# A Portable, Arbitrary Waveform, Multichannel Constant Current Electrotactile Stimulator

Jesse Cornman, Aadeel Akhtar, and Timothy Bretl

***Abstract*— In this paper, we present the design and performance of a portable, arbitrary waveform, multichannel constant current electrotactile stimulator that costs less than $30 in components. The stimulator consists of a stimulation controller and power supply that are less than half the size of a credit card and can produce ±15 mA at ±150 V. The design is easily extensible to multiple independent channels that can receive an arbitrary waveform input from a digital-to-analog converter, drawing only 0.9 W/channel (lasting 4-5 hours upon continuous stimulation using a 9 V battery). Finally, we compare the performance of our stimulator to similar stimulators both commercially available and developed in research.**

## I. Introduction

Electrotactile stimulation involves the application of current over the skin to elicit sensations such as vibration, touch, tingling, itching, pinching, pressure, and pain [1]. It is useful as a method for delivering information via haptic feedback to a user, such as touch/pressure feedback in a virtual reality setting [2] or in a hand prosthesis [3]. To elicit the most expressive sensations to a user with the most consistency over time, it is important that the stimulator can produce arbitrary waveforms using constant current stimulation [1], [4]. In order to make this type of stimulation accessible to everyone, it is also important to keep stimulators low-cost.

Many stimulators have been developed in research to study electrotactile stimulation. Two of the most prominent are from Poletto & Van Doren [5] and Schaning & Kaczmarek [6]. Commercial stimulators are also available, including the STMISOLA (BIOPAC Systems, Goleta, CA) and transcutaneous electrical nerve stimulators (TENS) such as the TENS 7000 (VQ OrthoCare, Irvine, CA). However, these stimulators consume too much power, are too large, are too expensive, or cannot produce arbitrary waveforms.

In this paper, we present the design and performance of a portable, arbitrary waveform, multichannel constant current electrotactile stimulator that costs less than $30 in components (Fig. 1). The paper is organized as follows: in Sec. II, we describe the design of our stimulation controller and power supply; in Secs. III-IV, we characterize the performance of our stimulator and compare our results to other stimulators with respect to power consumption, compliance voltage, peak current, rise time, cost, ability to produce an arbitrary waveform, and size. We also discuss the safety of our stimulator. In Sec. V, we conclude with improvements we can make to our stimulator in the future.

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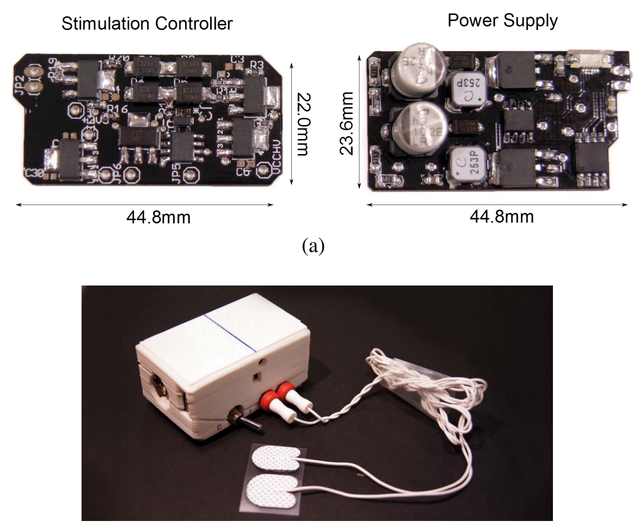


(a)

(b)

Fig. 1. Our stimulator consists of (a) a stimulation controller and a power supply that can attach to each other to form a single stimulation channel. (b) The stimulation channel can be housed with a 9 V battery, microcontroller, and electrodes in a compact, portable module.

## II. Design

The block diagram in Fig. 2 shows the overall layout of the device. We provide stimulation through an analog constant current controller coupled to a switched-mode DC-DC converter that boosts voltage from a battery pack (6-12 V) to ±200 V. The physical layout and specifications of our stimulator were motivated by the requirement of a portable multichannel stimulator that could be used in a wide variety of applications, especially for prosthetic hands. A separate microcontroller and digital-to-analog converter (DAC) board controls each stimulation channel independently. The stimulation channels were physically separated from the DAC board to allow for flexibility in the placement of the microcontroller and individual channels. This layout also gives users the ability to freely change the number of channels used for stimulation as necessary, as the bulk of the power consumption of the stimulator comes from the coupled stimulation controllers and high voltage DC-DC converters.

