix A

IRB Application and Paperwork

In order to complete human testing, an application must be submitted to the Institutional Review Board (IRB) for review. Along with the application that can be found on their website (http://www.wpi.edu/offices/irb/forms.html), supplemental material including the Informed Consent form, a brief (less than 5 pages) overview of the project and the procedure to be completed, and a certificate from the online human testing training course should be provided. Samples of the Informed Consent form and the link to take the online training course can all be found under the expedited forms link.

WORCESTER POLYTECHNIC INSTITUTE
Institutional Review Board
Application for Approval to Use Human Subjects in Research

WPI IRB use only
IRB # ________
Date: ________

This application is for: *(Please check one)* ☒ Expedited Review ☐ Full Review

Principal Investigator (PI) or Project Faculty Advisor: (*NOT a student or fellow; must be a WPI employee)* WPI IRB use only

Name: John A. McNeill Tel No: 1-508-831-5567 E-Mail Address: mcneill@ece.wpi.edu ☐

Department: Electrical and Computer Engineering

Co-Investigator(s): *(Co-PI(s)/non students)*

Name: ________ Tel No: ________ E-Mail Address: ________ ☐

Name: ________ Tel No: ________ E-Mail Address: ________ ☐

Student Investigator(s):

Name: Jennifer Legaspi Tel No: 315-409-8057 E-Mail Address: jlegaspi@wpi.edu ☐

Jordyn Rombola 860-280-5540 jrombola@wpi.edu

Name: Allison Thibault Tel No: 508-642-1704 E-Mail Address: allietbo53@wpi.edu ☐

Check if: ☒ **Undergraduate project** *(MQP, IQP, Suff., other)* MQP

☐ **Graduate project** *(M.S. Ph.D., other)* ________

Has an IRB ever suspended or terminated a study of any investigator listed above?
No ☒ Yes ☐ *(Attach a summary of the event and resolution.)*

Vulnerable Populations: The proposed research will involve the following (Check all that apply):
pregnant women ☐ human fetuses ☐ neonates ☐ minors/children ☐ prisoners ☐
students ☒ individuals with mental disabilities ☐ individuals with physical disabilities ☐

Collaborating Institutions: (*Please list all collaborating Institutions.)*

Locations of Research: (*If at WPI, please indicate where on campus. If off campus, please give details of locations.)*
Atwater Kent, 317B

Project Title: Analog Integrated Circuit Applications

Funding: *(If the research is funded, please enclose one copy of the research proposal or most recent draft with your application.)*

Funding Agency: ________ WPI Fund: ________

Human Subjects Research: *(**All** study personnel having direct contact with subjects **must** take and pass a training course on human subjects research. There are links to web-based training courses that can be accessed under the Training link on the IRB web site http://www.wpi.edu/offices/irb/training.html. **The IRB requires a copy of the completion certificate from the course or proof of an equivalent program**.)*

Anticipated Dates of Research:

Start Date: 10/29/2013 Completion Date: 12/19/2013

Instructions: Answer all questions. If you are asked to provide an explanation, please do so with adequate details. If needed, attach itemized replies. Any incomplete application will be returned.

1.) Purpose of Study: *(Please provide a concise statement of the background, nature and reasons for the proposed study. Insert below using non-technical language that can be understood by non-scientist members of the IRB.)*

We want to measure brain waves by placing non-invasive surface electrodes on the subject's head in order to demonstrate the performance of a portable EEG measurment device.

2.) Study Protocol: *(Please attach sufficient information for effective review by non-scientist members of the IRB. Define all abbreviations and use simple words. Unless justification is provided this part of the application must not exceed 5 pages. Attaching sections of a grant application is not an acceptable substitute.)*

A.) For **biomedical, engineering and related research**, please provide an outline of the actual experiments to be performed. Where applicable, provide a detailed description of the experimental devices or procedures to be used, detailed information on the exact dosages of drugs or chemicals to be used, total quantity of blood samples to be used, and descriptions of special diets.

B.) For applications in the **social sciences, management and other non-biomedical disciplines** please provide a detailed description of your proposed study. Where applicable, include copies of any questionnaires or standardized tests you plan to incorporate into your study. If your study involves interviews please submit an outline indicating the types of questions you will include.

C.) If the study involves **investigational drugs or investigational medical devices**, and the PI is obtaining an Investigational New Drug (IND) number or Investigational Device Exemption (IDE) number from the FDA, please provide details.

D.) Please note if any **hazardous materials** are being used in this study.

E.) Please note if any **special diets** are being used in this study.

3.) Subject Information:

A.) Please provide the exact number of subjects you plan to enroll in this study and describe your subject population. *(eg. WPI students, WPI staff, UMASS Medical patient, other)*

Males: Up to 3 Females: Up to 3 Description: Members of the MQP will be participating in the study, along with three other volunteers who are previous/current WPI students.

B.) Will subjects who do not understand English be enrolled?
No ☒ Yes ☐ *(Please insert below the language(s) that will be translated on the consent form.)*

C.) Are there any circumstances under which your study population may feel coerced into participating in this study?
No ☒ Yes ☐ *(Please insert below a description of how you will assure your subjects do not feel coerced.)*
Each MQP team member has already volunteered, and when explaining the project to the other participants, they were eager to volunteer.

