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Amplification and Filtering Stage for An In-Home Wireless Polysomnograph

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AMPLIFICATION AND FILTERING STAGE FOR AN IN-HOME WIRELESS POLYSOMNOGRAPH

By

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ABSTRACT

Obstructive sleep apnea is a sleep disorder that tends to be easily treatable but, for most sufferers, is undiagnosed. To lower the economic barriers to screening the design of a portion of a polysomnograph, or obstructive sleep apnea testing tool, is considered. A filtering and amplification stage is theorized, simulated, and constructed for testing. The stage, consisting of three cascaded 1st order Butterworth low pass filters and amplifiers, did not achieve the target signal to noise ratio for a sinusoidal waveform of amplitude 200μv, but showed promising signal to noise ratios for waveforms with amplitudes as low as 600μv. The stage was then designed for implementation on a printed circuit board such that the noise mitigation techniques of printed circuit board implementation would allow successful signal to noise ratios to be achieved.

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List of Abbreviations

ADC: Analog to Digital Conversion

EEG: Electroencephalograph

FFT: Fast Fourier Transform

LPF: Low Pass Filter

OSA: Obstructive Sleep Apnea

PCB: Printed Circuit Board

PSG: Polysomnograph

SNR: Signal to Noise Ratio

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INTRODUCTION

Obstructive Sleep Apnea (OSA) is a sleep disorder that occurs as a result of an obstruction of the airways during sleep [12]. These obstructions can occur for a number of reasons and cause a pause in breathing. These apneas starve the brain of oxygen and cause the patient to wake up temporarily. Although the patient does wake up, this phenomenon is brief to the point of not being noticed by the patient and is referred to as a microarousal. These microarousals can occur as frequently as 30 times per night, or as infrequently as once per night, but frequencies of 10 are not uncommon [12]. Although the microarousals are not noticed or cannot be recalled by the patient, they nevertheless result in a qualitative loss of sleep. In addition to the obvious effects of sleep deprivation such as fatigue and lethargy, other risks exist. Heart disease, obesity, anxiety, and greater risks after surgical procedures are all common problems faced by patients with chronic OSA [8]. Despite the severity of leaving the disorder untreated, the disorder tends to be easily treatable, either with a positive pressure device or a simple plastic mouth insert while sleeping. The disorder remains a problem for many Americans, however. While about 5% of Americans have OSA, as many as 80% of those with OSA are undiagnosed [8, 12]. Especially due to the problems associated with leaving the condition untreated, diagnosis is particularly important. One difficulty when considering OSA diagnosis is that in order to be diagnosed with OSA a patient must attend a hospital sleep center and stay several nights in a row. The difficulty in doing this is that not all hospitals have a sleep center for OSA testing, which could result in patients needing to travel large distances for this testing to take place [8]. Furthermore, the need to stay several nights in this sleep center is also cost prohibitive. Even when a patient would have time off from work, the travel, lodging, and testing costs remain.

Before OSA screening is discussed in detail, it becomes necessary to understand one of the underlying principles of sleep and awareness. Brain waves are placed into different categories depending on their frequency. The wave categories are Delta, Theta, Alpha, Beta, and Gamma waves. Each category of brainwaves is characterized by an increase in frequency. Delta waves are below 4Hz, Theta waves from 4 – 8Hz, Alpha waves from 9 – 13Hz, Beta waves from 14 – 30Hz, and Gamma waves from 30 – 80Hz [11, 14]. Each category of brain waves is associated with a particular state, or level, of consciousness or awareness. Delta waves, for example, are typically indicative of deep, dreamless sleep, whereas Gamma waves are indicative of hyper awareness or focus. As the frequency of the measured brain waves increase, so too does the level of awareness that can be expected in the patient [3, 11]. During sleep, the human brain passes through multiple stages of brain waves, exhibiting Theta waves during Rapid Eye Movement (REM) sleep, and Delta waves during deep, dreamless sleep. Brain wave frequency, then, are used as a metric for reliably determining the sleep state of a patient.

In order to screen for OSA, a device called a Polysomnograph (PSG) is used. This device uses sensors to measure electrical activity in the brain, as well as sensors to measure oxygen levels in the blood, and a strap to measure chest motion and breathing. The Polysomnography electrical measurements on the brain are taken by means of metal cup shaped probes placed on the scalp of a patient. Alternating voltages in the brain capacitively couple with the metal probes to produce corresponding voltages at those probes [23]. The frequency of the voltage signals measured in the brain correspond with sleep state; the presence of higher frequency waves indicates focus or wakefulness, whereas the prevalence of low frequency waves indicates reduced awareness or unconsciousness. A Polysomnography device is similar to an electroencephalography (EEG) device in that brain waves are measured by means of scalp

probes, but several differences exist. One of the most important differences is the number of probes. EEG devices commonly have as many as 32 or 64 probes, whereas PSG devices tend to have fewer, such as 8 or 16 [14]. Microarousals produce an effect on all the sensors of a Polysomnograph. As an obstruction occurs in the airways of a patient, the blood oxygen levels drop, and breathing slows or stops. As the blood oxygen levels drop, a microarousal would then be observed in the brain activity; the observed waves would suddenly increase in frequency. Following this quick increase in brain wave frequency, breathing suddenly resumes and blood oxygen returns to normal. After blood oxygen returns to normal then brain wave frequency returns to ordinary sleeping conditions [12].

There exist a number of different types of probes for performing PSG measurements. Wet and dry probes are the main two categories of probes, but subcutaneous probes exist as well. Subcutaneous probes are not typically used for OSA testing and are therefore not considered. Wet and dry probes differ in that dry probes are prepared and attached to the skin with a gel, and wet probes attach with a saline solution. The purpose of both the gel and saline is to reduce the resistance of the scalp so that a higher voltage will be induced on the probe, and therefore a cleaner reading may be obtained [14]. Wet probes are useful when considering customer rental of a screening device because they are disposable and easy to use. In order to attach wet probes to the skin, a cap is typically affixed and the probes are simply pushed on the back like a button to activate them. Wet probes are also more expensive than dry probes, however. Dry probes, although more difficult to place and prepare, are reusable and cheaper [14].

One possible solution to the inconvenience of OSA screening is the possibility of in-home testing. By being able to screen individuals in their own homes for this disorder, patients could gain greater insight into whether or not they should attend a sleep center. By adding a more

accessible screening layer to the process of OSA treatment, part of the OSA diagnosis process becomes more accessible to many patients. This paper investigates the design of a PSG component such that a PSG device utilizing that component could be more accessible to patients.

EXPERIMENTAL METHODS

One of the most prohibitive components of a Polysomnography (PSG) device is the brain-wave measurement portion of the device. Other needed sensors, like a blood oxygen sensor, are relatively inexpensive. The brain wave measurement portion of the device consists of the following components, which are outlined below in figure 1 [15].

