Biomedical Signal Processing and Control 6 (2011) 346–355

Contents lists available at ScienceDirect

Biomedical Signal Processing and Control

journal homepage: www.elsevier.com/locate/bspc

On the design of a flexible stimulator for animal studies in auditory prostheses☆

Douglas Kim [a], Vanishree Gopalakrishna [a], Song Guo [a], Hoi Lee [a], Murat Torlak [a], Nasser Kehtarnavaz [a], Arthur Lobo [b], Philipos C. Loizou [a,*]

[a] *University of Texas at Dallas, Department of Electrical Engineering, Richardson, TX 75080-3021, USA*
[b] *Signals and Sensors Research, Inc., McKinney, TX 75071-3191, USA*

ARTICLE INFO

Article history:
Received 1 June 2010
Received in revised form 24 January 2011
Accepted 11 May 2011
Available online 12 June 2011

Keywords:
Current stimulator
Bipolar current stimulation
Benchtop current stimulator
Portable research platform
Percutaneous animal studies
Auditory
Cochlear implant

ABSTRACT

The present paper describes the design of two stimulators (bench-top and portable) which can be used for animal studies in cochlear implants. The bench-top stimulator is controlled by a high-speed digital output board manufactured by National Instruments and is electrically isolated. The portable stimulator is controlled by a personal digital assistant (PDA) and is based on a custom interface board that communicates with the signal processor in the PDA through the secure digital IO (SDIO) slot. Both stimulators can provide 8 charge-balanced, bipolar channels of pulsatile and analog-like electrical stimulation, delivered simultaneously, interleaved or using a combination of both modes. Flexibility is provided into the construction of arbitrary, but charge-balanced, pulse shapes, which can be either symmetric or asymmetric.

Published by Elsevier Ltd.

1. Introduction

The field of cochlear implants has experienced considerable growth over recent years largely stimulated and driven by research on both human subjects and animals. Access to a flexible research platform is critical for the advancement of cochlear implants. While most implant manufacturers provide research speech processors for use on human subjects that allow researchers to develop and test new signal processing algorithms [1–3], most research labs are unable to use them due to limited technical resources. Furthermore, of those research processors, only one made by CRC/Hearworks (SPEAR3 processor) is wearable and portable [4–6], whereas the research processors provided by the other manufacturers are constrained for use only in laboratory environments.

While the availability of portable research speech processors for human studies is largely limited, the availability of portable multi-channel current stimulators for chronic neurophysiological experiments on animals are far fewer. Given that the performance of CI users may improve or change within a period of a few months, it is necessary that novel algorithms be evaluated under long-term use of the device. Such evaluations would provide more realistic assessments of the performance of new algorithms and new stimulation paradigms.

The design of several stand-alone current simulator sources have recently been proposed [7,8], but none that have been integrated into a complete system that is both portable and provides researchers with the ability to generate arbitrary biphasic current waveforms over a wide array of stimulation parameters. Some recently proposed stimulator designs integrate onto a complete system-on-chip (SOC) both the stimulator as well as the speech processor DSP core into a single implantable device [9,10]. While portable, these systems are intended for use on humans rather than animals and are not designed for research experimentation, and use DSP cores that execute a single fixed speech-processing algorithm. Systems more suitable for experimentation must be capable of implementing new and novel algorithms with relative ease such that researchers with limited programming skills are able to use them effectively. Such a system would also be valuable in animal studies assessing long-term effects of electrical stimulation, and would therefore bridge the gap and accelerate the transition from animal studies to clinical applications. The inability to easily reconfigure these systems with novel algorithms and their large form factor relative to the anatomy of feline or small primate species largely limits the use of these systems for animal studies.

In [11], we proposed the use of Personal Digital Assistants (PDAs) as a research platform for both human and animal studies

☆ Expanded Paper from the work presented at the 2nd International Symposium on Applied Sciences in Biomedical and Communication Technologies held 24 September 2009.

* Corresponding author.
E-mail address: loizou@utdallas.edu (P.C. Loizou).

1746-8094/$ – see front matter. Published by Elsevier Ltd.
doi:10.1016/j.bspc.2011.05.008

in cochlear implants. This research platform includes a stimulator unit which can be used for electrical stimulation in animal studies, a recording unit for collecting evoked potentials from human subjects and a portable processor for implementing and evaluating novel speech processing algorithms after long-term use. In the present paper, we describe the development of a bench-top stimulator designed for neurophysiological studies with animals (further details regarding the development of the portable speech processor designed for human studies can be found on our website[1]). We also describe ongoing work with a portable stimulator, which can be used for chronic animal studies. This PDA-based stimulator can allow the animals to roam freely in a laboratory environment without being tethered or confined in any way.