D.) Are the subjects at risk of harm if their participation in the study becomes known?
No ☒ Yes ☐ *(Please insert below a description of possible effects on your subjects.)*

E.) Are there reasons for excluding possible subjects from this research?

No ☐ Yes ☒ *(If yes, please explain.)*
Not many subjects are needed, so testing the prototype on ourselves makes the most sense. It is a two-term project, so taking more time to find other subjects would be wasteful.

F.) How will subjects be recruited for participation? *(Check all that apply.)*

☐ Referral: *(By whom)* ______________________

☒ Other: *(Identify)* Team members and other current/former WPI students volunteering

☐ Database: *(Describe how database populated)*

F.) Have the subjects in the database agreed to be contacted for research projects? No ☐ Yes ☐ N/A ☐

☐ Direct subject advertising, including: *(Please provide a copy of the proposed ad. All direct subject advertising must be approved by the WPI IRB prior to use.)*

☐ Newspaper
☐ Radio
☐ Television
☐ Internet
☐ Bulletin board
☐ Flyers
☐ Letters
☐ E-mail

G.) Are the subjects being paid for participating? *(Consider all types of reimbursement, ex. stipend, parking, travel.)*
No ☒ Yes ☐ *(Check all that apply.)* ☐ Cash ☐ Check ☐ Gift certificate ☐ Other: ________________
Amount of compensation ____________________________

4.) Informed Consent:

A.) Who will discuss the study with and obtain consent of prospective subjects? *(Check all that apply.)*
☒ Principal Investigator ☐ Co-Investigator(s) ☒ Student Investigator(s)

B.) Are you aware that subjects must read and sign and Informed Consent Form prior to conducting any study-related procedures and agree that all subjects will be consented prior to initiating study related procedures?	No ☐ Yes ☒
C.) Are you aware that you must consent subjects using only the IRB-approved Informed Consent Form?	No ☐ Yes ☒
D.) Will subjects be consented in a private room, not in a public space?	No ☐ Yes ☒
E.) Do you agree to spend as much time as needed to thoroughly explain and respond to any subject's questions about the study, and allow them as much time as needed to consider their decision prior to enrolling them as subjects?	No ☐ Yes ☒
F.) Do you agree that the person obtaining consent will explain the risks of the study, the subject's right to decide not to participate, and the subject's right to withdraw from the study at any time?	No ☐ Yes ☒
G.) Do you agree to either 1.) retain signed copies of all informed consent agreements in a secure location for at least three years or 2.) supply copies of all signed informed consent agreements in .pdf format for retention by the IRB in electronic form?	No ☐ Yes ☒

(If you answer No to any of the questions above, please provide an explanation.)

5.) Potential Risks: *(A risk is a potential harm that a reasonable person would consider important in deciding whether to participate in research. Risks can be categorized as physical, psychological, sociological, economic and legal, and include pain, stress, invasion of privacy, embarrassment or exposure of sensitive or confidential data. All potential risks and discomforts must be minimized to the greatest extent possible by using e.g. appropriate monitoring, safety devices and withdrawal of a subject if there is evidence of a specific adverse event.)*

A.) What are the risks / discomforts associated with each intervention or procedure in the study?
Depending on the electrodes used, they might be very cold to the touch. There is a small risk for electric shock, and if latex electrodes are used there is a possibility that someone will have a reaction if they are allergic to latex.

B.) What procedures will be in place to prevent / minimize potential risks or discomfort?
Altering the electrodes could potentially ruin the data to be collected, so the subject will be asked to relax in a comfortable position, and batteries will be used instead of a DC power supply to minimize the risk of electric shock. The subject will need to sign the Informed Consent form which will indicate that the electroded may be latex and that they should not participate if they are allergic. If there is a different type of electrode not made of latex, we can suggest to use that kind instead if the subject still would like to participate. If the subject feels uncomfortable at any time with the test then we will stop the test.

6.) Potential Benefits:

A.) What potential benefits other than payment may subjects receive from participating in the study?

There are no potential benefits other than helping prove the team's system works.

B.) What potential benefits can society expect from the study?

A portable EEG device that is less expensive would be a great improvement in the medical field.

7.) Data Collection, Storage, and Confidentiality:

A.) How will data be collected?

All of the data will be collected electronically as voltage readings from an analog to digital converter.

B.) Will a subject's voice, face or identifiable body features *(eg. tattoo, scar)* be recorded by audio or videotaping?
No ☒ Yes ☐ *(Explain the recording procedures you plan to follow.)*

C.) Will personal identifying information be recorded? No ☒ Yes ☐ *(If yes, explain how the identifying information will be protected. How will personal identifying information be coded and how will the code key be kept confidential?)*

D.) Where will the data be stored and how will it be secured?
The electronic data will be stored by the MQP students in a folder on a computer. Each file name will be labeled by which test number it is (i.e. if the subject is the first to be tested, then the file name will be Test1); no indentifiable information will be used to name the file. No other data will need to be collected about each subject that could be identifiable information. The MQP students will be primerialy responsible for data collection, and the Informed Consent forms for each of the MQP students will be given to the PI (McNeill) who will keep them locked in his private office.

E.) What will happen to the data when the study is completed?
The de-identified data will remain in the MQP folder and with the faculty PI indefinitely. Identifiable information will consist only of the Informed Consent documentation. These documents will be maintained by the PI for three years, and then destroyed.

