Portable Multi-Channel Electrotactile Haptic Feedback System

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1. Introduction

1.1 Objective

No commercially available upper limb prosthetic device has any meaningful form of touch feedback. Lack of touch feedback is one of the most fundamental problems with modern myoelectric prosthetic devices [1], [2]. Because of this problem, users can't grasp delicate objects without crushing them, or feel anything they touch with their prosthetic hand. Lack of sensory feedback results in reduced device embodiment, increased difficulty of use for manipulation, and overall leads to increased device abandonment. [3]. The solution to this problem is conceptually simple: find a way to restore sensory feedback to an amputee through their prosthetic device. However, from a technical standpoint, this goal can be a significant challenge. The two forms of sensory feedback that are most important for use and device embodiment are proprioception and tactile feedback; our objective is to provide tactile feedback. We will achieve this using a system of tactile pressure sensors which can be embedded in a prosthetic device and a multi-channel electrical stimulator which will non-invasively provide stimulation across surface electrodes. The pressure sensors will modulate the sensation intensity as a function of the pressure measured on the fingertips.

1.2 Background

Tactile feedback for prosthetics can be provided through the use of vibrotactile motors (small vibrating motors placed on the skin), non-invasive electrotactile, and invasive electrotactile (direct nerve stimulation). The two most viable techniques for a conventional myoelectric prosthetic are vibrotactile feedback, and non-invasive electrotactile [4]. Non-invasive electrotactile has a number of advantages over vibrotactile, including spatial resolution, device longevity, and flexibility in sensation quality and intensity. However, due to size, cost and power limitations nearly all conventional electrotactile stimulators are used exclusively in research; there are currently no commercial prostheses which use electrotactile stimulation as a form of sensory feedback. Commercial constant current stimulators do exist, but are all some combination of expensive, power hungry, and large. The commercial device we know of which comes close to our target specifications is the BIOPAC STMISOLA, which costs approximately $1500, consumes almost 3W quiescent power, and is too large to be mobile. To resolve these issues, our device will be significantly smaller and cheaper than existing commercially available devices.

Pressure sensing in robotic applications is currently an area of active research. There are a number of different methods used, including the use of fluids, capacitive sensors, and piezoelectric resistors. Most methods are either too expensive to be viable or too low-performance to be useful. A pressure sensing method which hits a good middle ground between performance and cost employs the use of barometric pressure sensors sealed in silicon, a technique employed by the companies TakkTile and PSYONIC. We intend to improve on this design by mounting an array of barometric sensors on a flexible PCB, allowing the device to easily mount and conform to a compliant robotic finger.

1.3 High Level Requirements

1.3.1 The stimulator must be small enough to unobtrusively worn or fitted into the socket of a prosthetic limb.
1.3.2 The device must be low cost: <200$ for the entire device
1.3.3 The device needs to be able to support up to 3 channels (minimum)

2. Design

The stimulator will consist of four different primary modules: the digital to analog converter (DAC), the high voltage power converters, the pressure sensors, and the analog constant current controllers. The DAC module will consist of a microcontroller and an 8 channel SPI DAC chip. The power supplies should be able to convert a battery voltage to between +/-200-300Vdc, to provide supply to the constant current controller. The pressure sensors will be arranged in clusters of up to 3, and will interface with I2C muxes. The constant current controller will output a current which will vary as a linear function of the voltage output from the DAC.The stimulator needs to provide high compliance voltage constant current stimulation signals across the user's skin. Constant current is superior to constant voltage for this application because it helps maintain a consistent sensation when electrodes shift, and protects the device if the electrodes short [5]. The output should be arbitrary waveform, to potentially allow sensation quality manipulation in future research; however, we intend to deliver 'biphasic' square pulses, which are a standard for electrotactile stimulation. The pulses will typically have a frequency of 50hz with a 1% duty cycle. The amplitude will vary as a function of the measured pressure on the corresponding channel. The overall design will be modular; the DAC will have 8 ports which will connect to current controller/high voltage power supply modules, and a port for I2C which will allow multiple addressable pressure sensor boards.