2. Bench-top stimulator: system overview and methodology

The bench top, bipolar stimulator or BT-BiSTM is a multichannel bipolar current source designed for acute experiments on percutaneous, animal cochlear implant systems. Created as the precursor to the portable version of the stimulation system, the bipolar configuration offers researchers the ability to study the effects of channel interactions on speech recognition particularly as a function of the electrode array configuration. Distortions in a desired amplitude envelope for frequency bands of interest arise as a result of electrical-field summations between nearby electrodes during simultaneous stimulation [12]. Narrowly spaced electrodes or bipolar pairs can be used in studying neural activation patterns relative to broader electric fields generated by bipolar pairs separated further apart from one other. Furthermore, while the BT-BiSTM is intended to be used primarily for bipolar stimulation, it is also capable of generating up to eight independent, time interleaved monopolar signals. Therefore, studies on the effects of time interleaved monopolar stimulation on speech perception can also be made with use of this device. Support for fully simultaneous as well as interleaved monopolar stimulation will be available in an upcoming monopolar version of the current stimulation system described briefly in Section 7.

The BT-BiSTM is a highly versatile platform capable of generating up to 8 simultaneous channels over a wide array of excitation patterns including both pulsatile and analogue-like, or combinations of both. Built around a 9-bit configurable current source chip [13], the BT-BiSTM platform possesses the following specifications:

- 8 independently controlled bipolar channels or up to 8 independently controlled *time interleaved* monopolar channels,[2] each electrically isolated and charge-balanced
- 5 V compliance voltage
- 1 mA maximum current amplitude per channel
- 9-bit current amplitude resolution per channel
- 4 μs minimum pulse width per channel
- 0 μs minimum interphase gap per channel
- 4 μs minimum interstimulus interval per channel
- 83.3 kHz maximum pulse rate per channel
- >50 MΩ output resistance per channel

With this platform, a wide array of stimulation techniques for cochlear implants can be tested on animals. By varying parameters such as current amplitude, pulse width, interphase gap, interstimulus interval (ISI) and pulse rate, a multitude of stimulation patterns can be created both in phase (simultaneous) or interleaved across multiple channels.

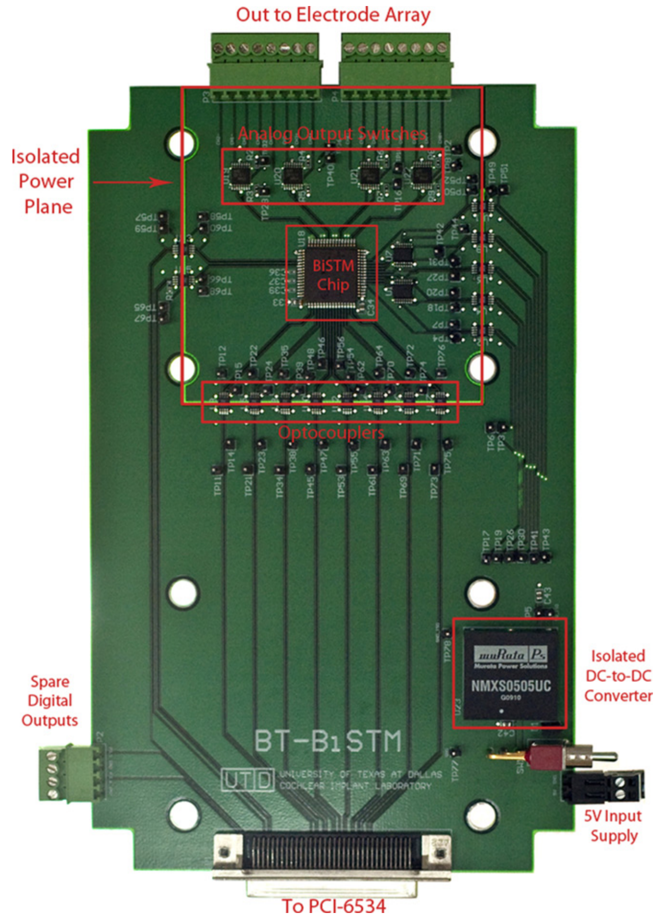

Fig. 1. BT-BiSTM 8-channel bipolar stimulator board.