F.) Can data acquired in the study adversely affect a subject's relationship with other individuals? *(i.e. employee-supervisor, student-teacher, family relationships)*

No.

G.) Do you plan to use or disclose identifiable information outside of the investigation personnel?
No ☒ Yes ☐ *(Please explain.)*

H.) Do you plan to use or disclose identifiable information outside of WPI including non-WPI investigators?
No ☒ Yes ☐ *(Please explain.)*

8.) Incidental findings: *In the conduct of information gathering, is it possible that the investigator will encounter any incidental findings? If so, how will these be handled? (An incidental finding is information discovered about a subject which should be of concern to the subject but is not the focus of the research. For example, a researcher monitoring heart rates during exercise could discover that a subject has an irregular heartbeat.)*

None anticipated.

9.) Deception: *(Investigators must not exclude information from a subject that a reasonable person would want to know in deciding whether to participate in a study.)*

Will the information about the research purpose and design be withheld from the subjects?
No ☒ Yes ☐ *(Please explain.)*

10.) Adverse effects: *(Serious or unexpected adverse reactions or injuries must be reported to the WPI IRB within 48 hours using the IRB Adverse Event Form found out at http://www.wpi.edu/offices/irb/forms.html. Other adverse events should be reported within 10 working days.)*

What follow-up efforts will be made to detect any harm to subjects and how will the WPI IRB be kept informed?
If any harm comes to a subject, one of the student investigators will contact the WPI IRB office immediately after the incident.

11.) Conflict of Interest: *(A conflict of interest occurs when an investigator or other key personnel in a study may enjoy material benefits based on study results. Relationships that give rise to a conflict of interest or the appearance of a conflict of interest must be disclosed in the informed consent statement provided to study subjects. More information, including examples of relationships that require disclosure and those that do not, can be found here.)*

A.) Do any of the investigators listed on this application have a potential or actual conflict of interest with regard to this study?
a. Investigator (name) _John McNeill________________ No ☒ Yes ☐
b. Investigator (name) _All MQP Students________________ No ☒ Yes ☐
c. Investigator (name) _ ________________ No ☐ Yes ☐

B.) If any of the answers to 11A. are "Yes," please attach an explanation of the nature of the conflict to this application and identify appropriate language for use in the consent form. Examples of consent language are found on the IRB website, here.

C.) Does each investigator named above have a current WPI conflict of interest disclosure form on file with the appropriate supervisor/department head? No ☐ Yes ☒

D.) Do any of the investigators' COI forms on file with WPI contain information regarding this research?
No ☒ Yes ☐
a. If "Yes," identify the investigator(s) ________________________

12.) Informed consent: *(Documented informed consent must be obtained from all participants in studies that involve human subjects. You must use the templates available at http://www.wpi.edu/offices/irb/forms.html to prepare these forms.* ***Informed consent forms must be included with this application.*** *Under certain circumstances the WPI IRB may waive the requirement for informed consent.)*

Investigator's Assurance:

I certify the information provided in this application is complete and correct.

I understand that I have ultimate responsibility for the conduct of the study, the ethical performance of the project, the protection of the rights and welfare of human subjects, and strict adherence to any stipulations imposed by the WPI IRB.

I agree to comply with all WPI policies, as well all federal, state and local laws on the protection of human subjects in research, including:

- ensuring the satisfactory completion of human subjects training.
- performing the study in accordance with the WPI IRB approved protocol.
- implementing study changes only after WPI IRB approval.
- obtaining informed consent from subjects using only the WPI IRB approved consent form.
- promptly reporting significant adverse effects to the WPI IRB.

Signature of Principal Investigator John A McNeill Date 10/1/2013

Print Full Name and Title John A. McNeill Associate Professor

Please return a signed hard copy of this application to the WPI IRB c/o Ruth McKeogh 2nd Floor Project Center
Or email an electronic copy to irb@wpi.edu
If you have any questions, please call (508) 831-6699.

Portable Brain Wave Sensing (EEG) Device

Introduction

The purpose of this experiment is to verify the performance of a portable electroencephalography (EEG) device by comparing the measured brain waves to the different types of waves the brain produces. By taking measurements of the waves with the noise we are expecting to see and then taking measurements eliminating as much noise as possible, we can verify that the waves being measured are true and accurate.

Protocol

After securing written informed consent, the subjects will be seated in a chair or asked to lie down while the testing is being performed. In each experiment, the subject will also be asked to wash their face, particularly their forehead where the electrodes will be placed. The electrode system in place will either be clinical EEG disposable electrodes (possibly made of latex) or reusable electrodes (not made of latex) that will need to be sanitized before the next use. If possible, multiple pairs will be purchased so a fresh pair can be used for each participant. They will be secured to the subject using medical tape or soft bandaging. Some electrodes may have self-adhesive. If time permits, the electrodes will be affixed to the inside of a hat or a headband so that no bandaging will be necessary.

Two types of measurements will be made during the procedure. A baseline measurement will need to be taken in order to compare signals read. Eye blinking, moving their eyes, speaking, and other body movements will cause unwanted artifacts to appear in the waveforms. By having the participants make these movements first, we can compare the signal taken while they are making these movements to a signal when they are not moving at all. It's impossible for them to stay perfectly still, so we will be able to see what noise from what movement contributes to the output.