2.1 Block Diagram

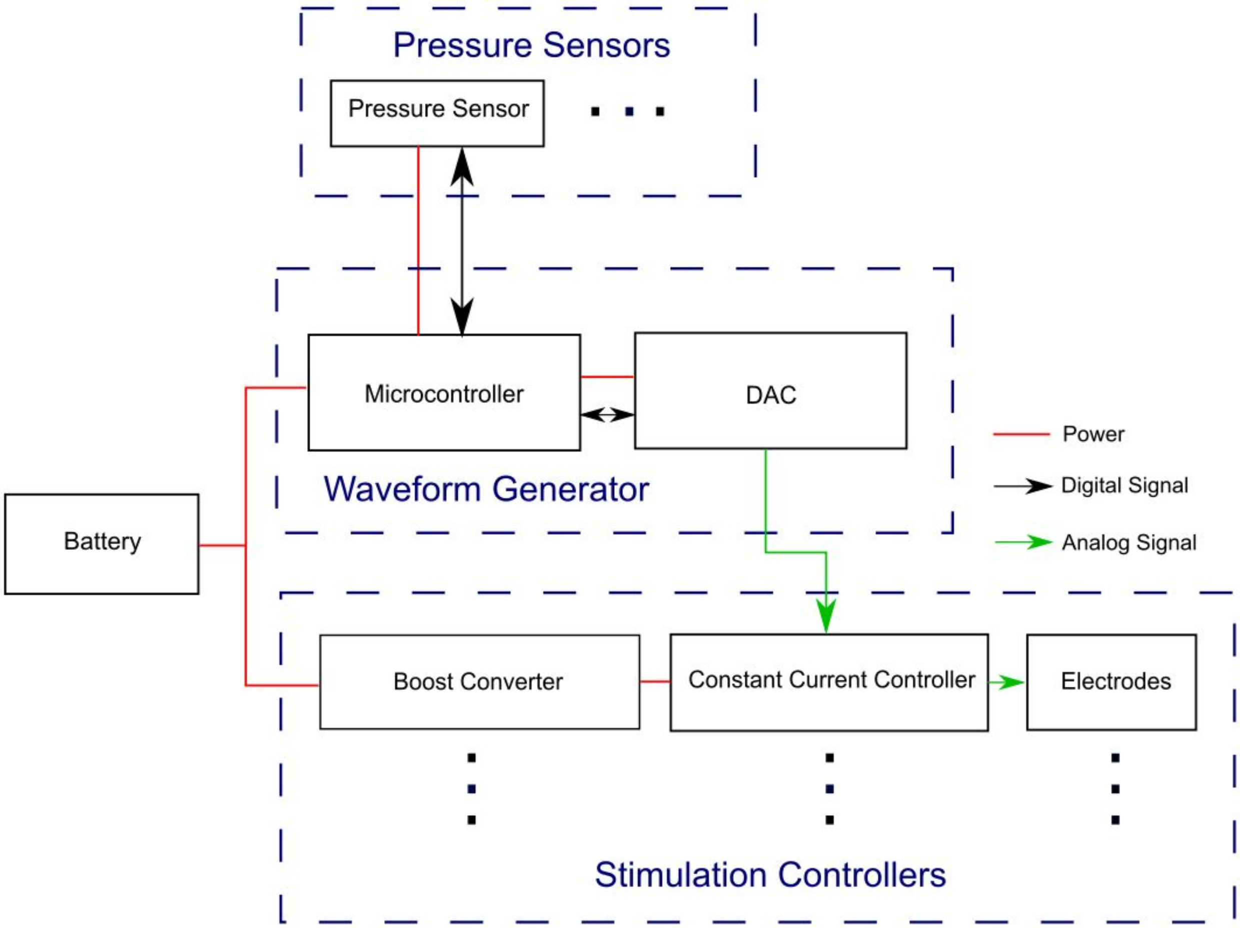

Figure 1. Block Diagram

2.2 Physical Design

As described above, our design will be modular; i.e., pressure sensors and high voltage stimulation controllers can be disconnected and connected as needed. We want our design to be small enough that it can be configured to fit into a socket of a lower limb amputee. An example of such a socket is depicted in the image below:

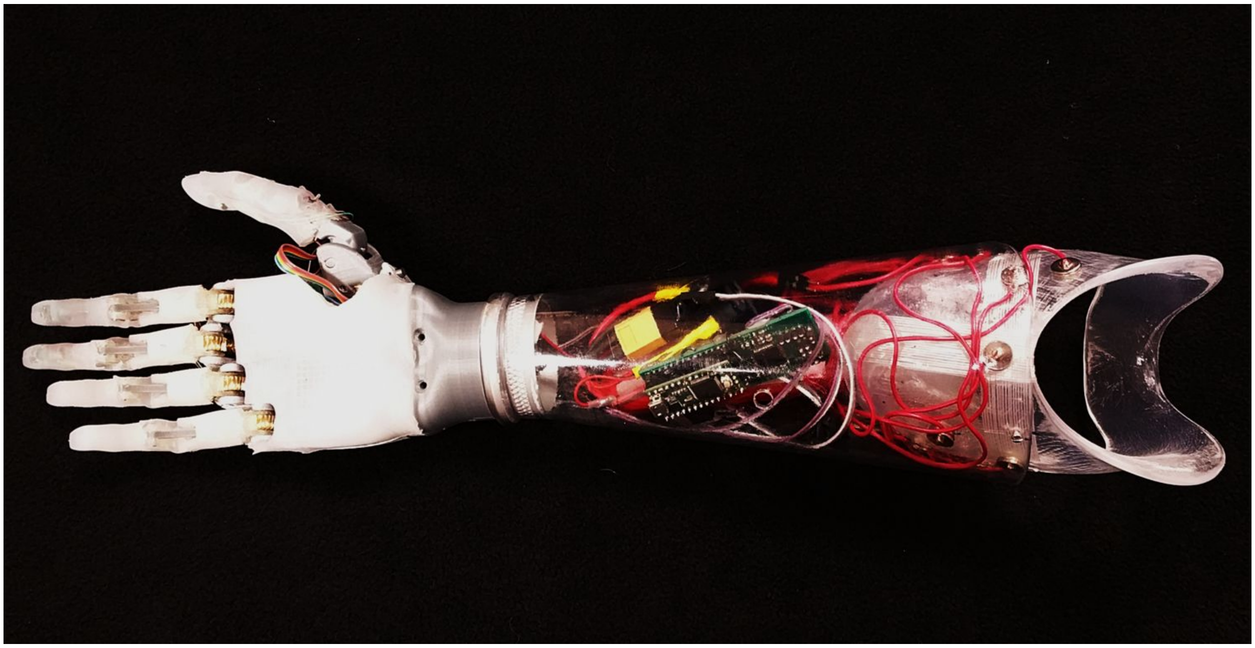

Figure 2: Example socket (right) and PSYONIC's myoelectric prosthetic hand (left)
(Photo credit: PSYONIC)

Another constraint on the physical design is that the pressure sensor board must be able to conform to the compliant fingers in PSYONIC's prosthetic hand. The figure below shows how the board should be mounted in CAD.

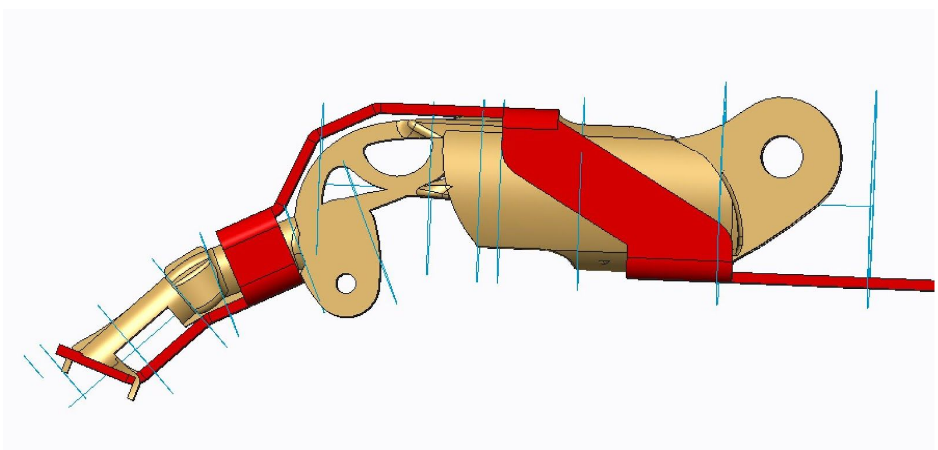

Figure 3: Compliant finger with required required board outline
(Photo/Finger Design credit: PSYONIC, Kyung Yun Choi)

2.3 Block Description

2.2.1 Constant current controller

The controller must take in a voltage (+/-10V) and use it as a reference for a controlled current. It will use a bootstrapped modified howland current source, to allow for high compliance voltage.