### A. Stimulation Controller

Figure 3 shows a schematic of our stimulation controller. To convert $V_{in}$ from a DAC to a constant current output, we used an improved Howland current source design due to its bipolar operation and fast response times [5], [7].

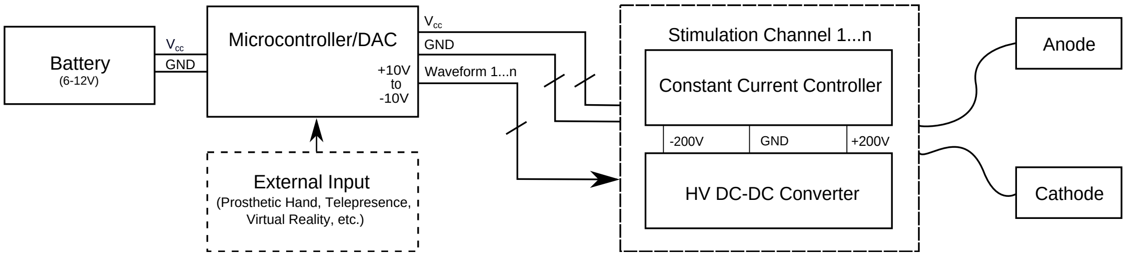


Fig. 2. Hardware block diagram showing a high-level overview of our stimulator.

The voltage-to-current input-output relationship of the improved Howland current source can be tuned using a single sense resistor, allowing the use of an integrated difference amplifier with high-precision internal resistors. This design is desireable because it reduces error in the voltage-current relationship resulting from lower-precision resistors [7].

Because skin impedances can typically range from 10 kΩ to 100 kΩ on the forearm [4], electrotactile stimulators need to handle high voltages (typically between 150-300 V [6]). Conventional amplifier integrated circuits have limited supply voltages ($\leq \pm 18$ V), so a bootstrap network [8] was implemented, which linearly sweeps the supply rails of the operational amplifiers in the improved Howland current source as a function of the voltage drop across the load, and allows for a significant increase in compliance voltage. An improved Howland current source has been used for constant current electrotactile stimulation by Poletto & Van Doren [5]; however, they used high voltage operational amplifiers in a bridge configuration in order to achieve a high compliance voltage. The use of high voltage operational amplifiers is undesirable for three reasons. First, the amplifiers used (PA85A, Apex Microtechnology, Tucson, AZ) are expensive ($>$ \$200). Second, these amplifiers typically draw 21 mA quiescent current, resulting in approximately 9 W dissipated per amplifier in their circuit (+430 V, -15 V rails) with no load. Finally, the use of a bridge configuration to double the compliance voltage requires floating the load, which is undesirable for safety reasons [6]. Our bootstrap technique is inexpensive to implement ($<$ \$5) and can achieve the same compliance voltages without having to float the load.

A major issue with the improved Howland current source is its susceptibility to common-mode latch up [7]. This issue can be resolved in general with careful power supply sequencing [5], [8]. However, the simplest solution is to take advantage of the availability of inexpensive, high common-mode voltage integrated difference amplifiers, such as the INA149 (Texas Instruments, Dallas, TX), AD8479 (Analog Devices, Norwood, MA), or LT6375 (Linear Technologies, Milpitas, CA). Our selection of the INA149 was based on the implementation of a bootstrapped Howland current source described by Caldwell [9], which demonstrates how to use the high common-mode voltage feature to protect a device from common-mode latch up without changing the voltage-to-current input-output relationship. The input-output relationship of our improved Howland current source is $I_{out} = V_{in}/R_s$. $V_{in}$ can vary between $\pm 10$ V. By choosing $R_s$ to be 500 Ω, our improved Howland current source can produce currents ($I_{out}$) between $\pm 20$ mA. However, due to restrictions placed in our power supply, the functional output range of the device is between $\pm 15$ mA (see Sec. IV-A). In a typical bootstrap topology [9], only two bootstrapped transistors are used. When using this topology to set the $\pm 15$ V rails of the operational amplifiers, the maximum possible power dissipated (assuming the output is shorted) in each bootstrapped transistor would be approximately 3.7 W, requiring large thermal pads to protect the transistors from damage. To resolve this issue we implemented a 2-stage cascaded version of the bootstrap network [8], which reduced the voltage drops across each transistor, keeping the maximum power dissipation below 2.5 W in any single transistor. This technique can be scaled to increase the compliance voltage of the circuit as well, while still using small transistors with relatively low voltage and power ratings (300 V, 3 W).