Depending upon the demands of particular applications, the BT-BiSTM is capable of generating up to eight independently controlled and highly complex bipolar signals each varying in pulse width, interphase gap, pulse rate and current amplitude. A portion of the output channels can be generated simultaneously in phase while others can be interleaved in time. The time duration of a single cycle of such complex stimulation patterns (multiple cycles can continuously be generated in a periodic fashion for extended periods of time) is a function of the shortest non-zero time constraint of the waveform – typically the interphase gap. For instance, if the minimum required interphase gap of a channel output is 2 μs then the output sampling rate of the bipolar stimulator chip, that is the rate at which a single analog current output sample can be generated by the onboard current stimulator chip, must be chosen sufficiently high enough in order to maintain an adequate time resolution for the specified interphase gap. Chosen as such is an output sampling rate of 0.1 μs which corresponds to 12.8 s of a single stimulation pattern for all 8 channels that can be stored in memory and then continuously regenerated in a loop-like fashion over extended periods of time. For applications with looser time constraints that can be met with slower output sampling rates, even lengthier stimulation patterns can be stored and generated.

In addition to the 8-channel current source BT-BiSTM board shown in Fig. 1, the software needed to control the board is also available in an easy to use and user-friendly graphical user interface (GUI) built on top of the National Instruments (NI) graphical programming environment, LabVIEW. When combined with NI's

[1] http://www.utdallas.edu/~loizou/cimplants/PDA/.

[2] Support for fully simultaneous as well as interleaved monopolar stimulation will be available in an upcoming monopolar version of the current stimulation system described briefly in Section 7.

32 channel, high speed digital output board, the PCI-6534,[3] which serves as the hardware control interface between any PC equipped with a standard PCI slot and the BT-BiSTM board, the BT-BiSTM GUI greatly simplifies the task of using the board in order to output a desired set of electrical stimulation patterns by eliminating the need of users to program the board. Waveform parameters such as pulse rate or pulse width may simply be entered into the GUI for a desired set of stimuli without knowledge of the underlying sequence of digital control signals needed to control the BT-BiSTM board, thus reducing the time required in learning how to use the platform and allowing researchers to focus their efforts on conducting animal experiments.

3. BT-BiSTM hardware architecture

Shown in Fig. 1 is a photograph of the BT-BiSTM board. At the core of the board is the 9-bit configurable current source chip, simply referred to as the BiSTM chip. The BiSTM chip is designed to provide programmable anodic and cathodic current pulses for stimulation [13]. By using a dynamic biasing scheme, the stimulator can realize 9 bits of resolution with a single 7-bit binary-weighted digital to analog converter (DAC). Hence, good linearity and a small implementation silicon area are achieved simultaneously. Moreover, active cascade output stages are used in the BiSTM chip to achieve high output impedance. Output impedance is further improved with the use of stacking MOS structures which can minimize hot-carrier effects and maintain output current accuracy through large voltage compliance.

Each of the board's 8 bipolar outputs are electrically isolated from the line power supplying the board in order to avoid problems incurred by ground loops and voltage spikes. Electrical isolation is achieved with use of the NMXS0505UC isolated DC-to-DC converter made by Murata and the IL711 optocoupler manufactured by Nonvolatile Electronics (NVE).

The NMXS0505UC divides the 5 V input supply voltage into two separate power/ground planes: (1) tied to line power or that of the electrical circuit within a building and (2) one apart from line power, having no direct electrical connection to the 5 V input supply. Once sufficiently isolated, the BiSTM current output signals no longer share a common reference with any other electrical equipment which may be attached to a test animal such as neural recording systems, thereby eliminating the formation of potential ground loops between various devices and further eliminating the risk of physically harming the animal or incurring distortions associated with electrical artifacts during recordings. In addition to the electrical isolation through the BT-BiSTM power supply circuit provided by the NMXS0505UC, the Murata IL711 optocouplers further isolate the 30 digital control signals entering the 68-pin Dsub connector at the base of the circuit board. Thus, all sources of electricity attached to the board, both for power and control, are isolated from the outputs of the BiSTM chip.