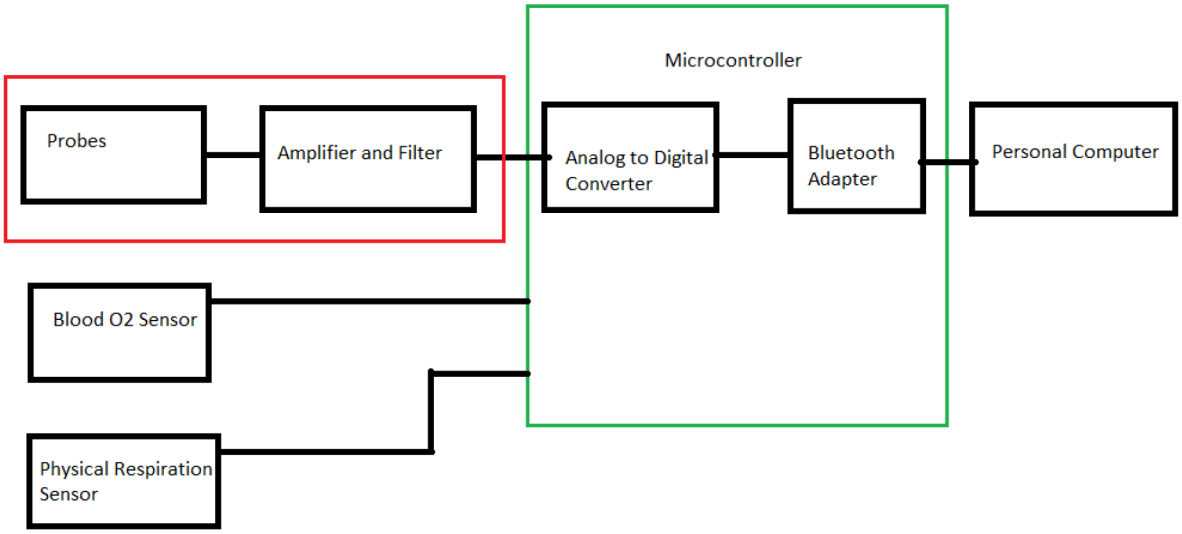

Figure 1: Functional Block Diagram of a Bluetooth Wireless PSG Device

The portion of the diagram in red is specifically responsible for measuring voltage in the brain. PSG or Electroencephalography amplification and signal processing stages typically cost

around $20,000 [6]. In order to increase accessibility, the design of a less expensive system is considered.

The purpose of the amplification and filtering stage is to amplify the probe voltage such that it is within input range of the microcontroller for the device. This stage also filters the input signals based on input signal frequency, which is discussed later. As the microcontroller selected for the project was the BTH-1208LS, a purpose-built Analog to Digital Conversion (ADC) microcontroller, the amplifier gain would be configured according to the requirements of the BTH-1208LS. Probe voltages due to brain activity are extremely small, no larger than ±200μV [14]. The BTH-1208LS, for the finest resolution, allowed for an input of ±1V. In order to obtain the lowest possible quantization noise for the analog to digital conversion step, a gain of 5,000 V/V would need to be realized by the amplifier, or the amplifier would increase the magnitude of the input signal by 5,000.

Another problem to be addressed was the number of channels for the device. Commercial PSG devices typically have from 8 to 16 channels, but only as few as 4 channels are needed for making sleep-state measurements [14]. Voltage is a difference in electrical potential energy and must be measured relative to another point. In standard PSG practice, all channels are measured with respect to a common reference, usually the earlobe of the patient [4, 14]. This is because humans sometimes have a DC voltage that is not equal to 0, with respect to the ground, and it is necessary to obviate this DC content for the amplifier to function as intended. By connecting a probe to the earlobe of the patient and having 4 probes as differential inputs to the BTH-1208LS, with respect to the earlobe reference, four differential channels of the BTH-1208LS could be utilized. The BTII-1208LS possessed a maximum sampling rate of 1024S/s in Bluetooth mode, divided by the number of active channels [10]. In accordance with the Nyquist Criterion to

prevent aliasing, that yielded a maximum signal frequency of 128Hz, considering 4 active differential channels. Because the maximum frequency in the passband for the filter was 30Hz, aliasing was not a concern for signals in the passband.

In consideration for the design of this amplification stage and filter, an anti-aliasing hardware filter must be considered. There is no way to undo aliasing, and 60Hz noise must be taken into consideration. Industry standard EEG software performs mathematical transforms on available data for customizable filtering needs [16]. Such software, such as OpenEEG, provides a high order 60Hz notch filter option to remove 60Hz noise from the data. A problem for an opamp amplification stage is that if the 60Hz noise has a sufficiently high power content, the output of the opamp will become saturated and unable to display any useful data [4]. Considering measuring a small voltage of ±200μV, 60Hz noise is likely to saturate the amplifier output at least some of the time. Considering a device meant to be utilized in the home of a patient, 60Hz noise tolerance is crucial [18].

In order to eliminate 60Hz noise from consideration, the filtering stage of the amplifier and filter is utilized. A Low Pass Filter (LPF) is a circuit that attenuates, or reduces the magnitude of, input signals that have a frequency above the center frequency for the filter. For a LPF to be used in a PSG, a passband of 0.6Hz – 30Hz is considered. Only Gamma waves have frequencies above 30Hz, up to 80Hz, and indicate extreme focus on the part of the patient. Such waves need not be measured to indicate that a microarousal has occurred [12], and so the center frequency of the Low Pass Filter may be lowered to 45Hz. 45Hz is selected as the center frequency instead of 30Hz because of the roll off of the filter. While ideal LPFs instantly attenuate any signals above the center frequency, real filters gradually attenuate signals at a rate of 20dB/Decade per order of the filter. For a maximally flat passband filter, or Butterworth filter,

a tradeoff exists between attenuating 60Hz as much as possible, and not attenuating 30Hz [7, 17]. Therefore, a center frequency of 45Hz was selected such that 60Hz noise would be attenuated as much as possible without adversely impacting the passband.

Electromagnetic noise is a significant obstacle in the way of cleanly amplifying such a small input voltage. 60Hz noise, generated as a result of the powered wiring in American homes, can easily overwhelm a signal as small as 200μV [3]. Fortunately, electromagnetic shielding is easy to implement for Printed Circuit Boards (PCBs). A shielded PCB would be highly resistant to a wide range of noise frequencies [5]. In order for this filtering and amplification stage to be mass produced, it would also need to be implemented as a PCB. As such, designing a PCB implementation of this amplification and filtering stage was necessary.

The measure of the success of an amplification and filtering circuit is the Signal to Noise Ratio (SNR) for that circuit. Some literature suggested that 20dB, or a ratio of 10, as a guideline for PSG signals [21]. The SNR of the circuit would be measured for different input frequencies and would be measured in the following manner. MATLAB, an engineering and mathematics software system, would be used to compare the noise profile of the measured signal to the input signal. MATLAB's built in signal to noise ratio function would be used to find the SNRs. The input signal would be assumed to be ideal, and would be mathematically generated within MATLAB. The noise profile would then be acquired by subtracting the assumedly ideal signal from the measured signal. This assumption would not account for abnormalities in the supply waveform, which were certainly present. SNR determination for the circuit could be improved by sampling the input waveform and using it instead of an ideal approximation, but the ideal approximation erred on the side of requiring a more noise-resistant circuit. SNR was crucial for determining if the signal was amplified cleanly enough for the circuit to have been useful.

RESULTS & DISCUSSION

Investigation into circuit implementation began with considering what operational amplifier, or opamp, would be selected for use in the circuit. Opamps are an analog device that can be implemented in a circuit for signal amplification, filtration, or both. Although a number of opamps would have performed acceptably, the op37 was selected for having excellent noise characteristics of 3nV/√Hz [2], a dual-ended supply, for being inexpensive, and because it was on-hand for ease of testing. After the op37 had been selected as the opamp with which the filter and amplifier would be constructed, the center frequency of the filter was to be investigated.