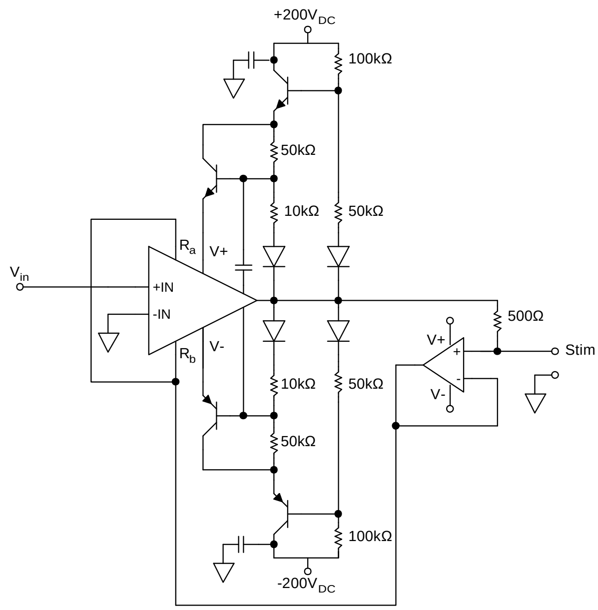


Fig. 3. Schematic of the stimulation controller with an improved Howland current source and cascaded bootstrap network.

## B. Power Supply

In order to produce stimulation signals at an acceptable compliance voltage, we required a DC-DC converter which could be powered from a battery pack and would produce $\pm 200$ V. Figure 4 shows a schematic of our power supply design. Due to the high current and voltage gain require-

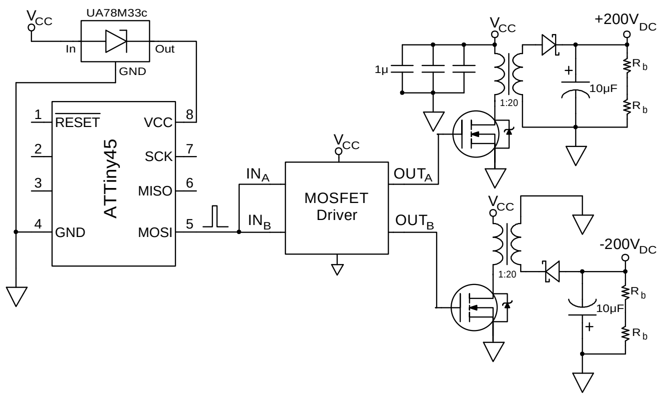


Fig. 4. Schematic of the ±200 V power supply. The power supply is a flyback converter that takes in a low DC voltage ($V_{CC}$ = 6-12 V) from a battery and converts it to two unregulated high DC voltages (±200 V). The ATTiny45 serves as a switching controller for the two 1:20 transformers.

ments, we used a flyback converter design. In our design, two separate 1:20 turns ratio transformers (LPR6235-253PMR, Coilcraft, Cary, IL) produced the positive and negative high voltage rails, with the primary coil driven at 31.7 kHz with a 5% duty cycle between 6-12 V.

To prevent any potential safety issues arising from routing ±200 V wires across a user's body, the power supplies were designed to be efficient and collocated with the stimulation controller. We used an ATTiny45 microcontroller (Atmel, San Jose, CA) as the switching controller on the power supply, which is programmable and can easily be modified for voltage regulation or limiting the total power available to the device. However, because the power supply output voltage does not affect the functionality of the stimulation controller as long as it stays relatively close to ±200 V, active voltage regulation is not necessary for this device.

## III. Performance

We tested our stimulator with a $V_{in} = 0.5$ V ($I_{out} = 1$ mA), 1 ms square pulse input across four different resistive loads (10 kΩ, 25 kΩ, 50 kΩ, and 100 kΩ) and recorded the steady state voltage. The voltage drop and current across each load are shown in Figs. 5a-5b, respectively. Table I compares characteristics of our stimulator to other stimulators that were developed in research (Poletto & Van Doren [5], Schaning & Kaczmarek [6]) or are commercially available (STMISOLA, TENS 7000).

We observed rise times of $< 2$ µs (comparable to those of Poletto & Van Doren [5] and Schaning & Kaczmarek [6]) and overshoot ranging between 0-30% of the target current (higher at lower resistances). Similar overshoot was found in the stimulator of Poletto & Van Doren, but due to the short duration and therefore negligible charge, the overshoot is of no physiological significance [5]. The primary strength of our stimulator lies in the power consumption, size, and cost. Furthermore, our stimulator and the STMISOLA are the only stimulators listed that have been designed to produce arbitrary waveforms.