Requirement	Verification
1. Must be constant current. 2. Must be able to provide +/-15mA. 3. Compliance voltage must be at least +/-150V. 4. Must have fast rise times (<2us). 5. Must have noise that is not perceptible in the stimulation output (low ringing, good PSRR). Target noise is <2Vpp.	1. Connect resistors of varying impedances to the stimulator and measure the voltage drop. Verify the current waveform is consistent each of these resistive loads. 2. Provide the appropriate input voltage, measure the drop across a known resistance 3. Send +/-1mA across a 150k resistor, measure the voltage drop across this resistor and ensure it is +/-150V. 4. Send a square wave input signal, measure the stimulator rise time with an oscilloscope 5. Measure the noise with an oscilloscope

2.2.2 Boost Converter

The power supply will convert voltage from a battery to approximately +/-200V. A flyback topology will be used. The primary coil will be driven with a high power mosfet. For flexibility, the switching source will be an Attiny85 microcontroller.

Requirement	Verification
1. Must be able to provide at least +/-200V. The input voltage can range from 9-12V, depending on the type of battery desired. The output need not be regulated. 2. When connected to the constant current source, the stimulator system will not dissipate more than 1W (quiescent), to ensure practicality in a mobile setting. 3. Noise must be low enough that the current controller PSRR can compensate for it. Noise must be low enough to not have any physiological effect; an upper maximum is +/-2V.	1. Measure the output with no load (bleed load) and verify the output voltage with an oscilloscope or multimeter 2. Measure the voltage drop across a low value resistor in series with the battery when connected to the stimulator, which should send 0mA across some typical load resistance (10k). Log this voltage drop with an oscilloscope, and use the data to calculate power. 3. Check the noise with an oscilloscope.

2.2.3 Microcontroller

The microcontroller will provide an interface between the pressure sensors and DAC, by taking in pressure data via I2C, using that information to modulate a waveform in program memory, and sending that information over SPI to the DAC chip. We will reference the open source hardware schematics from teensy 3.2 to break out a MK20DX256VLH7 microcontroller to interface with the DAC.

2.2.4 DAC

The DAC chip will provide 8 channels of programmable voltage, which will be used as setpoints for the analog constant current sources. The DAC108S085 meets all the listed requirements.

Requirement	Verification
1. The chip must be SPI controlled; this is to allow the central microcontroller to be configured as either I2C master or slave, depending on whether it interfaces with the pressure sensors directly or through a different microcontroller. 2. The DAC PCB must be capable of producing 50Hz bipolar square pulses between -10V and 10V on all eight channels, with the same phase.	1. Check to make sure SPI communication works by measuring output waveforms with an oscilloscope. 2. Verify that the output waveform is correct with an oscilloscope.

2.2.5 Pressure Sensors

The pressure sensors will be connected in sets of up to 3 per board. They will connect to a low profile I2C mux. Each mux will have a hardware-settable device address to allow multiple sets of devices to connect to the same I2C master. The barometric pressure sensor and mux will be mounted on a flexible PCB, to allow the board to fit inside a compliant prosthetic finger. In order for the barometric pressure sensors to be adapted to measure tactile pressure, the entire board will be vacuum sealed in rubber or silicon. We will use the TCA9548 for our I2C mux, and either the MPL115a2 or the MPL3115a2 for as our pressure sensor.

Requirement	Verification
1. The pressure sensor data has manageable drift and hysteresis, if present; the pressure sensor data should be post-processed such that the output must not cause the baseline stimulation to drift enough to feel a difference (typical JND is .1mA).	1. Log the pressure sensor data into a file and plot it in matlab. Verify the post-processed output does not result in drift which would create a perceptible change in stimulation (JND .1mA).

2.2.6 Battery

We will likely use an 11.1V lithium polymer battery in our design, although this choice could change. If we do use an 11.1V lithium polymer battery, we will purchase one with a built in overcurrent/undervoltage protection module. We will select our battery chemistry based on the power consumption of the stimulator, to achieve our target battery life.

Requirement	Verification
1. The battery or batteries must be able to power the device for at least 5 hours.	1. We will power the device and dissipate current 1ma across a 50k resistor for 5 hours, and verify that the stimulation waveform has not decreased in amplitude.