Two jobs had to be performed by this circuit: amplification of a 200uV signal to 1V, and attenuating any content above a certain frequency. Separate configurations exist for creating amplifier or filter opamp circuits, but a number of arrangements, or topologies, exist for single-opamp filters and amplifiers. The Butterworth filter is one such topology, allowing for a single active element to act as a linear amplifier and a filter. The advantage of the Butterworth filter over alternative topologies is that the passband for a Butterworth filter is maximally flat, flatter than the passband of Chebyshev or Elliptical filters [13]. Butterworth filters are also very easy to implement and comparatively insensitive to changes in resistor and capacitor values. In order to increase the roll off rate of the overall filter, the approach of cascading multiple 1st order Butterworth filters together in series was selected. This approach was selected due to the ease of implementing the total required gain of 5,000 in several increasing stages, as well as the simplicity of configuring 1st order Butterworth filters [9]. Implementing a large gain such as 5,000 in several increasing stages is also a good practice for amplifier design for noise mitigation [22].

The cutoff frequency for a Low Pass Filter (LPF) is the frequency at which the output amplitude has been attenuated to approximately 70%, or -3dB, of the input amplitude. For this reason, the center frequency is also called the 3dB frequency. For a LPF, inputs of a frequency below the center frequency are slightly attenuated, whereas inputs of higher frequencies than the center frequency are greatly attenuated. Below is a graph showing the relationship between input frequency and output amplitude for a 1st order Butterworth LPF.

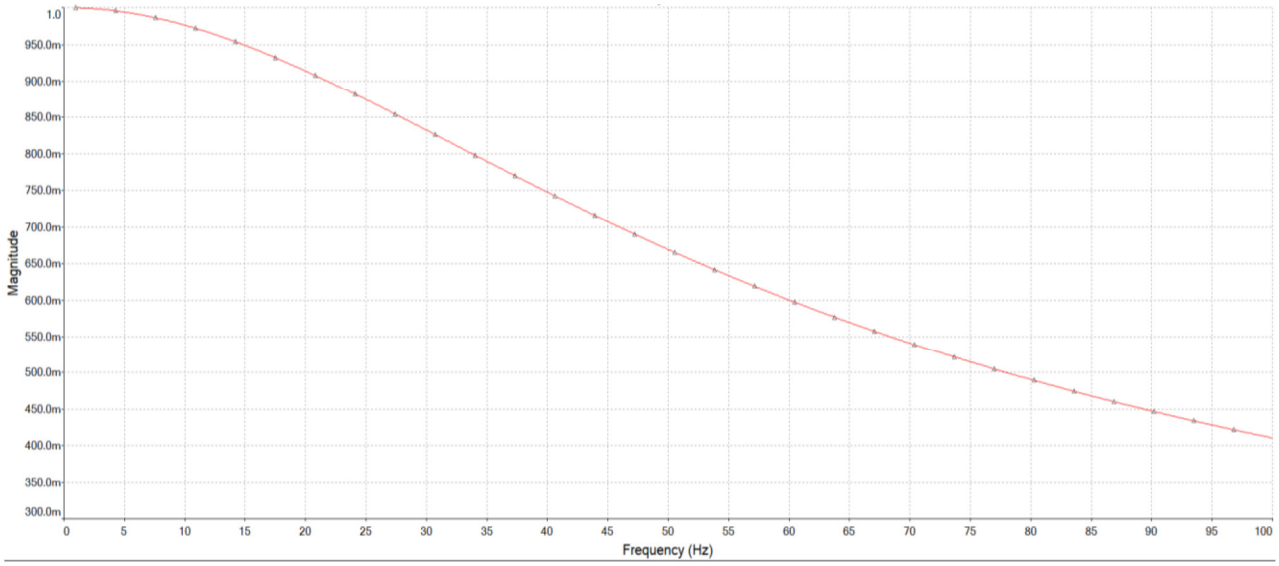

Figure 2: Bode plot for 1st order Butterworth LPF

When the cutoff frequency for each Butterworth LPF was considered, it was important to keep the purpose and operating conditions of the circuit in mind. In order to determine sleep state, the highest frequency brainwaves that needed to be analyzed were Beta waves, ranging from 14 to 30Hz. Additionally, the device was expected to be able to significantly attenuate 60Hz noise, noise from the environment in which it was expected to operate. Therefore, the

cutoff or center frequency for each Butterworth LPF sub-element was implemented as 45Hz. For 1st order Butterworth filters, the implementation of this cutoff frequency was straightforward. Following Butterworth topology below for a first order filter, values of R2 and C were selected to tune the filter.

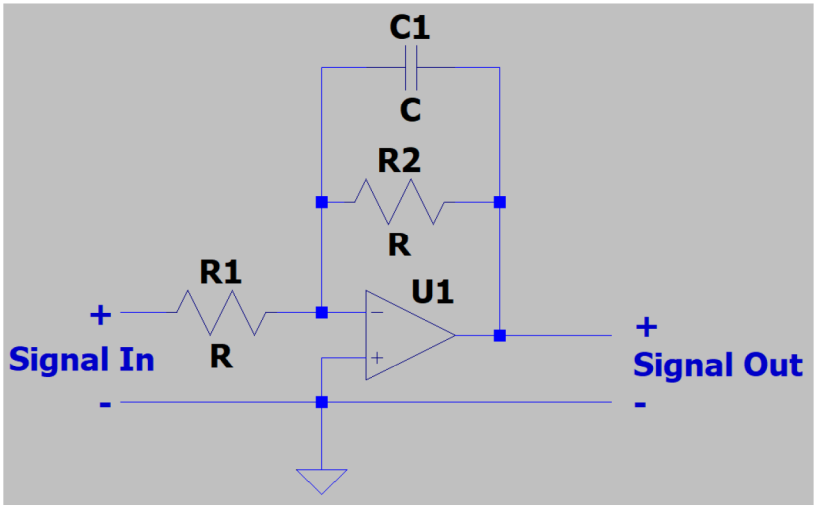

Figure 3: First order Butterworth LPF topology

$$f_c = \frac{1}{2\pi R_2 C}$$

Equation 1: First order Butterworth center frequency as a function of R2 and C

Cascading these 45Hz Butterworth LPFs in series in order to implement staged gain resulted in an overall lower center frequency. For example, cascading three 45Hz 1st order Butterworth LPFs in series was found to produce an overall cutoff frequency of about 25Hz. Varying the

center frequency of each Butterworth LPF component of the overall filter served to non-linearly change the center frequency of the cascaded system.