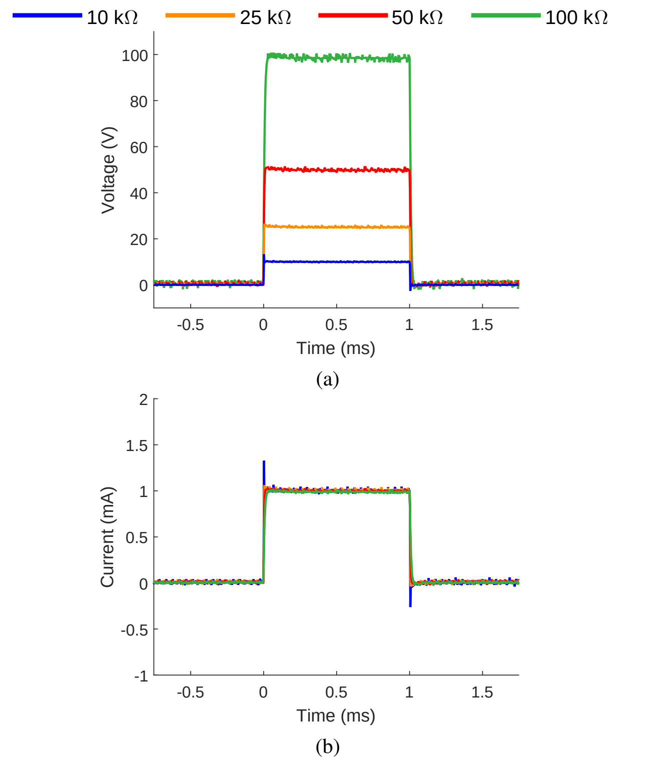


Fig. 5. (a) The voltage drop and (b) current across four resistive loads in response to a $V_{in} = 0.5$ V ($I_{out} = 1$ mA), 1 ms square pulse input.

## IV. Discussion

### A. Power

While our power supply can deliver ±200 V, we limited the duty cycle of the square wave driving the transformer to a conservative 5% in order to prevent overheating in the transformer. Limiting the duty cycle resulted in a small average power in our power supply circuit, reducing our output compliance voltage to ±150 V at ±15 mA. However, the output compliance can easily be increased at no additional cost in two ways. First, we can increase the duty cycle, increasing the average power delivered to the transformer. However, increasing the duty cycle too much can result in overheating the power supply. Second, if even higher compliance voltages are desired, a transformer with a higher turns ratio could be used, such as the equivalently-priced LPR6235-123QMR 1:50 transformer (Coilcraft, Cary, IL). Changing the transformer would require adjusting the resistor values in the bootstrap network to compensate for the increased rail voltage. We chose not to make these changes because our electrodes are large enough (20 mm x 15 mm) to make this current and compliance voltage acceptable for a wide range of stimulation intensities.

The quiescent power dissipation of one channel, including losses in the ±200 V converter, is only 0.91 W, which is significantly lower than that of comparable devices, such as the STMISOLA, whose quiescent power dissipation we measured to be 2.7 W (including 12 V to ±200 V conversion), the stimulator of Poletto & Van Doren, which dissipated at least 18 W in the amplifier alone [6], as well as the stimulator of Schaning & Kaczmarek, which dissipated 1.6 W in the

TABLE I

COMPARISON OF DEVICE CHARACTERISTICS BETWEEN VARIOUS STIMULATORS.

| Stimulator | Quiescent Power (W/channel) | Compliance Voltage (V) | Peak Current (mA) | Rise Time (μs) | Cost (USD) | Arbitrary Waveform? (Y/N) |
|---|---|---|---|---|---|---|
| Our stimulator | 0.9 | $\pm$150 | $\pm$15 | $< 2$ | $< \$30$ | Y |
| Poletto & Van Doren [5] | >18 | 800 | 25 | $< 1$ | $> \$1300$ | N |
| Schaning & Kaczmarek [6] | 1.6 | $\pm$600 | $\pm$20 | $< 2$ | $> \$700$ | N |
| BIOPAC STMISOLA | 2.7 | $\pm$200 | $\pm$100 | $< 10$ | $1500* | Y |
| TENS 7000 | - | 50 | 100 | - | $30* | N |

*These costs reflect the list prices of these stimulators rather than the costs of components.

amplifier [6]. The power dissipation could be further reduced if the same mobile power supply was used with an amplifier that is not arbitrary waveform (i.e. only square pulses), and even further if the device produced constant voltage pulses instead of constant current pulses, as is commonly found in most commercial TENS units. Regardless, in its current state a single channel will operate for approximately 4-5 hours continuously from a conventional 9 V alkaline battery.