2.4 Schematics and Functionality

2.3.1 Constant Current Controller

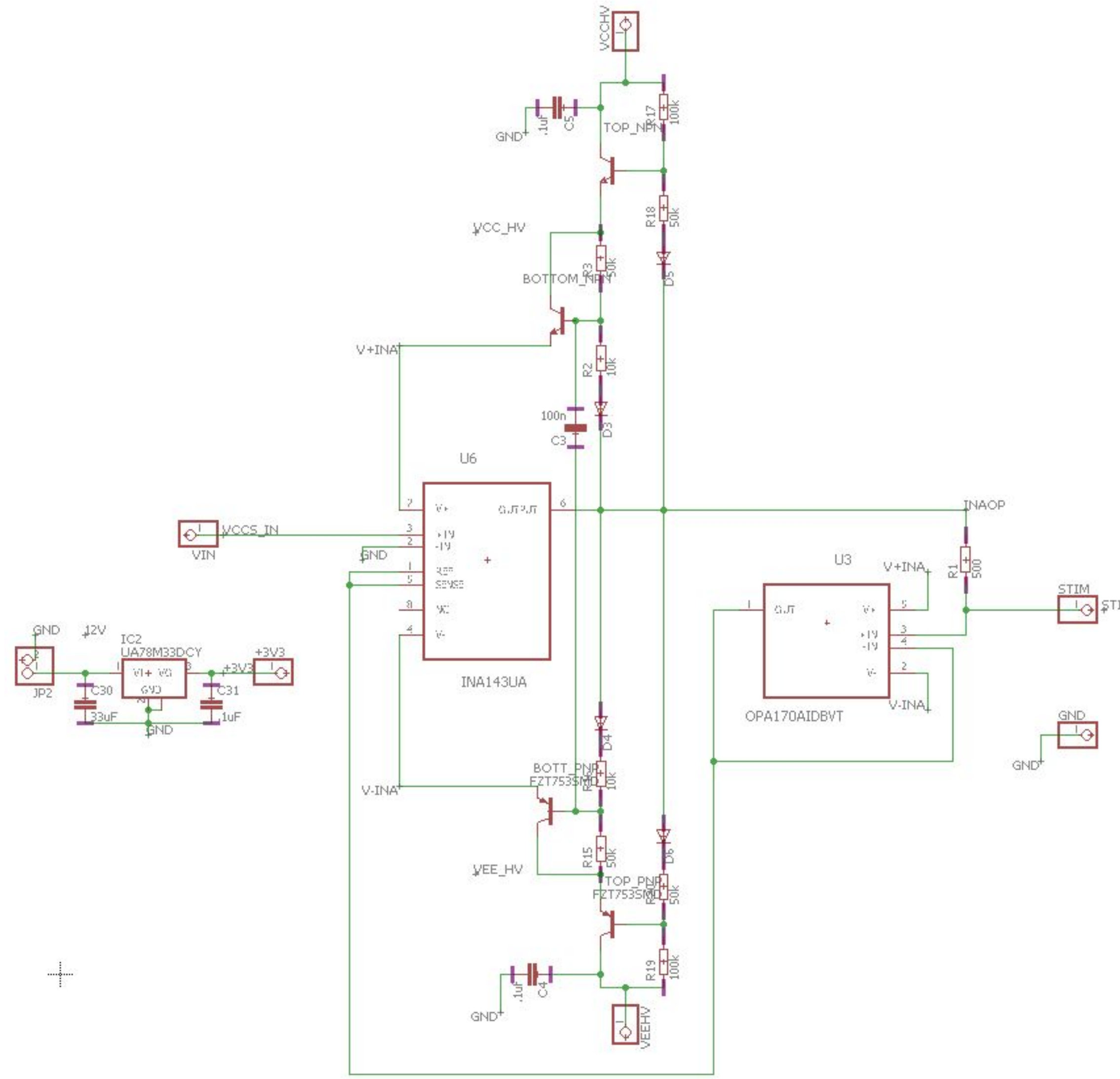

Figure 4: Constant Current Controller

The above circuit depicts the design of the voltage to current converter. The converter consists of two primary components. The first is a howland current source with two modifications, which we adapted from Caldwell [8]. The first is a commonly implemented modification which buffers the output of the traditional howland current source for more robust operation. The second is to connect the REFB and REFA pins to the output of the buffer, which deviates slightly from the classic howland topology without changing the input/output relationship; this was done to allow the +IN and -IN pins on the INA149 to be used, thus allowing us to take advantage of the chip's high common mode rejection. This effect could have also been achieved using an external resistor, but the former method slightly reduces component count and cost without changing the input/output relationship of the device. Solving the system for the relationship between Vin and Iout yields:

$$\frac{V_{in}}{R_{gain}} = I_{out}$$

Eq 1

Where Vin is generated from the DAC, R_{gain} is R_1 in the above schematic, and I_{out} is the output current, which will be actively maintained by the control circuitry. The derivation of this equation uses an approximation of an ideal operational amplifier, and therefore only captures steady state behavior and not the transient controller response. In our device, the gain resistor value is 666 ohms, allowing the min/max current for the device to be +/-15mA

The second component is the bootstrap network, which we adapted from both Brown and Caldwell [7] [8]. This network allows the compliance voltage of the device to be significantly increased, without having to resort to expensive high voltage operational amplifiers. The bootstrap network works by sweeping the op-amp supply rails as a linear function of the the output voltage. Our device uses a cascaded bootstrap network, where the high voltage rails are stepped down in two stages before reaching the INA149 an OPA170.

The figure below was produced by a SPICE simulation of the bootstrap cascade sweeping the supply rails (Vn and Vp), as well as the voltage drop of the stimulator across a 10k load (Vo) with a +/-1V 50Hz sinusoidal input. The peak current in this figure is 1.5mA, which can be found by applying equation 1.

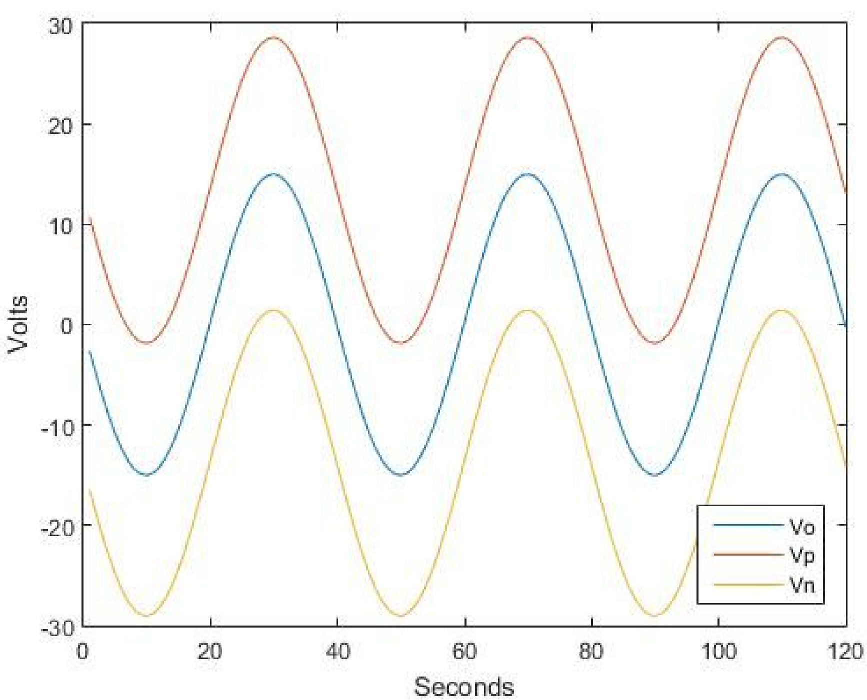

Figure 5: Stimulator output and bootstrap cascade.