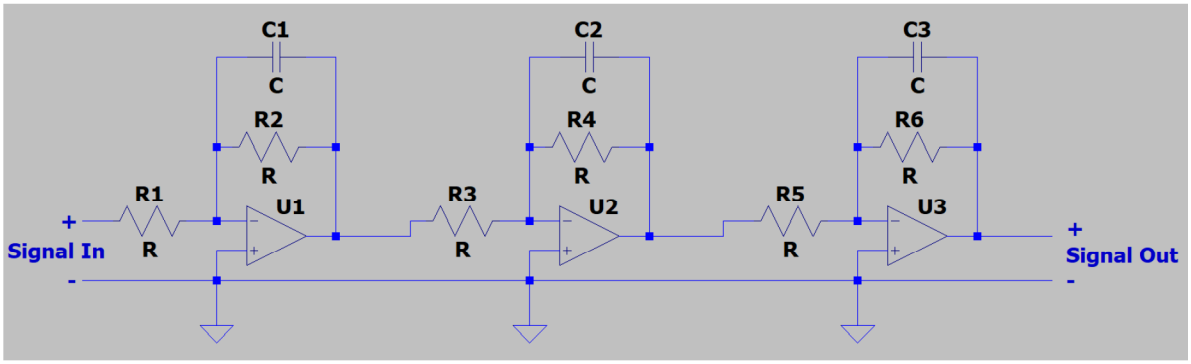

Figure 4: Topology of cascaded 1st order Butterworth filters

For 1st order Butterworth LPFs of center frequency 45Hz, this topology was found to only slightly attenuate frequencies in the passband. In exchange, signals above 30Hz were more heavily attenuated. 60Hz content, for example, was found to be attenuated by 14dB, or to 20% of the original amplitude, while 30Hz content was only attenuated by 5dB, or to 55% of the original amplitude. Slightly attenuating some frequencies in the pass band in order to more greatly attenuate frequencies above the cutoff frequency was an acceptable tradeoff. The attenuation of higher frequency content in the passband could be undone via software scaling, with an inconsequential increase in quantization noise for those frequencies as a result [1].

For a cascaded series of 1st order Butterworth filters, placing multiple filters in series sees an increase in total gain proportional to the gain of the amplifier added. For example, cascading

three amplifiers together in series, each with a gain of 5V/V, will result in a system gain of 5 x 5 x 5 = 125. The gain of a 1st order Butterworth filter is expressed below in equation 2 [17].

$$G(\omega) = \frac{1}{\sqrt{1 + \frac{\omega^2}{\omega_c}}}, for\ \omega = angular\ frequency, \omega_c = center\ frequency$$

Equation 2: Gain of first order Butterworth LPF

As the gain of a cascaded system of amplifiers is equal to the product of each individual amplifier, the gain of a system of three cascaded 1st order Butterworth LPFs is given by the equation below [19, 20].

$$G(\omega) = \left(\frac{1}{\sqrt{1 + \frac{\omega^2}{\omega_c}}} \right)^3$$

Equation 3: Gain of a system of three cascaded 1st order Butterworth LPFs

This equation matches results obtained from AC analysis of a simulated system of three cascaded 1st order Butterworth LPFs. A graph of equation 3 may be seen below in figure 5, and a simulation Bode plot, or frequency response plot may be seen below in figure 6.