The BiSTM chip has at each of its 8 bipolar outputs a constant compliance voltage of 5 V when actively generating a signal or when at rest with no current flow. Depending upon the application, the 5 V found at each of the 8 bipolar outputs of the chip may be either disconnected or left attached from the output connectors located on the top of the board when in rest, and of course, attached when active. Achieved by passing each of the 8 BiSTM outputs through an electronically controlled single-pole-single-throw (SPST) switch, the 5 V compliance voltage may be applied or removed from the implanted electrode array as needed.

[3] http://sine.ni.com/nips/cds/view/p/lang/en/nid/13505.

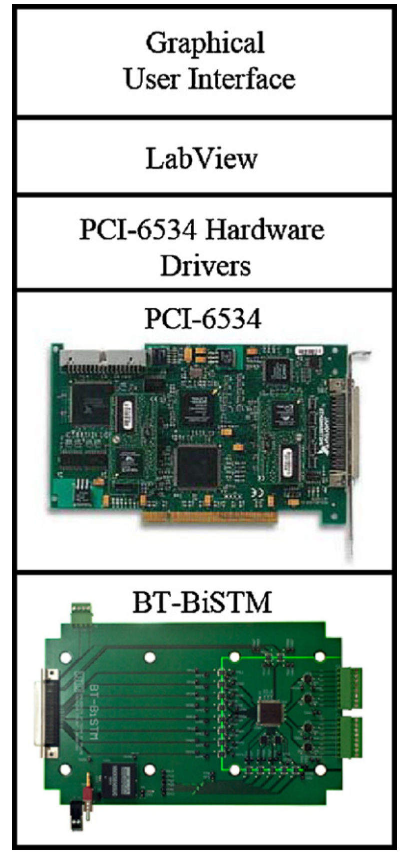

Fig. 2. BT-BiSTM stimulation platform control stack.

Also available on the BT-BiSTM board are 2 spare digital control signals. Equipped with 32 high-speed digital output signals, two of the PCI-6534 outputs are unused and routed out to the left-most connector of the board where it is available to users if needed.

4. PCI-6534 control interface

Attached through a cable to the 68-pin Dsub connector located at the base of the BT-BiSTM board is the PCI-6534 digital output card shown in Fig. 2. The PCI-6534 is equipped with 32 high speed digital output signals which serve as the input control signals to the BT-BiSTM board (though only 30 are needed) where each channel has a maximum data rate of 20 Mbits/s. To ensure accurate generation of waveform stimuli, all the necessary control signals for a given set of stimulation parameters are first stored onto the PCI-6534 64 MBytes of onboard memory and then transmitted to the BT-BiSTM in a repeated pattern at a data rate based upon the onboard 20 MHz clock. Doing so guarantees that the desired timing parameters such as pulse width or inter-stimulus interval are maintained at the outputs with high precision.

A LabVIEW GUI has been created to simplify the task of controlling the BT-BiSTM board by allowing the user to simply specify a set of desired stimulation parameters, without concern of how the underlying 30 digital control signals function to control the board. Essentially, the user is only required to learn how to use the LabVIEW GUI with only minimal knowledge of how the lower levels of the control stack operate (Fig. 2). Fig. 3 shows a snapshot of the LabVIEW GUI. As can be seen, the user can easily change the stimulation rate, the pulse width, pulse amplitude and can also select individual channels to be stimulated simultaneously, interleaved or combinations thereof.