2.3.2 Power Supply

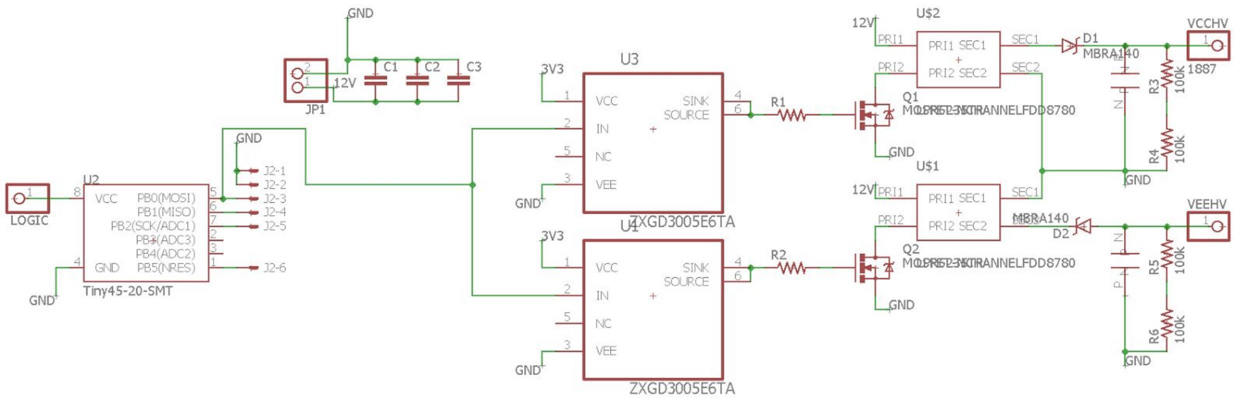

Figure 6: Power supply schematic

The above circuit depicts the design of the high voltage power supply. The design drives two small, high turn ratio transformers with high current mosfets. One transformer is rectified such that the output voltage is positive, and the other is rectified such that the output voltage is negative. We will use high impedance bleed resistors on the output to stabilize the output of the transformer. The coils will be driven via mosfet drivers with an ATTINY45 microcontroller as the switching source, to allow us to programmatically control the power dissipated in the primary coils of the transformers.

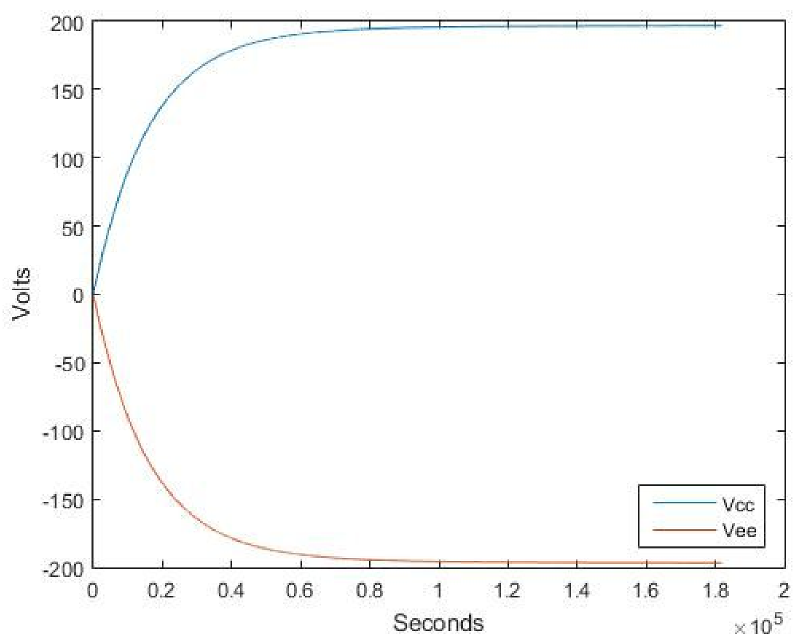

Figure 7: Power supply startup

Pictured above is a SPICE simulation showing the power supply rails during startup. The rails are particularly well matched due to the relative isolation between the drivers, which means that this power supply could potentially be used with devices more susceptible to common mode latch up.

2.3.3 Pressure sensor

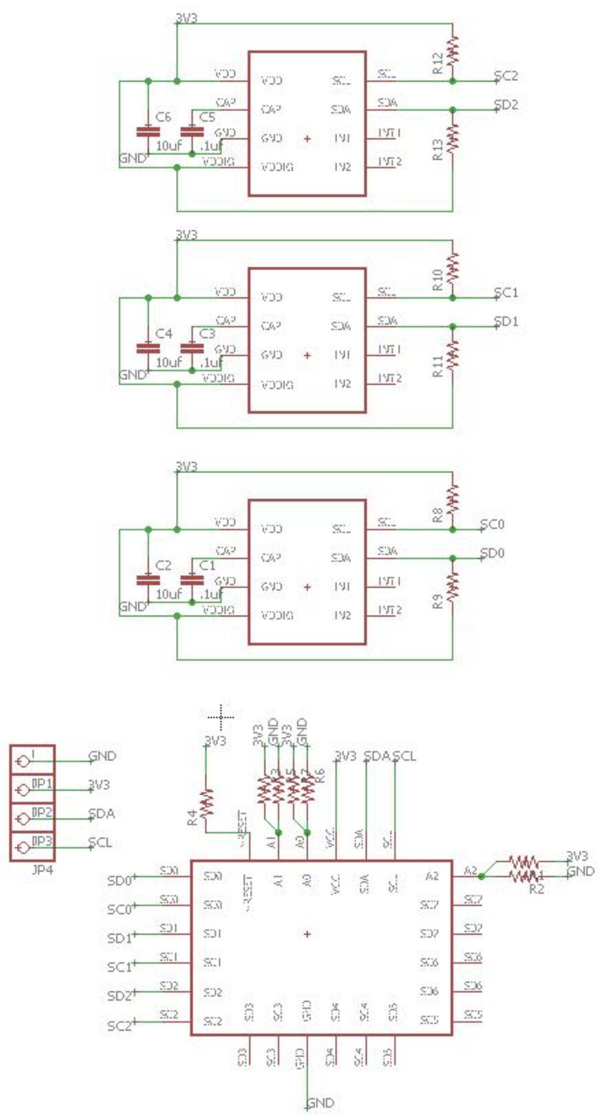

Figure 8: Pressure sensor schematic

The above figure shows the schematic capture for the flexible pressure sensor PCB. The pressure sensor will consist of an I2C mux and 3 MPL3115A2 barometric pressure sensors. The I2C mux has a base address which can take up to 8 values, based on the value of three address pins, allowing up to 8 sets of these 3 fingers to operate on one I2C bus at a time. The layout of this board will be designed to conform to the compliant fingers on PSYONIC's prosthetic hand prototype. Below is an image of the layout of this board:

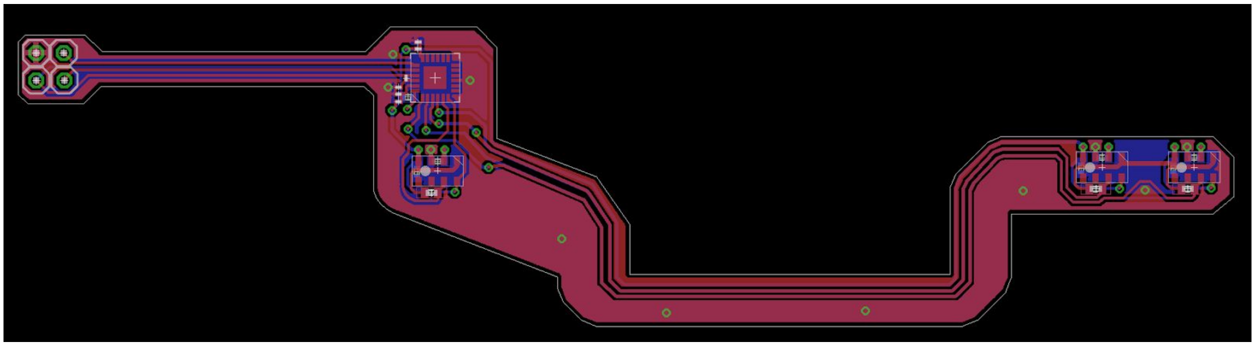

Figure 9: Pressure sensor board layout

2.3.4 DAC board

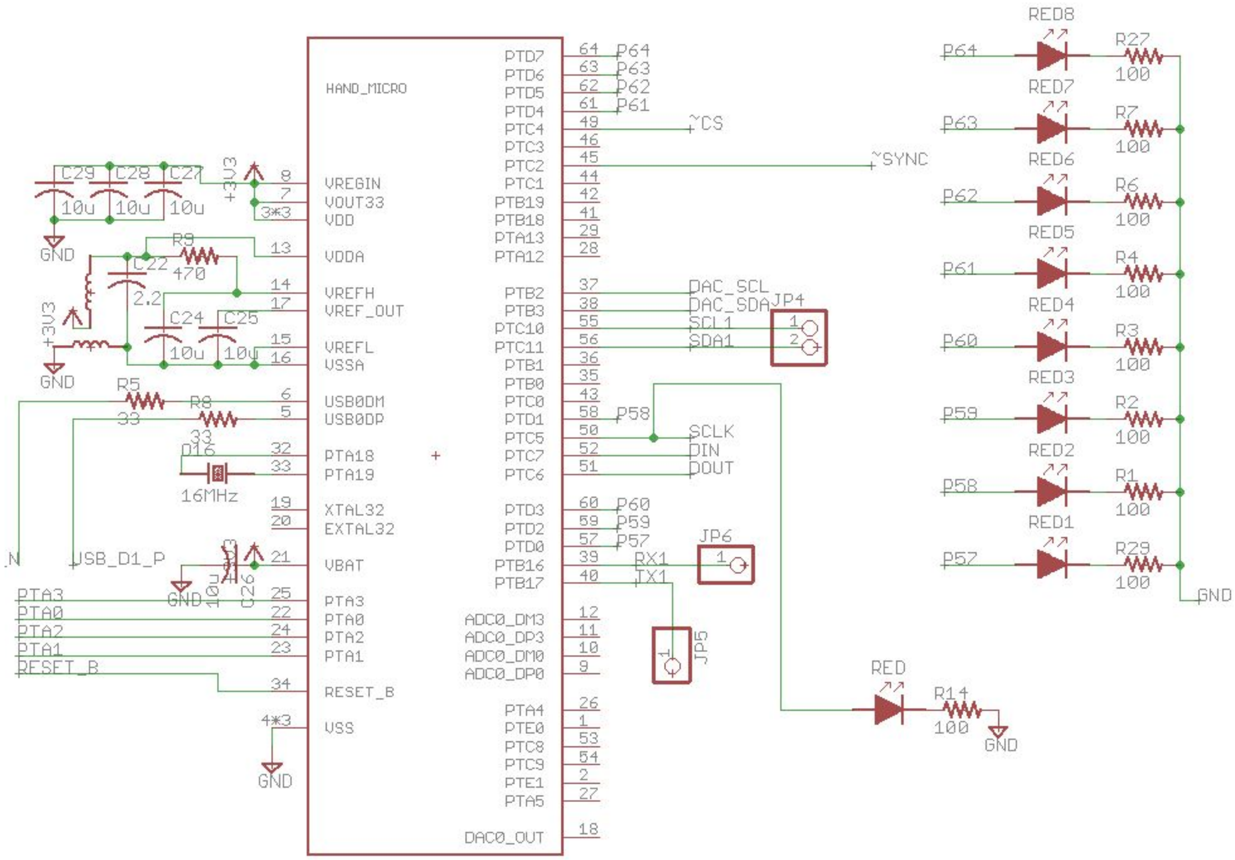

Figure 10.1 DAC board schematic: microcontroller and status LEDS

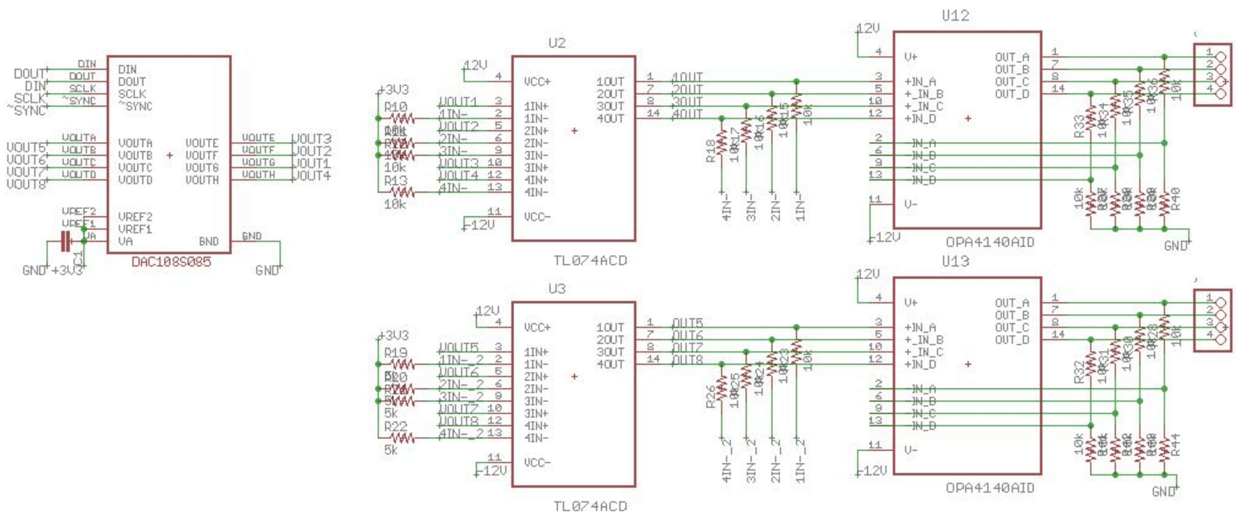

Figure 10.2: DAC board schematic: DAC and post-amplifiers

The figure above shows the schematic for the DAC board. The DAC board will break out an MK20DX256VLH7 microcontroller based on the open-source schematics for the teensy 3.2 microcontroller [9]. The microcontroller have an I2C bus broken out to provide an interface with the pressure sensors, and will interface with a DAC108S085 DAC over SPI on the same PCB. The DAC needs to provide bipolar signals between -10V and 10V, and does not support this natively; to provide bipolar outputs within that range, we will implement an op amp circuit based recommended from the device datasheet which will amplify and center the DAC output, which is natively 0-3.3V [10]. The DAC will also be capable of taking in user input over USB serial, for the purpose of calibrating the stimulation output.

2.5 Software

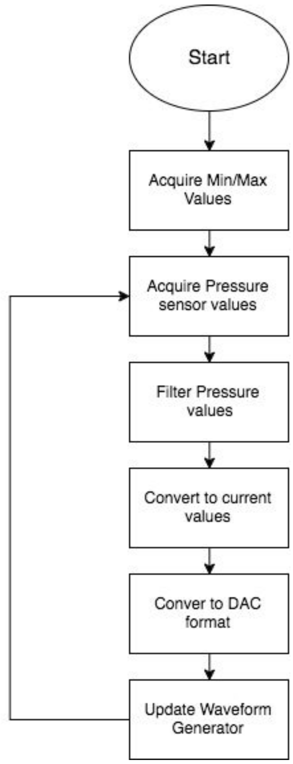

Figure 11: Software Flowchart

Pictured above is a flowchart depicting the software flow for the stimulator when connected to a human subject. The software flow begins by acquiring the minimum and maximum current values for the subject. This process is performed using a GUI on a computer; the DAC will send a waveform with an intensity controlled by the user. The minimum intensity will be recorded when the user barely feels the stimulation and the maximum will be recorded when the user feels a strong stimulation intensity. The pressure sensor values are then polled, post-processed using time domain features such as the standard deviation and mean, and transformed to a current value which ranges between the minimum and maximum recorded values. These desired current values are converted to integer values which represent the magnitudes of the reference voltage waveforms. A separate subroutine will take these magnitudes in and generate the appropriate waveform by updating the registers of the DAC, which produces the appropriate tracking reference for the analog control.