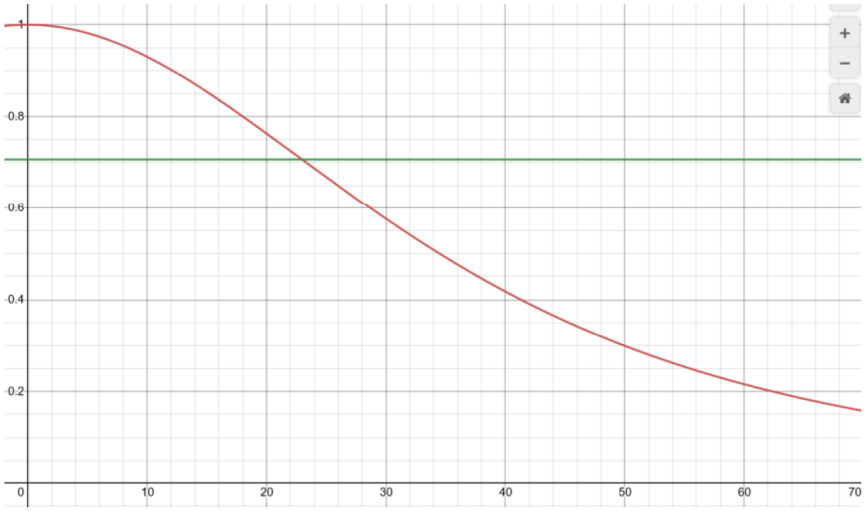

Figure 5: Graph of equation 3

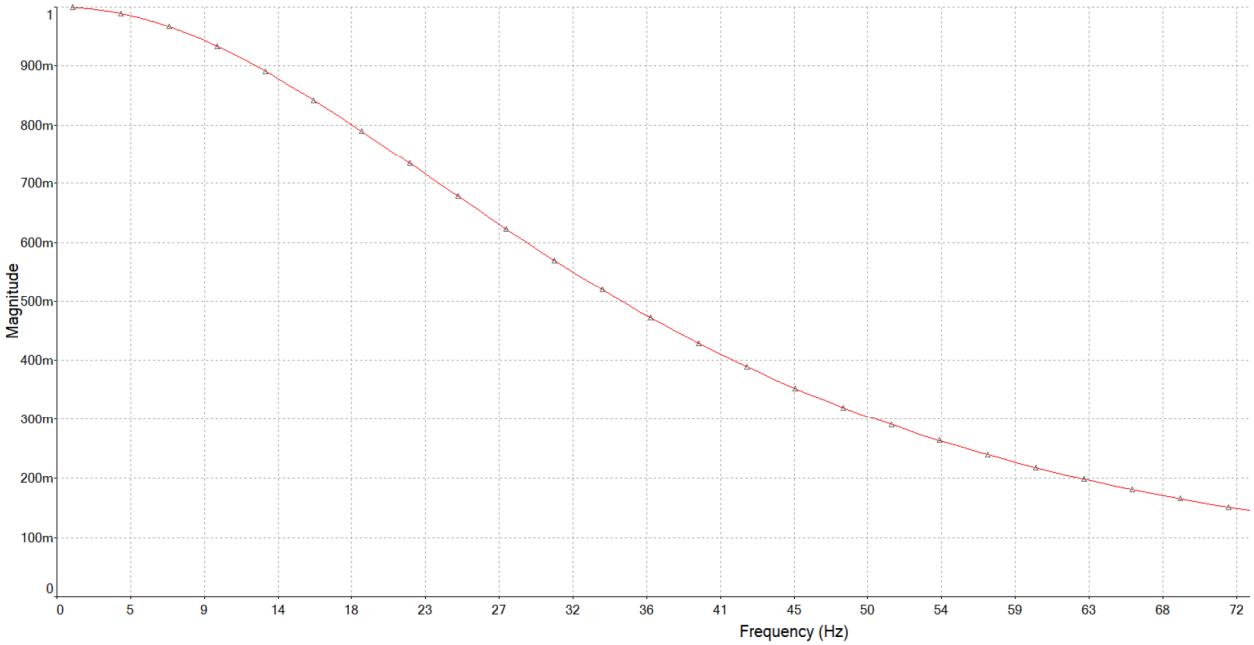

Figure 6: Bode plot of 3 cascaded 1st order Butterworth LPFs

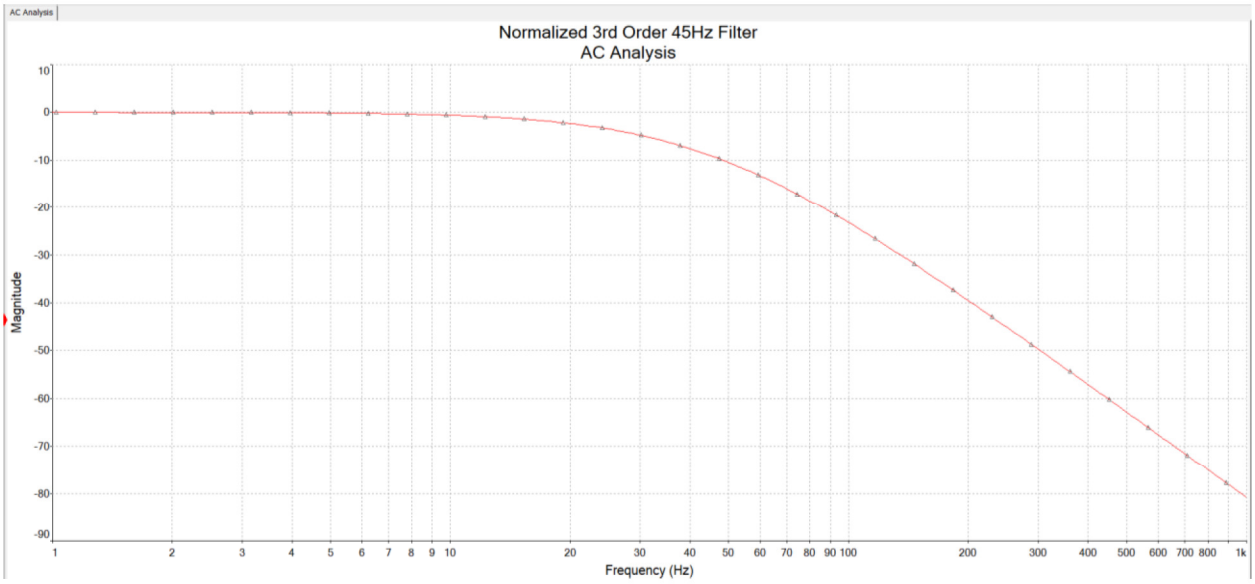

Figure 6.2: Bode plot of 3 cascaded 1st order Butterworth LPFs, logarithmic graph

Four circuits were simulated to investigate the attenuation of 30Hz and 60Hz content. Each of these circuits was a series of 1st order Butterworth LPFs, and each circuit differed based on the number of stages in series. Circuits with one through four stages were simulated. Table 1, below, demonstrates the trend that an increase in number of series filters had on the attenuation of 30Hz and 60Hz content.

Number of Series Butterworth Filters	30Hz Amplitude (V)	60Hz Amplitude (V)	60Hz/30Hz Amplitude Ratio
1	0.831	0.596	0.717
2	0.689	0.357	0.518
3	0.575	0.214	0.372

4	0.477	0.125	0.262

Table 1: Filter response per filter order, for Butterworth fc = 45Hz

For an increase in number of cascaded filters, the attenuation of 60Hz content increased. The attenuation of 60Hz content increased faster than the attenuation of 30Hz content, as can be seen in the increasing amplitude ratio column. Unfortunately, while attenuation of 60Hz content is desirable, attenuation of 30Hz content below a certain point is unacceptable. Attenuation of a 30Hz signal could be undone via scaling on software after the filter, but attenuation of the signal beyond a certain point would result in an unacceptable loss of resolution. In order to avoid reducing the magnitude of the 30Hz content below 50% of the original value, a series of three filters was selected. This value represented a compromise between attenuating 60Hz content and leaving 30Hz content unattenuated.

An additional consideration to take into account was that there would be four channels in this design. Four channels were found to be the lowest number of channels a PSG could have and still reliably differentiate sleep states. For the chosen topology, three opamps would be required per channel. Adding one opamp to a single channel would result in four total opamps being added to the design. These opamps would result in an increased product price, additional power consumption, and an increase in board size and cost. Therein was additional incentive to produce a working design with fewer active components.

Now that a design had been finalized, a circuit could be derived, constructed, and tested. As a three stage design was decided upon, the 5,000 gain was split into gain stages of 10, 20, and 25,

sequentially. Figure 7 below shows the circuit diagram for one channel of the amplifier and filter, with all the other filters being identical.

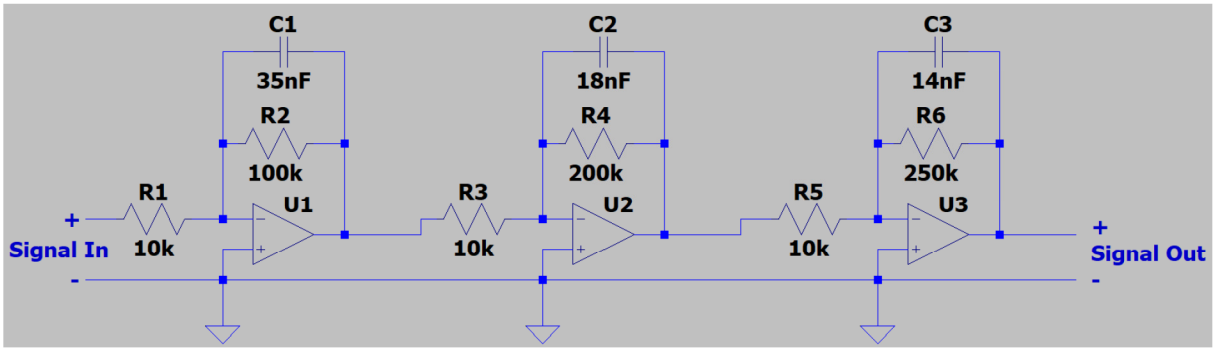

Figure 7: Amplifier and filter single channel circuit diagram

The circuit was designed by selecting the resistor and capacitor values for the Butterworth topology. An R1 value of 10kΩ was selected in order to provide sufficient input impedance for the opamps. Next, R2 values were selected for each stage to produce the required gain at that stage. This meant that from signal input to signal output, the R2 values were 100kΩ, 200kΩ, and 250kΩ. Finally, the capacitor values were calculated. Equation 4, below, was used to determine the correct capacitor value for each stage.

$$C = \frac{1}{2\pi R_2 f_c}, for\ R_2\ in\ Butterworth\ topology, f_c = center\ frequency$$

Equation 4: Capacitor value for 1st order Butterworth filter

Considering that the center frequency for each stage was to be 45Hz, and that the R2 values for the stages had been determined, the capacitor values could be determined. After the circuit was established for each stage of the amplifier and filter, the stages were connected from output to inverting input. It is also worth noting that each of the Butterworth filter implementations uses the inverting input for each opamp. The consequence of utilizing the inverting input for an opamp is that the phase of the output signal is shifted by 180° compared to the input signal. This phase shift is not important for differentiating sleep state [14], however, as only the power content within the frequency ranges need be compared. This circuit, then, represented one channel of a four-channel device, the need for four channels stemming from the minimum number of probes to successfully measure brain waves.

After the circuit for one channel had been realized, the circuit was implemented on a breadboard for testing. The op37e was used as the operational amplifier, and the circuit was assembled with a total gain of 200 instead of 5,000. This was done in order to test a decreasing progression of input voltage amplitudes, as noise was expected to be a significant impediment to this testing. Electromagnetic interference was expected to prevent the breadboard implementation of the circuit from being able to cleanly amplify and filter a 200uV input signal. Therefore, larger amplitude signals were tested, with their amplitudes decreased following successful results. An image of the breadboard circuit may be seen below in figure 8.