Then, subjects will be asked to remain as still as possible and not move their head in order to make sure the reading is correct and accurate. Different types of music may be played for a short amount of time while a measurement is taken. These will be thirty-second clips of random country, classical, pop, and dub step songs. The subject may also be read a relaxation script. The entire protocol is expected to require not more than one hour of time per subject. Testing will be suspended immediately if the subject expresses any discomfort. The two procedures mentioned will vary between test subjects, i.e. we may not play them any music and just read the relaxation script or we may only play music.

Methods of Analysis

The data will be processed using amplification techniques in order to see the small signals being read and filtering techniques that will remove any noise within the signal. The system will be connected to a computer to display the signals onto the screen, where they will be compared with typical waveforms.

Informed Consent Agreement for Participation in a Research Study

Investigator: John A. McNeill

Contact Information: **ECE Department**
WPI
100 Institute Road
Worcester, MA 01609
Tel. 508-831-5567, Email: mcneill@ece.wpi.edu

Title of Research Study: Analog Integrated Circuit Applications: Acquisition of Brain Wave Signals (Electroencephalography) using a Small, Portable Device

Introduction

You are being asked to participate in a research study. Before you agree, however, you must be fully informed about the purpose of the study, the procedures to be followed, and any benefits, risks or discomfort that you may experience as a result of your participation. This form presents information about the study so that you may make a fully informed decision regarding your participation.

Purpose of the study:

In this experiment, we will investigate the electrical activity within your brain. This information will be used to confirm the functionality of the device used to measure the waves by being able to detect different types of brain waves. A surface electrode will measure this electrical activity within the body.

Procedures to be followed:

You will be seated or asked to lie down for most of the experiment. The surface recording electrodes will be placed on your forehead to monitor the electrical activity seen at the skin. We may tape them onto your forehead, or ask you to wear a hat with the electrodes attached inside. We will ask you to lay/sit still and not move your eyes for certain parts of the experiment when the measurements are being recorded. This task will last no more than a minute. Rest (about 3-5 minutes) will be provided between each task. Your participation will last for a total of 1 hour.

Risks to study participants:

There is some possibility of minor discomfort due to the electrodes being placed on your forehead, i.e. the tape that will be used to secure them might cause some irritation. There is also a chance for minor electric shock, which will be mostly eliminated by using batteries instead of a DC supply. Some electrodes may be made of latex. If you have a latex allergy, please inform the person conducting this study. There will be another pair of electrodes not made of latex that will be used.

Benefits to research participants and others:

There is no direct benefit to you.

Record keeping and confidentiality:
Records of your participation in this study will be held confidential so far as permitted by law. However, the study investigators, the sponsor or it's designee and, under certain circumstances, the Worcester Polytechnic Institute Institutional Review Board (WPI IRB) will be able to inspect and have access to confidential data that identify you by name. Any publication or presentation of the data will not identify you.

Compensation or treatment in the event of injury: In the unlikely event of physical injury resulting from participation in the research, you understand that medical treatment may be available from WPI, including first aid emergency care, and that your insurance carrier may be billed for the cost of such treatment. No compensation for medical care can be provided by WPI. You further understand that making such medical care available, or providing it does not imply that such injury is the fault of the investigators. You do not give up any of your legal rights by signing this statement.

For more information about this research or about the rights of research participants, or in case of research-related injury, contact:
Prof. John A. McNeill, ECE Department, WPI, 100 Institute Road, Worcester, MA (Tel. 508-831-55677). You may also contact the IRB Chair (Professor Kent Rissmiller, Tel. 508-831-5019, Email: kjr@wpi.edu) and the University Compliance Officer (Michael J. Curley, Tel. 508-831-6919, Email: mjcurley@wpi.edu).

Your participation in this research is voluntary. Your refusal to participate will not result in any penalty to you or any loss of benefits to which you may otherwise be entitled. You may decide to stop participating in the research at any time without penalty or loss of other benefits. The project investigators retain the right to cancel or postpone the experimental procedures at any time they see fit. Data obtained in this experiment will become the property of the investigators and WPI. If you withdraw from the study, data already collected from you will remain in the study.

By signing below, you acknowledge that you have been informed about and consent to be a participant in the study described above. Make sure that your questions are answered to your satisfaction before signing. You are entitled to retain a copy of this consent agreement.

____________________________ Date: ____________________
Study Participant Signature

____
Study Participant Name (Please print)

_____________________________________ Date: ____________________
Signature of Person who explained this study

Appendix B

RFI Filter Cutoff Frequency

To derive the cutoff frequency for the RFI filter, half circuit analysis was used on the RFI filter to obtain the circuit shown in Figure B.1. In order to split the circuit in half, this would mean splitting the capacitor in "half." Since capacitors add like resistors in parallel, we just double the C_1 capacitor value so that when it is added with the capacitor from the other half, it will be the correct value.