2.6 Tolerance Analysis

In order to properly maintain the rail voltages of the constant current controller, the resistors of the bootstrap network should be properly tuned. The supply voltage produced by the bootstrap cascade is given by the following equations:

$$V_p = \frac{(\frac{(V_{cc}-.6)R_1+V_{out}R_2}{R_1+R_2} - .6)R_3 + V_{out}R_4}{R_3 + R_4} \quad \text{Eq 3}$$

$$V_n = \frac{(\frac{(V_{ee}+.6)R_5+V_{out}R_6}{R_5+R_6} + .6)R_7 + V_{out}R_8}{R_7 + R_8} \quad \text{Eq 4}$$

Where:

$$R_1 = R_5 = R_4 = R_8 = 50K \quad \text{Eq 5}$$

$$R_2 = R_6 = 100K \quad \text{Eq 6}$$

$$R_3 = R_7 = 10K \quad \text{Eq 7}$$

An upper bound on our power supply rails is:

$$V_{cc} = 250V \quad \text{Eq 8}$$

$$V_{ee} = -250V \quad \text{Eq 9}$$

Substituting these values, we come to the final result that:

$$V_p - V_n = 21.9556V \quad \text{Eq 10}$$

Which holds regardless of the value of V_{out} (i.e. V_{out} terms cancel) The above relationship holds for all V_{out} s.t.:

$$V_{ee} < V_{out} < V_{cc} \quad \text{Eq 11}$$

The maximum power supply range of the INA149 is $40V_{pp}$, and the power supply range for the OPA170 is $36V_{pp}$; 21.9V is well within this range, so we can be sure that the bootstrap network successfully protects the devices to which it supplies power by not exceeding their maximum allowable power supply voltage. This result can also be seen in Figure 4, a SPICE simulation showing V_p, V_{out}, and V_n over time.

Common mode latch up, which occurs when the input to an op amp exceeds the rail voltage, is a significant potential risk in constant current . The howland current pump as well as the bootstrapping technique are susceptible to common mode latch up, and if not properly handled, could result in device failure on startup. Our primary method of compensating for this problem is by selecting a difference amp with common mode rejection that exceeds the maximum value of our power supply: the common mode rejection range of the INA149 is +/-275V, and our power supply rails will not exceed +/-250V.