### B. Size

The reduction in size of our stimulator when compared to others is also significant. Each stimulation channel is 44.8 mm x 23.6 mm x 15.4 mm (length x width x height) (Fig. 1), approximately 15x smaller than the amplifier channel of the stimulator of Schaning & Kaczmarek [6]. While Poletto & Van Doren do not report a size, based on their components, our stimulator would be significantly smaller. Compared to the BIOPAC STMISOLA, our stimulator is 50x smaller, though their stimulator was not designed to be portable. While TENS units are small enough to be handheld and portable, they lack arbitrary waveform outputs and are usually constant voltage, which limits the expressiveness and consistency of the stimulation delivered to users [1].

### C. Cost

Our stimulator is also significantly cheaper than most comparable stimulators. Bootstrapping allowed us to take advantage of inexpensive low voltage operational amplifiers to achieve controller performance similar to that of Poletto & Van Doren without having to resort to expensive and power-hungry high voltage operational amplifiers. A significant component of the stimulator costs for both Poletto & Van Doren and Schaning & Kaczmarek comes from the use of expensive power supplies and DC-DC converters. Even without considering the costs of the power supplies, the stimulation controller used by Schaning & Kaczmarek alone costs more than $60, and that of Poletto & Van Doren costs more than $550.

### D. Safety

Our stimulator is battery powered, thus removing the need for complex AC-mains isolation schemes [6]. Additionally, the output could be capacitively coupled to the user to limit the total deliverable charge if the amplifier or input source were to fail [5]. Additional features, such as no-load detection on the output and output suppression can also be added to shut off the device if the electrodes become disconnected while the stimulator is powered.

## V. CONCLUSION

We presented the design and performance of an inexpensive, portable multi-channel electrotactile stimulator capable of delivering arbitrary constant current waveforms between $\pm$15 mA at $\pm$150 V. The design of this stimulator is highly flexible, allowing for tradeoffs between power consumption and compliance voltage. Future work will involve changing the amplifier, transformer, and resistor values to increase power efficiency and compliance voltage. Finally, we plan to test this stimulator on patients with upper limb amputations to evaluate long-term sensorimotor prosthetic control.

## VI. ACKNOWLEDGEMENTS

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## REFERENCES

[1] K. A. Kaczmarek, J. G. Webster, P. B. y Rita, and W. J. Tompkins, "Electrotactile and vibrotactile displays for sensory substitution systems," *IEEE Trans. Biomed. Eng.*, vol. 38, pp. 1–16, Jan 1991.

[2] D. Pamungkas and K. Ward, "Electro-tactile feedback system to enhance virtual reality experience," *Int. J. Comp. Theory Eng.*, vol. 8, no. 6, p. 465, 2016.

[3] A. Akhtar, K. Y. Choi, M. Fatina, J. Cornman, E. Wu, J. Sombeck, C. Yim, P. Slade, J. Lee, J. Moore, D. Gonzales, A. Wu, G. Anderson, D. Rotter, C. Shin, and T. Bretl, "A low-cost, open-source, compliant hand for enabling sensorimotor control for people with transradial amputations," in *Conf. Prof. 2016 IEEE EMBC*, pp. 4642–4645, Aug 2016.

[4] A. Akhtar, B. Boyce, and T. Bretl, "The relationship between energy, phase charge, impedance, and perceived sensation in electrotactile stimulation," in *Conf. Proc. 2014 IEEE HAPTICS*, pp. 69–74, Feb 2014.

[5] C. J. Poletto and C. L. V. Doren, "A high voltage, constant current stimulator for electrocutaneous stimulation through small electrodes," *IEEE Trans. Biomed. Eng.*, vol. 46, pp. 929–936, Aug 1999.

[6] M. A. Schaning and K. A. Kaczmarek, "A high-voltage bipolar transconductance amplifier for electrotactile stimulation," *IEEE Trans. Biomed. Eng.*, vol. 55, pp. 2433–2443, Oct 2008.

[7] P. Horowitz, W. Hill, and I. Robinson, *The Art of Electronics*. Cambridge University Press, 3rd ed., 2015.

[8] G. King and T. Watkins, "Bootstrapping your op amp yields wide voltage swings." EDN, May 13 1999.

[9] J. Caldwell, "A high-voltage bidirectional current source," tech. rep., Texas Instruments, 2013.