Letting $s = j\omega$, writing the impedance divider equation gives:

$$\frac{V_{out}}{V_{in}} = \frac{\frac{2}{sC_1}}{\frac{2}{sC_1} + R_1} \tag{B.1}$$

Multiplying top and bottom by sC_1 and dividing everything by 2 gives:

$$\frac{V_{out}}{V_{in}} = \frac{1}{1 + \frac{sC_1R_1}{2}} \tag{B.2}$$

To solve for the cutoff frequency, we find $\left|\frac{V_{out}}{V_{in}}\right|$ of the transfer function when it is equal to $\frac{1}{\sqrt{2}}$:

$$\frac{\sqrt{1^2}}{\sqrt{1^2 + (\frac{sC_1R_1}{2})^2}} = \frac{1}{\sqrt{2}} \tag{B.3}$$

Whatever is under the radical in the denominator must equal to 2 so:

$$2 = 1 + \left(\frac{2\pi f_{3dB} C_1 R_1}{2}\right)^2 \tag{B.4}$$

Solving for f, we find F_{3dB} to be:

$$f_{3dB} = \frac{1}{\pi C_1 R_1} \tag{B.5}$$

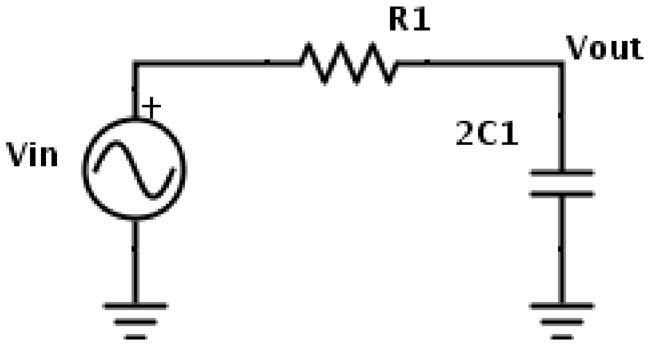

Figure B.1: RFI Filter Half Circuit

Appendix C

High-Pass and RFI Filter Transfer Function

There is no buffering between the high-pass filter and the RFI filter before the in-amp, so there is some interaction between the two filters that affects the actual poles. The equations in this appendix show the step-by-step procedure to finding these poles. The transfer function and poles of the circuit in Figure C.1 is what we are solving for. The circuits in this section with C_2 are actually $2C_2$ and this is reflected in the calculations.

Using $s = jw$, $\tau_1 = R_1C_1$ and $\tau_2 = R_2C_2$:

$$R_{thev} = \frac{R_1}{1 + s\tau_1} \tag{C.1}$$

$$V_{thev} = \frac{s\tau_1}{1 + s\tau_1} \tag{C.2}$$

The Thevenin Equivalent model is shown in Figure C.2.

$$V_{out} = \frac{\frac{2}{sC_2}}{\frac{2}{sC_2} + R_2 + \frac{R_1}{1+s\tau_1}} \left(\frac{s\tau_1}{1 + s\tau 2} \right) \tag{C.3}$$

Multiplying top and bottom by sC_2 gives:

$$V_{out} = \frac{2}{2 + s\tau_2 + \frac{sR_1C_2}{1+s\tau_1}} \left(\frac{s\tau_1}{1 + s\tau 2} \right) \tag{C.4}$$

Further simplification reveals:

$$V_{out} = \frac{2s\tau_1}{s^2\tau_1\tau_2 + 2s\tau_1 + s\tau_2 + sR_1C_2 + 2} \tag{C.5}$$

The denominator is then regrouped to $ax^2 + bx + c$ form:

$$V_{out} = \frac{2s\tau_1}{s^2\tau_1\tau_2 + s(2\tau_1 + \tau_2 + R_1C_2) + 2} \tag{C.6}$$

The top and bottom are then divided by the a term:

$$V_{out} = \frac{\frac{2s\tau_1}{\tau_1\tau_2}}{s^2 + s\frac{(2\tau_1+\tau_2+R_1C_2)}{\tau_1\tau 2} + \frac{2}{\tau_1\tau_2}} \tag{C.7}$$

Knowing:

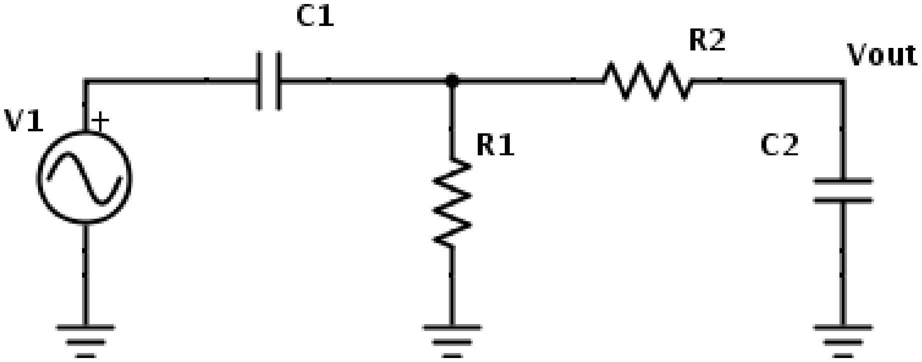

Figure C.1: High-Pass-RFI filter circuit

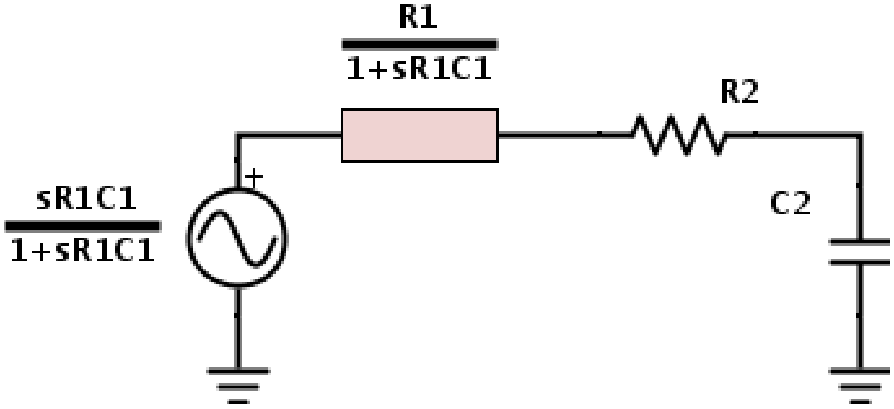

Figure C.2: Thevenin Equivalent High-Pass-RFI filter circuit

$$s^2 - (p_1 + p_2) + p_1 p_2 \tag{C.8}$$

Assuming $p_2 >> p_1$:

$$p_2 = \frac{2\tau_1 + \tau_2 + R_1 C_2}{\tau_1 \tau_2} \tag{C.9}$$

$$p_1 p_2 = \frac{1}{\tau_1 \tau_2} \tag{C.10}$$

Plugging Equation C.9 into Equation C.10 we find p_1 to be:

$$p_1 = \frac{1}{2\tau_1 + \tau_2 + R_1 C_2} \tag{C.11}$$

Then, each of the poles can be rewritten to show the original pole multiplied by an "interaction term":

$$p_1 = \underbrace{\frac{1}{\tau_1}}_{Original Pole} \underbrace{\left(\frac{1}{2 + \frac{\tau_2}{\tau_1} + \frac{C_2}{C_1}}\right)}_{Interaction Term} \tag{C.12}$$

$$p_2 = \underbrace{\frac{1}{\tau_2}}_{Original Pole} \underbrace{\left(2 + \frac{\tau_2}{\tau_1} + \frac{C_2}{C_1}\right)}_{Interaction Term} \tag{C.13}$$

Appendix D

Noise Analysis

There are three noise sources due to the front-end filtering, including the thermal noise from the resistors and the input bias current for the instrumentation amplifier. The circuit in Figure D.1 shows the half circuit for the analysis. The transfer function for each of the sources with respect to the output (V_A) will be found. The circuits in this section with C_2 are actually $2C_2$ and this is reflected in the calculations.

For each source, the other sources were shut off to find the transfer function for only that source.

The Thevenin equivalent was found for the circuit at the node before R_2.

$$R_{thev} = \frac{R_1}{1 + s\tau_1} \tag{D.1}$$

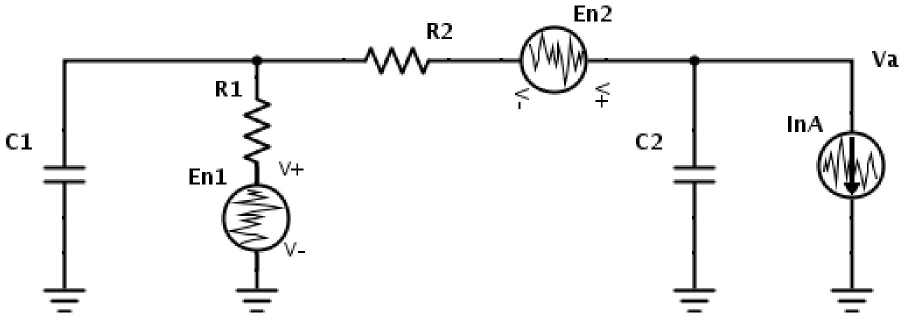

Figure D.1: Noise Analysis Circuit

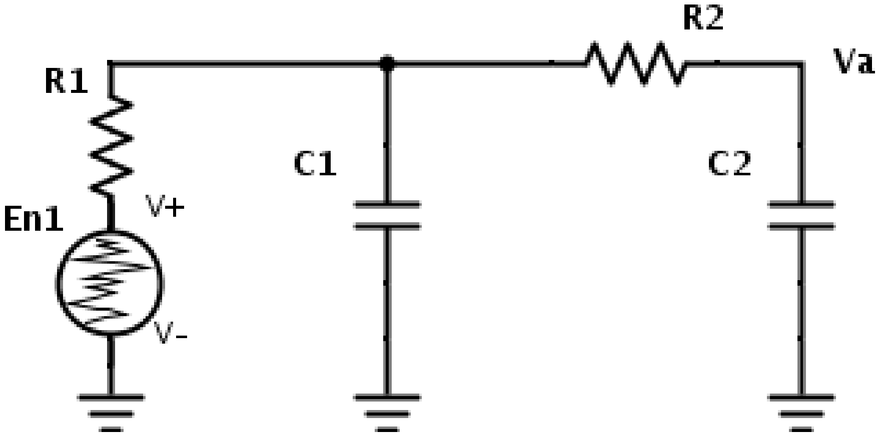

Figure D.2: E_{n1} Noise Analysis Circuit

$$V_{thev} = \frac{1}{1 + s\tau_1} \tag{D.2}$$

Then replacing R_1, C_1, and E_{n1} with these components, we find the total transfer function to be:

$$\frac{V_A}{E_{n1}} = \frac{\frac{2}{sC_2}}{\frac{2}{sC_2} + R_2 + \frac{R_1}{1+s\tau_1}} \left(\frac{1}{1 + s\tau_1} \right) \tag{D.3}$$