The worst case scenario for device failure is that the user would be exposed to the DC rail voltage, which would occur only if there were multiple failures in the bootstrap cascade and howland current pump. Because the power available to our power supply is limited, this failure would result in a painful but non-lethal shock. This is a significant impact, but it is extremely unlikely that this failure would occur; therefore the risk of using our device is low enough that it should be safe for prolonged use.

3 Costs and Schedule

3.1 Costs

We estimate our development costs to be $30/hour, 12 hours a week for two people. For 16 weeks, the total development cost is:

$$2\frac{\$30}{hr}\frac{12hr}{wk}16wks = \$11520$$

Eq 2

Our estimated component cost for each of the four modules is detailed below:

Part	**Quantity**	**Cost**
LPR6235 (Transformer)	2	$1.40
1uf 0805	3	$.30
6 pin FFC connector	1	$1.07
STD15NF10 (N-MOSFET)	2	$2.40
10uF 250V Capacitor (Custom package)	2	$1.58
MOSFET Driver	2	$1.18
ATTINY45	1	$1.11
10uF 0402 Capacitor	3	$.54
MPL3115A2 (Barometric pressure sensor)	3	$12.96
TCA9548A (I2C Mux)	1	$1.50
10uF 0603 Capacitor	10	$.99
2.2uF 0603 Capacitor	2	$0.20
33uF 0603 Capacitor	1	$0.13
MK20DX256VLH7 (Main Microcontroller/DAC microcontroller)	1	$6.50
UA78M33C (3.3V regulator)	2	$1.16
9 pin FFC connector	1	$1.23
16 Mhz Crystal	1	$0.75
DAC108S085 (DAC chip)	1	$7.59

600 Ohm 0603 Ferrite Bead	2	$0.40
LT1054 (Voltage Inverter)	1	$4.27
TL074 (quad op amp)	2	$1.02
666 Ohm 0603 Resistor	1	$0.10
Gen Purpose SMA Diode	4	$0.96
INA149	1	$6.96
300V NPN SOT-223 Transistor	2	$1.90
300V PNP SOT-223 Transistor	2	$2.04
1KV SMA Schottky Diode	2	$0.94
0 Ohm 0603 Jumper	10	$0.10
1M 1206 Resistor	10	$1.80
0 Ohm 0201 Jumper	10	$.26
1K 0402 Resistor	10	$0.11
.1uF 0402 Capacitor	10	$0.12
33 Ohm 0402 Resistor	10	$0.17
100 Ohm 0402 Resistor	10	$0.12
10K 0402 Resistor	25	$0.19
470 Ohm 0402 Resistor	25	$0.19
LED 0603	10	$1.96
10K 0603 Resistor	10	$0.10
100K 0603 Resistor	10	$0.10
49.9K 0603 Resistor	10	$0.13
.1uF 0603 Resistor	10	$0.18
OPA170 (Op amp)	10	$12.58
Total Component Cost:	$77.89	

We anticipate the combined unit cost of all four PCBs to be <$20, bringing the total device cost at low volume to approximately $97.89. It is difficult to get a good price comparison because there are not currently many comparable devices on the market. The only comparable device (stimulator ONLY not pressure sensor) is the BIOPAC STMISOLA, which is only 1 channel of stimulation without a waveform generator or pressure sensor, and costs approximately $1500.

3.2 Schedule

Week	Jesse	Vikram
02/05/17	Introduce Vikram to the project, high level overview of what needed to be done	Install Teensy Chip Software, Understood the different components of the project
02/12/17	Prepare Mock Design Review doc	Prepare Mock Design Review doc, Set up Teensy software to be arduino compatible
02/19/17	Work on layout, Complete Design Review, finalize schematics	Complete Design Review, Polled data from pressure Sensors
02/26/17	Rev 1. pcb ordered for stimulator, Remake pressure sensor circuit	Understand the circuit design of the pressure sensor module, Remake pressure sensor circuit
03/05/17	When boards arrive, populate rev 1 and test, Complete Soldering assignment	Transform Pressure sensor values to meet Min/Max conditions, Complete Soldering assignment
03/12/17	Debug any potential hardware issues with the power supply and constant current controller, if present. Assist Vikram with coding.	Convert values into current values by using linear transformation
03/19/17	Debug hardware. Verify correctness of DAC output.	Complete waveform generation software
03/26/17	Work on individual progress report.	Debug software if there are issues.
04/02/17	If necessary, order rev 2	Start working on GUI.
04/09/17	Populate and test rev 2 if necessary	Continue GUI work and debug if needed.
04/16/17	Mock Demo Presentation	Debugging DAC and pressure sensors.
04/23/17	Final Demo, Final Paper	Final Demo, Final Paper
04/30/17	Finish Final Paper	Finish Final Paper

4. Safety and Ethics

The most apparent safety risk in this project is the potential delivery of a dangerous shock; this hazard is endemic to devices of this nature. We intend to pay close attention to the IEEE Code of Ethics #1, which is the most relevant to this project; electrical stimulation has significant potential for rehabilitative care, but if mishandled can result in serious injury [6]. To avoid a dangerous shock, we will limit the total power available to the power supply and the maximum amount of current deliverable from the current controller. Before testing the device on a person, we will exhaustively test the device with resistive loads, and verify the thermal and electrical stability of the device. When testing, we will also ensure electrodes are placed in an arrangement which will prevent current from passing through the heart. It is also possible to limit the total deliverable energy in each pulse by AC coupling the output with a capacitor; however we do not think this is a necessary precaution if the listed conditions are met. By limiting the output current, waveform energy, and compliance voltage, we will comply with the standards presented in IEC 601-2-10 for nerve and muscle stimulators.

Another potential risk in this project comes from the use of batteries. If we decide to use lithium ion or lithium polymer batteries in our project, we will follow all safety procedures defined in the ECE 445 battery document. Battery charging is outside the scope of this project, and a battery charging/battery management system will not be implemented in our design; instead, we will use off the shelf protection circuitry and chargers. If disposable 9V alkaline batteries are practical for this device, we will use those instead of lithium ion or polymer batteries.

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