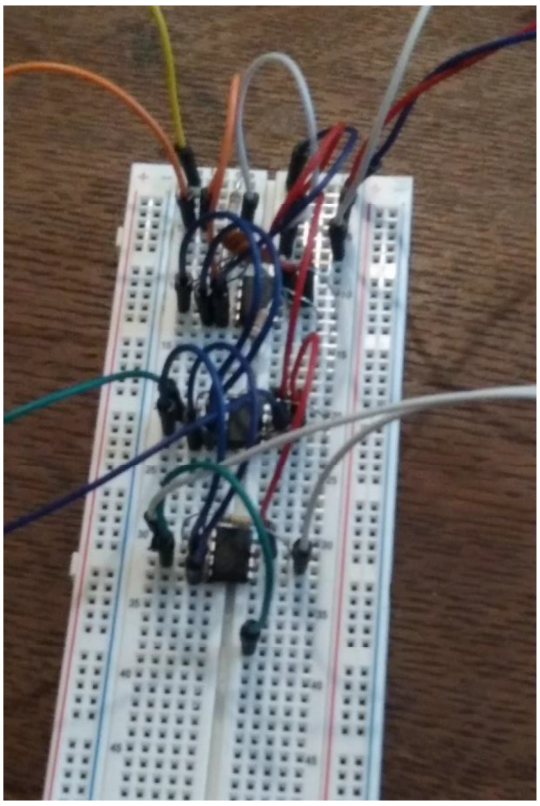

Figure 8: Picture of breadboard implementation of the circuit

Testing was accomplished by supplying the circuit with a voltage waveform from a Digilent Analog Discovery 2 module and reading the output with a Tektronix mixed signal oscilloscope. Input waveforms from the Analog Discovery 2 module were passed through a resistor divider in order to achieve low magnitudes. Input waveforms were visually inspected to ensure that they were not distorted. Lower magnitude waveforms, like 600uV and 200uV, possessed moderate amounts of noise, but were not significantly distorted.

As opposed to a Printed Circuit Board (PCB), a circuit realized on a breadboard and using long wires is much more highly susceptible to noise. This noise can be introduced into the circuit on intermediary connecting wires and allowed into the circuit from the unfiltered power supply. Figure 9 below shows one of the waveforms captured while testing this circuit.

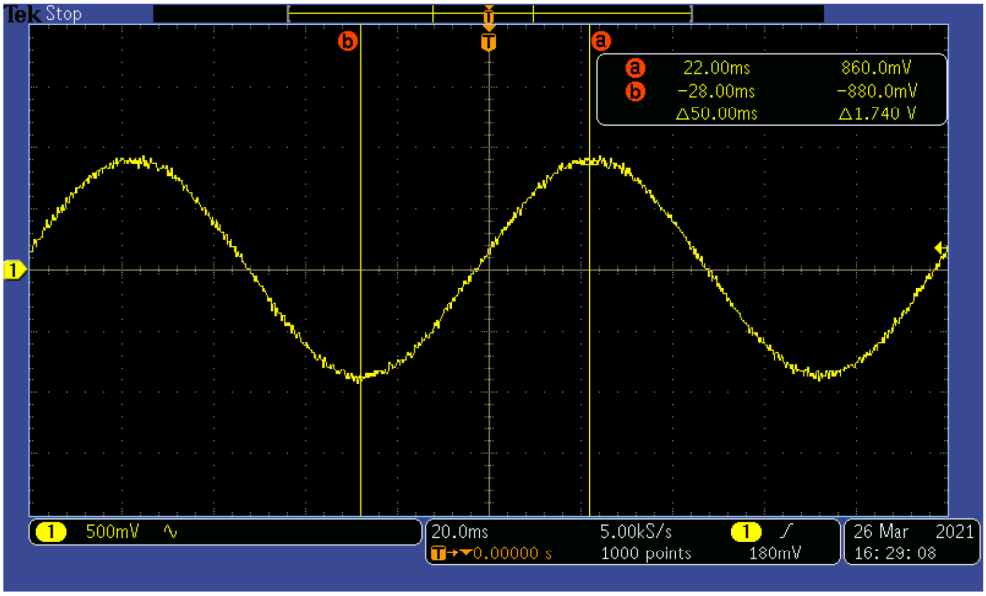

Figure 9: 200 gain filter output for 20Hz 5mV sinusoidal input

Testing was carried out on this 200 gain 25Hz lowpass filter at a number of amplitudes and frequencies. Output voltage waveforms were recorded for analysis at amplitudes of 5mV, 1mV, 600µV, and for frequencies of between 1Hz and 70Hz. The frequency response may be seen below in figures 10, 11, and 12 for this 200 gain 25Hz lowpass filter and amplifier.

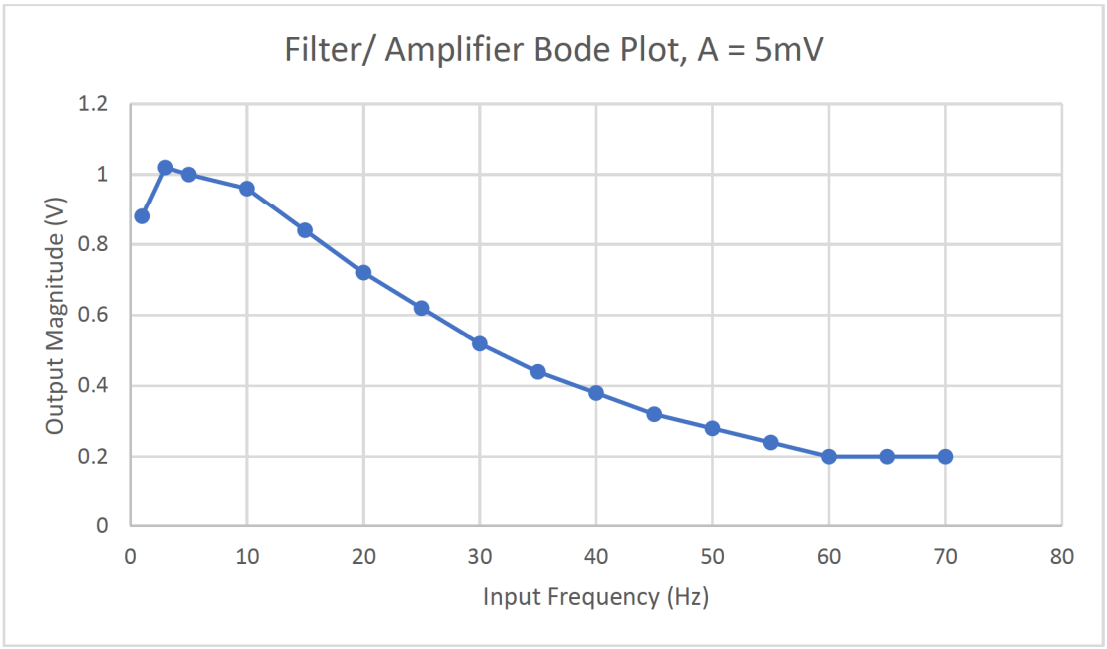

Figure 10: Frequency response for 200 gain 25Hz lowpass filter and amplifier, input of 5mV