Multiplying the numerator and denominator by sC_2 and then distributing everything, we find:

$$\frac{V_A}{E_{n1}} = \frac{2}{s^2\tau_1\tau_2 + s(2\tau_1 + \tau_2 + R_1C_2) + 2} \tag{D.4}$$

Finally, dividing by $\tau_1\tau_2$ we find:

$$\frac{V_A}{E_{n1}} = \frac{\frac{2}{\tau_1\tau_2}}{s^2 + s\frac{2\tau_1+\tau_2+R_1C_2}{\tau_1\tau_2} + \frac{2}{\tau_1\tau_2}} \tag{D.5}$$

So the pole that this noise source should affect is the first pole that makes the "low-pass" part of the band-pass configuration.

$$p_1 = \frac{1}{\tau_1} \left(\frac{2}{2 + \frac{\tau_2}{\tau_1} + \frac{C_2}{C_1}} \right) \tag{D.6}$$

The circuit in Figure D.3 shows the circuit for the second thermal noise source analysis.

The parallel combination of R_1 and C_1 is added in series with R_2 to find the equivalent resistance:

$$R_{eq} = \frac{R_1 + R_2(1 + s\tau_1)}{1 + s\tau_1} \tag{D.7}$$

Then using KCL at the V_A node:

$$\frac{V_A}{E_{n2}} = -\frac{\frac{2}{sC_2}}{\frac{R_1+R_2(1+s\tau_1)}{1+s\tau_1}} \tag{D.8}$$

Multiplying top and bottom by sC_2 and $1 + s\tau_1$ we find:

$$\frac{V_A}{E_{n2}} = -\frac{\frac{2(1+s\tau_1)}{\tau_1\tau_2}}{s^2 + s\left(\frac{\tau_2+R_1C_2}{\tau_1\tau_2}\right)} \tag{D.9}$$

The second noise source should mostly affect the second pole:

$$p_2 = \frac{1}{\tau_2}\left(\frac{\tau_2}{\tau_1} + \frac{C_2}{C_1}\right) \tag{D.10}$$

Finally, the last noise source is the input bias current to the in-amp. Figure D.4 shows the circuit for this noise analysis.

The equivalent resistance was found by combining $((R_1||C_1) + R_2)||C_2$. This equivalent resistance turns out to be:

$$R_{eq} = \frac{\frac{R_1+R_2(1+s\tau_1)}{1+s\tau_1}\left(\frac{2}{sC_2}\right)}{\frac{R_1+R_2(1+s\tau_1)}{1+s\tau_1} + \left(\frac{2}{sC_2}\right)} \tag{D.11}$$

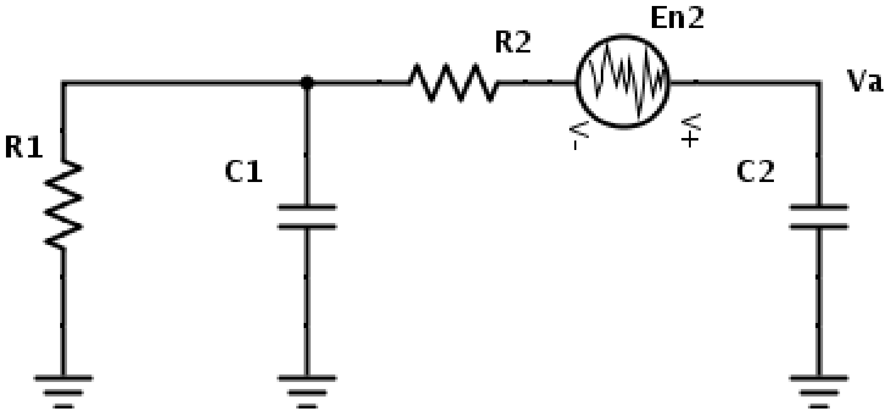

Figure D.3: E_{n2} Noise Analysis Circuit

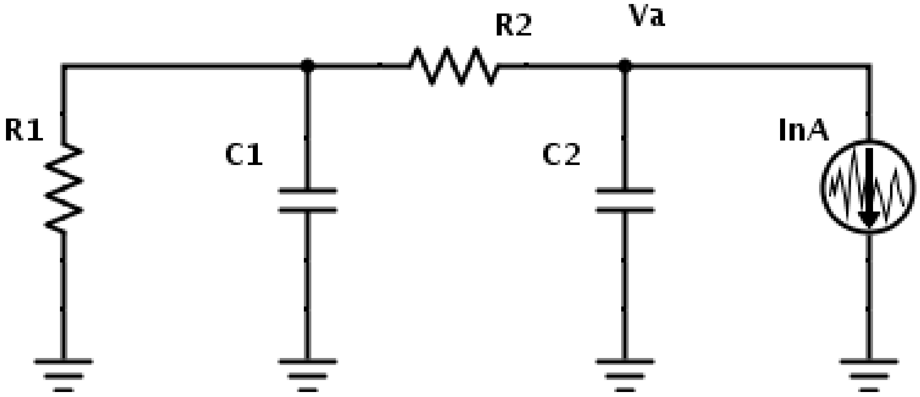

Figure D.4: I_{na} Noise Analysis Circuit

Multiplying the numerator and the denominator by $sC_2(1_s\tau_1)$, we find:

$$\frac{V_A}{I_{na}} = \frac{2R_1 + 2R_2(1 + s\tau_1)}{s^2\tau_1\tau_2 + s(2\tau_1 + \tau_2 + R_1C_2) + 2} \tag{D.12}$$

