Portable Multi-Channel Electrotactile Haptic Feedback System

Team 2 — Jesse Cornman and Vikram Mudaliar
ECE 445 Project Proposal — Spring 2017
TA: John Capozzo

1. Introduction

1.1 Objective

One of the most fundamental problems with modern myoelectric prosthetic devices is the lack of effective sensory feedback [1], [2]. This problem 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 non-invasively will 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. Therefore, it is important that our device is small, low power, and low cost, to make it viable as a commercial device.

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

The stimulator must be portable
The device must be low cost: <100$ for the entire device
The device needs to be able to support up to 8 channels
The device needs to be modular (channels can be connected and disconnected easily)

2. Design

The stimulator will consist of three different primary modules: the digital to analog converter (DAC), the analog constant current stimulation controller, and the pressure sensors. 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 constant current controller will output a current which will vary as a linear function of the voltage output from the DAC. The pressure sensors will be arranged in clusters of up to 3, and will interface with I2C muxes.

2.1 Block Diagram

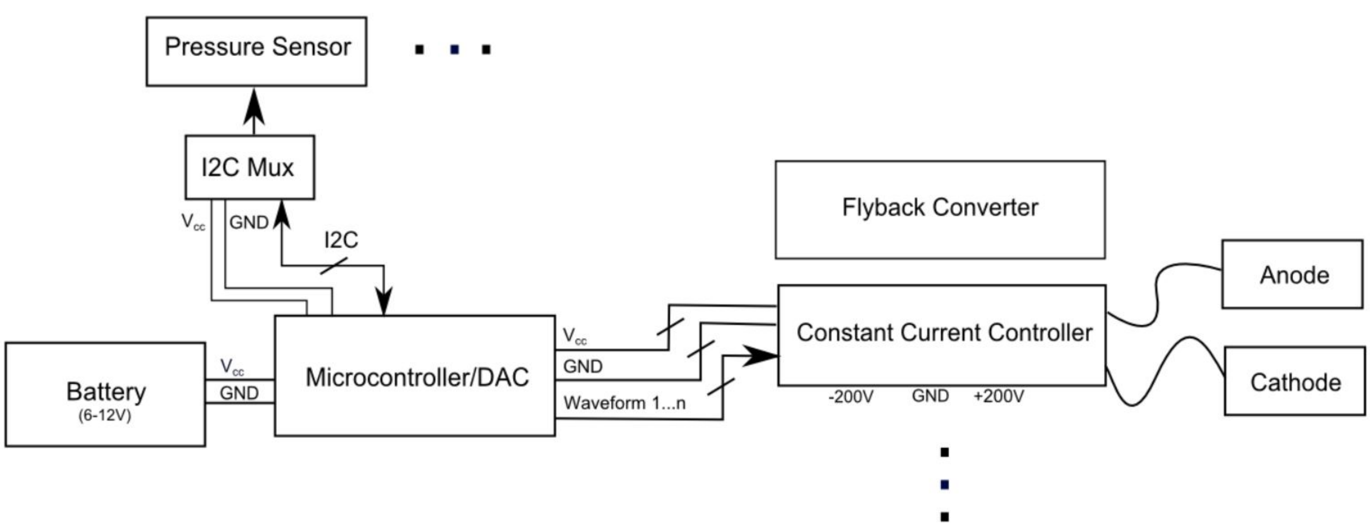

Figure 1. Block Diagram

2.2 Functional Overview

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; 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.

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 1: Must be able to provide +/-15mA.
Requirement 2: Compliance voltage must be at least +/-150V.
Requirement 3: Must have fast rise times (<2us).
Requirement 4: Must have noise that is not perceptible in the stimulation output (low ringing, good PSRR).

2.2.2 Power supply

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 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.
Requirement 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.
Requirement 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.

2.2.3 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.We will reference the open source hardware schematics from teensy 3.2 to break out a MK20DX256VLH7 microcontroller to interface with the DAC chip.

Requirement 1: The outputs need to be able to 'queued', meaning all the pulses have precisely the same phase; this is so the signal can be easily ignored while acquiring EMG signals in a prosthetic.

Requirement 2: 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.

2.2.4 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 1: The pressure sensors must not drift significantly; i.e. they always settle to the same baseline value.

2.2.5 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.

2.3 Risk Analysis

Common mode latch up is a significant potential risk in this project. 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. We will compensate for this problem by selecting operational amplifiers with a high common mode rejection ratio, and for further redundancy, by designing a power supply with well-matched rails (one rail does not power on before the other).

Pressure sensor drift and hysteresis are unwanted effects that we have to watch for, and could arise from mechanical properties of the sensor. If the sensor drifts from its baseline over time, the stimulator may start to send stimulation even when no contact has been made. If we encounter this problem, we will try various signal processing methods to try and rectify it, such as band pass filtering to remove low frequency artifacts, etc. Hysteresis will limit the expressiveness of the sensor; for instance, a fast sequence of taps may not be registered. Based on our evaluation of sensors sold by TakkTile, if our manufacturing technique is correct we should be able to avoid significant hysteresis.

3. 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.

References

[1] Biddiss E, Beaton D, Chau T. Consumer design priorities for upper limb prosthetics. Disabil. Rehabil. Assist. Technol. 2(6), 346-357 (2007).

[2] Kyberd PJ, Wartenberg C, Sandsjö L et al. Survey of upper-extremity prosthesis users in Sweden and the United Kingdom. J. Prosthet. Orthot. 19(2), 55-62 (2007)

[3] April Lauer, Kathy Longenecker Rust, Roger O. Smith. Factors in Assistive Technology Device Abandonment: Replacing “Abandonment” with “Discontinuance”. (2015) [Online]. Available: http://www.r2d2.uwm.edu/atoms/archive/technicalreports/tr-discontinuance.html. [Accessed: 07- Feb- 2017].

[4] C. Antfolk, M. D’Alonzo, B. Rosén, G. Lundborg, F. Sebelius, C. Cipriani, Sensory feedback in upper limb prosthetics. (2013) [Online]. Available: https://pdfs.semanticscholar.org/5cf2/d32093fe0d37de39c399ed1d6cef9f6ae884.pdf [Accessed: 07- Feb- 2017].

[5] H. Kajimoto, “Electro-tactile Display: Principle and Hardware”, in *Pervasive Haptics*, Tokyo, Japan; Springer, 2016, ch. 5 sec. 5.3.1, pp. 89.

[6] Ieee.org, "IEEE IEEE Code of Ethics", 2016. [Online]. Available: http://www.ieee.org/about/corporate/governance/p7-8.html. [Accessed: 8-Feb- 2017]