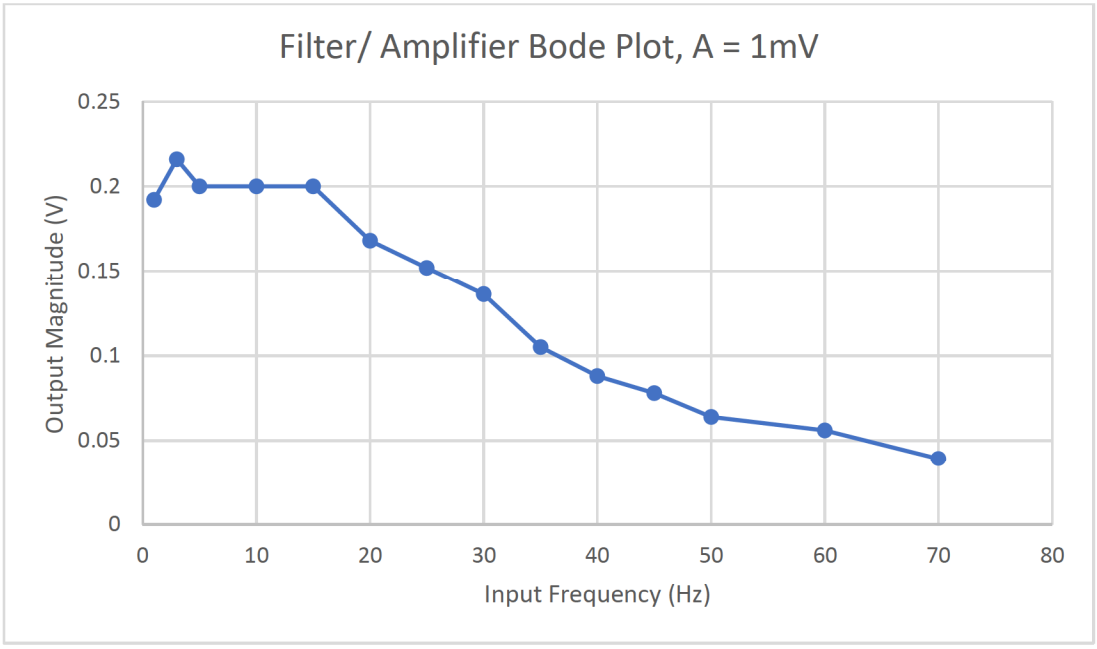

Figure 11: Frequency response for 200 gain 25Hz lowpass filter and amplifier, input of 1mV

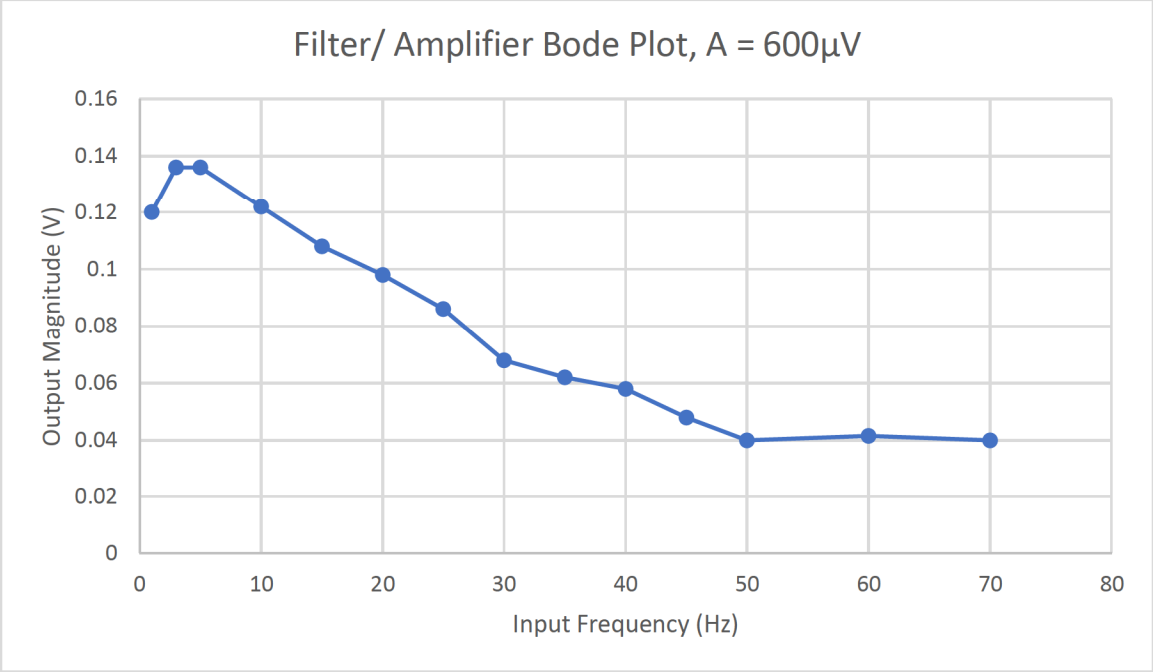

Figure 12: Frequency response for 200 gain 25Hz lowpass filter & amplifier, input of 600µV

The response of the 200 gain circuit indicates that the circuit is behaving in accordance with the theoretical model as well as the simulation. In order to represent the data another way, the output waveform for the 5mV 10Hz waveform was converted into the frequency domain using a Fast Fourier Transform (FFT) in Matlab, and may be seen below in figure 13.

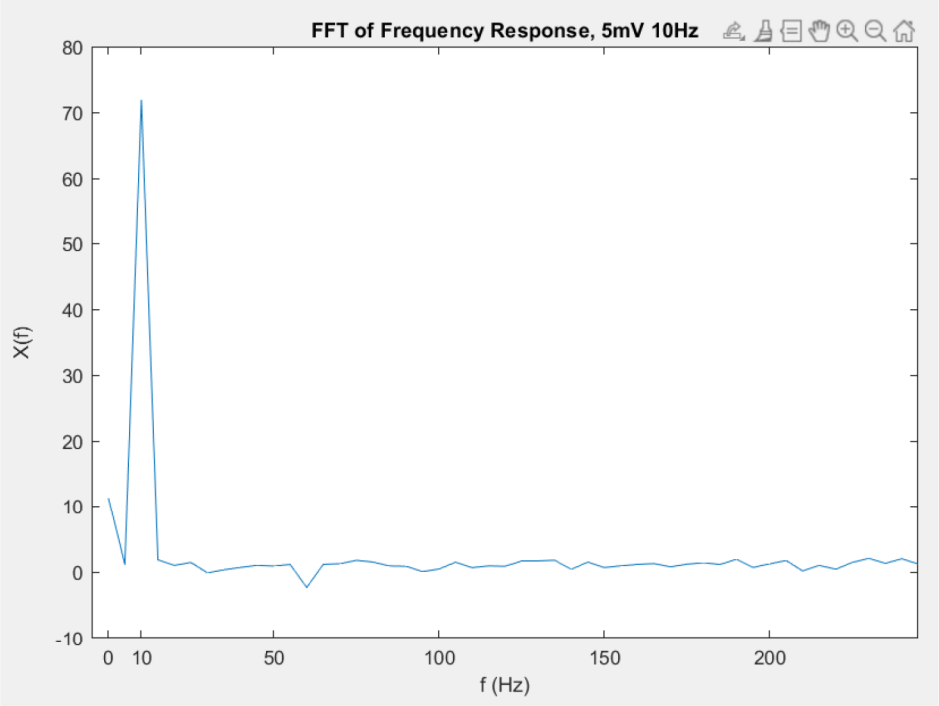

Figure 13: FFT of frequency response for 5mV 10Hz input

For figure 13, the frequency content at values other than 10Hz may clearly be seen. This results in noise on the output voltage waveform that can impact the ability of the channel to transmit data correctly. Therefore, the signal to noise ratio of the circuit was investigated for different frequencies and amplitudes. The below table, table 2, provides the Signal to Noise Ratios (SNRs) calculated for each type of input into the circuit. The below SNRs were calculated using the built in SNR function in MATLAB. Oscilloscope waveform captures were converted to MATLAB arrays using Tektronix scope software and a python script.