Dividing everything by $\tau_1\tau_2$ we find the final transfer function to be:

$$\frac{V_A}{I_{na}} = \frac{\frac{2R_1+2R_2(1+s\tau_1)}{\tau_1\tau_2}}{s^2 + s\left(\frac{2\tau_1+\tau_2+R_1C_2}{\tau_1\tau_2}\right) + \frac{2}{\tau_1\tau_2}} \tag{D.13}$$

This noise source sees both poles:

$$p_1 = \frac{2}{2\tau_1\tau_2 + R_1C_2} \tag{D.14}$$

$$p_2 = \frac{2\tau_1 + \tau_2 + R_1C_2}{\tau_1\tau_2} \tag{D.15}$$

Appendix E

Microcontroller Code

E.1 Arduino Code

```
#include <SoftwareSerial.h>
SoftwareSerial xbee(2,3); //rx, tx
Int heartbeat =0;
Int heartbeat2=0;
String sendval;
Void setup()
Serial.begin(57600)
Xbee.begin(19200) //xbee baud rate
Void loop()
```

```
Heartbeat=analogRead(A0); //read from pin 0, ADC input
Sendval=String(heartbeat)+ “\n”; //convert number to string, add new line
Xbee.print(sendval); //send this to the xbee
Delay(10); //do this every 10ms
```

E.2 ADuC7021 Code

```
/************************************************************************
Author : ADI - Apps www.analog.com/MicroConverter
Date : Sept. 2005
File : ADCtimer.c
Hardware : Applicable to ADuC702x rev H or I silicon Currently targeting ADuC7026.
Description : Performs an ADC conversion every 100 us using timer0 overflow alternatively on Channel 0 and 1 sending the results through UART at 9600bps
************************************************************************/
#include <ADuC7021.h> // Include ADuC7021 Header File
void My_IRQ_Function(void); // IRQ Function Prototype
```

```
void senddata(short to_send);

void ADCpoweron(int);

char hex2ascii(char toconv);

void delay(int);

int itoa(int in, char *str, int len);

int leap =0;

//int dummy = 0;

int main (void) {

POWKEY1 = 0x01;

POWCON = 0x00; // 41.78MHz

POWKEY2 = 0xF4;

//ADC configuration

ADCpoweron(20000); // power on ADC

ADCCP = 0x04;

ADCCON = 0x4E2; // start conversion on timer 0 // ADC Config: fADC/2, acq. time = 2 clocks => ADC Speed = 1MSPS

REFCON = 0x01; // connect internal 2.5V reference to VREF pin

// Setup tx & rx pins on P1.0 and P1.1

GP2CON = 0x002;
```

```
// Start setting up UART at 9600bps
COMCON0 = 0x80; // Setting DLAB
COMDIV0 = 0x88;
COMDIV1 = 0x00;
COMCON0 = 0x03; // Clearing DLAB
// for test purposes only GP0CON = 0x10100000; // enable ECLK output on P0.7, and ADCbusy on P0.5
IRQ = My_IRQ_Function; // Specify Interrupt Service Routine
IRQEN = ADC_BIT; // Enable ADC IRQ ( 0x80 )
// timer0 configuration
T0LD = 0x6600; // 26112/2.61125MHz = 10ms
T0CON = 0xC4; // count down, periodic mode
//GP4DAT = 0x04000000; // Configure P4.2 as output
while(1)
{
}
return 0 ; }
/********************************************************************/
/* Interrupt Service Routine */
```

```
/*******************************************************************/
void My_IRQ_Function()
{ if (leap<52224) {
senddata (ADCDAT >> 16);
leap++;
}
else{
T0CLRI = 0x00;
leap=0;
}
return ;
}
void senddata(short to_send)
{
int d = 1000; //initialize integer
char str[6] = {‘0’, ‘0’, ‘0’, ‘0’, ‘0’, ‘0’}; //create string to hold ascii values of adc value
int len = itoa(to_send, str, 6); //changed dummy to to_send (from counter test), convert adc value to ascii
```

```
while(!(0x020==(COMSTA0 & 0x020))){} //send a LF whenever this is true
COMTX = 0x0A;
for(d = 0; d < 4; d++) //this can only be between 0 and 4096 so only 4 places needed
{
while(!(0x020==(COMSTA0 & 0x020))){}
COMTX= str[4 - 1 - d]; //send the character from the string in reverse order
}
while(!(0x020==(COMSTA0 & 0x020))){}
COMTX = 0x0D; // output CR
// dummy++; //increment the dummy variable and then reset when it’s 4096
// if(dummy==0x00001000) // for counter test
// dummy=0x00000000;
}
char hex2ascii(char toconv)
{
if (toconv<0x0A)
```

```
{
toconv += 0x30;
}
else
{
toconv += 0x37;
}
return (toconv);
}
void delay (int length)
{
while (length >=0)
length–;
}
void ADCpoweron(int time)
{ ADCCON = 0x20; // power-on the ADC
while (time >=0) // wait for ADC to be fully powered on
time–;
}
```

```
int itoa(int in, char *str, int len)
{ int c = 0;
while(in != 0 && c < len)
 str[c++] = (in % 10) + ‘0’;
in /= 10;
}
return c;
}
```