	5mV	1mV	600µV

10Hz	23.35dB	20.12dB	20.37dB
30Hz	16.93dB	12.27dB	15.27dB

Table 2: SNR for different input frequencies and magnitudes

Considering that the circuit was implemented on a breadboard with unfiltered power supplies and plenty of stray capacitance and inductance, the SNRs are fairly good. For even the 600μV signal at 10Hz, a magnitude only 3 times larger than the target amplitude of 200μV, the SNR was 20.37dB. Although this SNR surpasses the guideline of 20dB offered in the literature [21], the measurements at 30Hz are lower than 20dB. That the measurements taken at the high end of the passband dip below 20dB means that this circuit would have violated the guideline if the amplitude had been 200μV. Finally, the amplitude of 200μV had to be tested on a physical circuit.

In order to get the highest resolution possible on the output signal, the gain of the circuit was reworked. Resistors and capacitors for the cascaded 1st order Butterworth LPF system were replaced to achieve a total system gain of 5,000. The center frequency of each filter was kept the same, at 45Hz, then causing a center frequency for the whole system of approximately 25Hz. When the circuit was tested with an input of 200μV, the output was completely overwhelmed by noise. No degree of a useful signal could be extracted from the measured waveforms. Although the physical limitations of the breadboard circuit had not been enough to overwhelm the 600uV input, the 200uV input was completely overwhelmed by noise. Fortunately, this noise is mitigated for a circuit implemented in a Printed Circuit Board (PCB).

For a circuit implemented in a PCB, numerous measures exist for limiting the impact that electromagnetic noise has on the circuit. All of the traces, or wires, for the circuit are designed to be as short as possible, the power traces are low pass filtered to remove high frequency noise, and the ground for the circuit is implemented as a planar layer in the board. All of these techniques were utilized in designing an implementation of this circuit on a PCB. Electromagnetic shielding is much easier to implement for a PCB, which combined with the other noise mitigation practices listed above serve as the basis for a more noise-resistant circuit. Below are two figures, figure 14 and figure 15, that illustrate the circuit diagram of the PCB as well as the routing and component placement for the board.

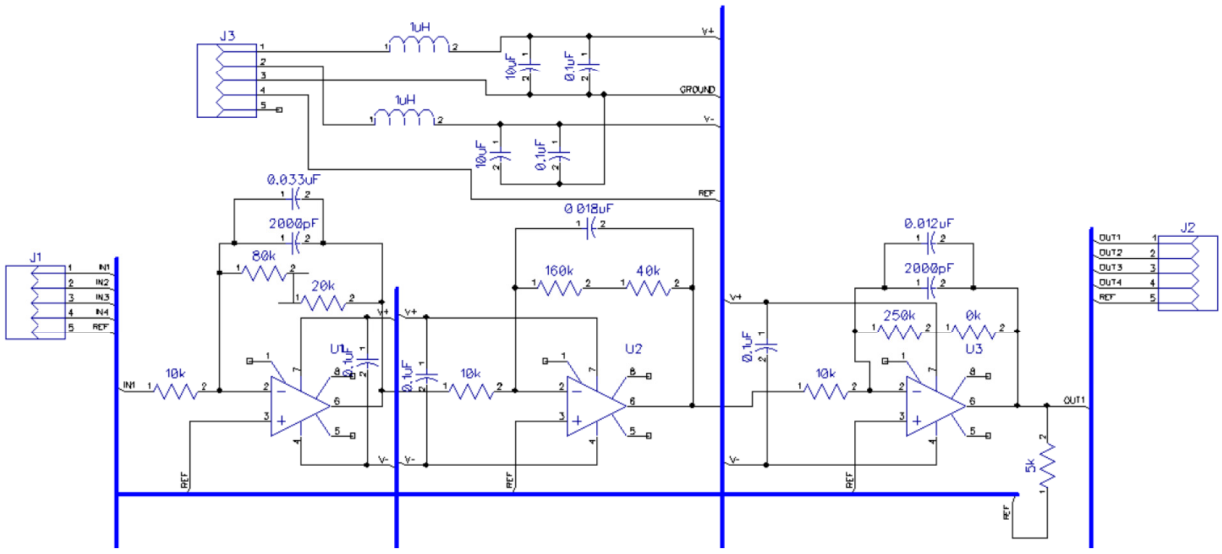

Figure 14: Circuit Diagram for power filtering and one channel of four channel PCB circuit

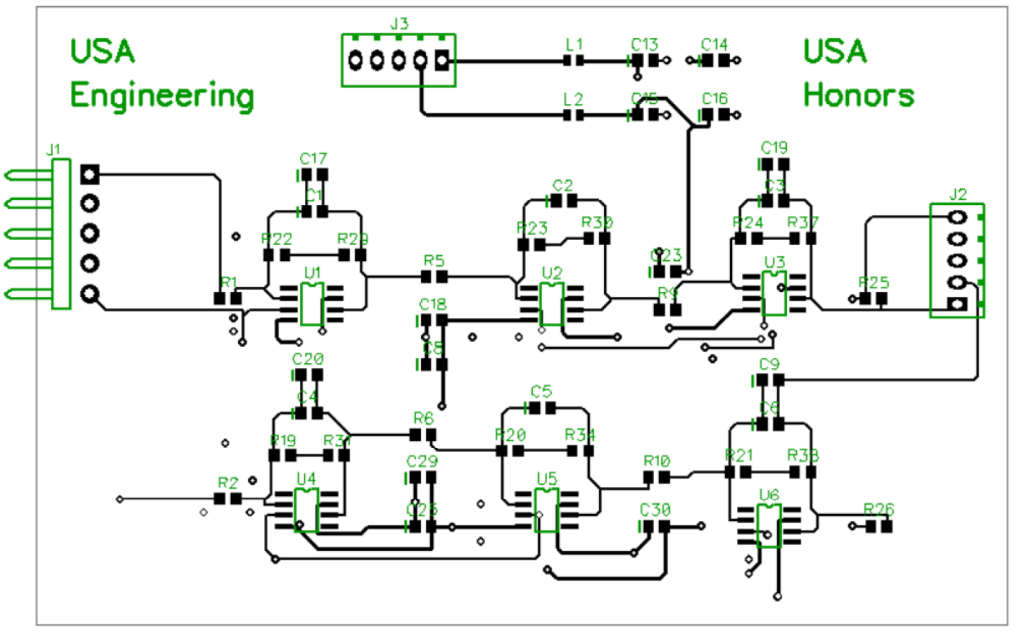

Figure 15: Routing and layout for top layer of PCB

The board also featured a connector, J1 on the diagram, designed to adapt to a common type of EEG probe. The board design was submitted to the PCB manufacturer PCBway and quoted at $138.60 per board. This price would only be lowered in the event of mass-manufacturing and bulk-order discounts.

The results obtained through testing the breadboard circuit are highly encouraging with respect to the success of the same circuit implemented on a PCB. SNRs measured from inputs as low as 600µV were able to surpass the 20dB guideline for low frequency content in the passband. It is likely that, through implementing the cascaded 1st order Butterworth LPF as a PCB with noise-mitigation practices, the circuit would cleanly amplify a 200µV 0.6Hz – 30Hz signal while attenuating out-of-band noise. This board would then be able to be used to amplify measurements made from the scalp of a patient in order to record those voltages and determine

sleep state from them. This method of determining sleep state could then be used in conjunction with blood oxygen sensors and physical respiration sensors in order to be part of a PSG and offer more accessible screening for OSA.

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