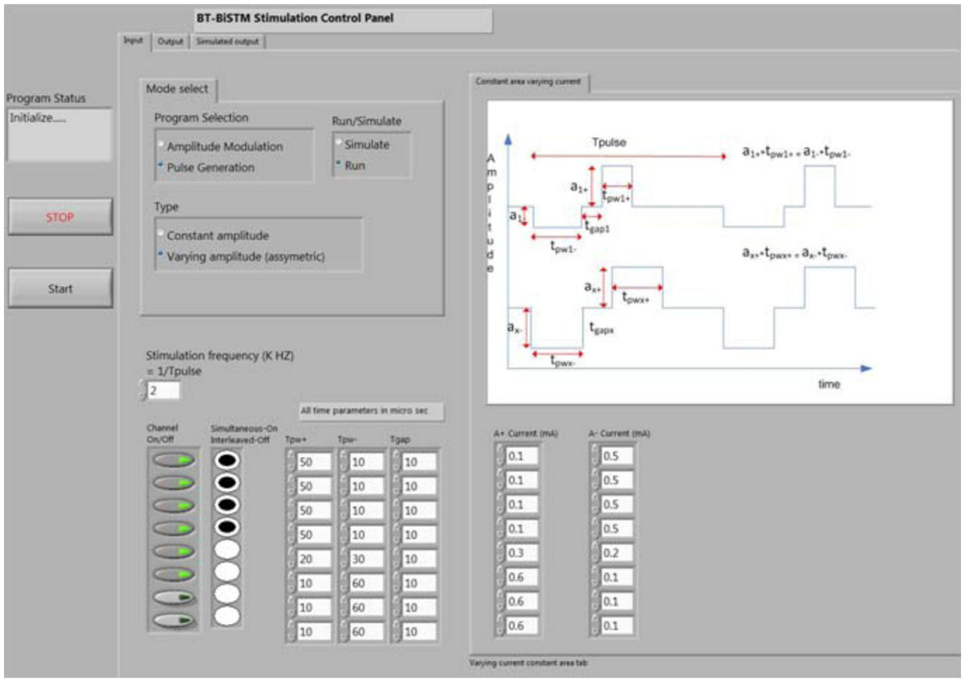

Fig. 3. LabVIEW GUI used for specifying and changing stimulation parameters.

The PCI 6534 card can be controlled using the NI-DAQmx driver. This driver provides a number of optimized application programming interfaces (APIs) for configuring the channels, setting up the buffers and performing acquisition. In the Windows environment, these APIs can be called from LabVIEW or .NET languages such as Microsoft Visual C or Microsoft Visual Basic. The developed GUI in LabVIEW uses the subVIs provided by the DAQmx toolbox to configure the 32bit digital channels and to set a data transfer rate based upon the board's internal clock. DAQmx can also be used to initialize the buffers within the onboard memory which helps to achieve a higher output rate independent of the PCI bus bandwidth.

As shown in Figs. 4 and 5, the LabVIEW GUI also incorporates a simulator with which the output waveforms can be viewed for a given set of desired waveform parameters without actually transmitting the control signals. With the BT-BiSTM board offline, the simulator can be used to set the appropriate parameters and to also help users visualize the output waveforms prior to executing their experiments.

Also shown in Fig. 6 is an additional debugging tool built within the LabVIEW GUI which generates the sequence of 32 bit control words associated with the 32 digital control signals driving the BT-BiSTM board. This table of control words can also be used as a cross reference for any other programming language method chosen to control the board.

Additionally, if needed, users are able to create custom applications that control the BT-BiSTM board in programming environments other than LabVIEW. By using the C/C++ library of hardware drivers provided with the PCI-6534 and the BiSTM *User's Guide*, which describes in full detail the function of each of the chip's 30 control signals (available upon request), users may create cus-

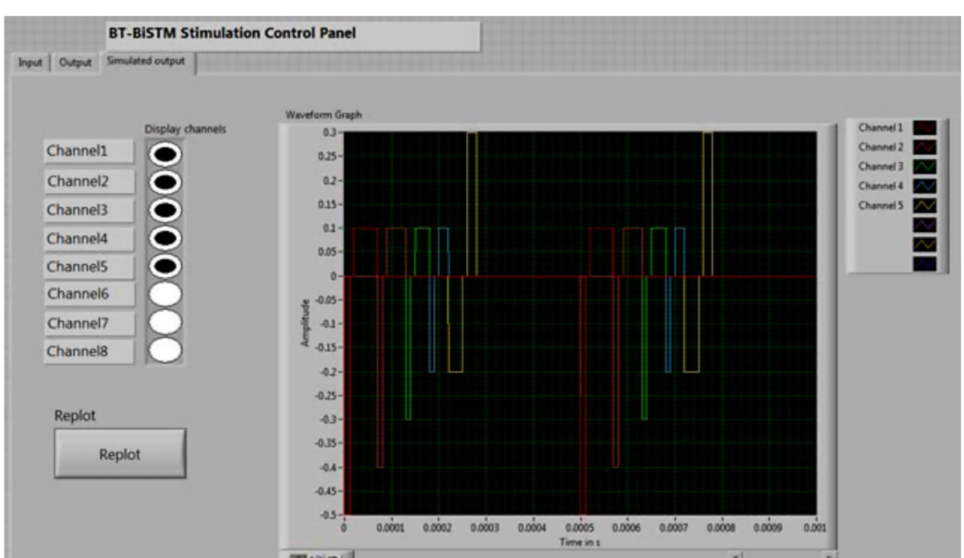

Fig. 4. LabVIEW GUI screen shot of simulated output for varying current interleaved stimulation.

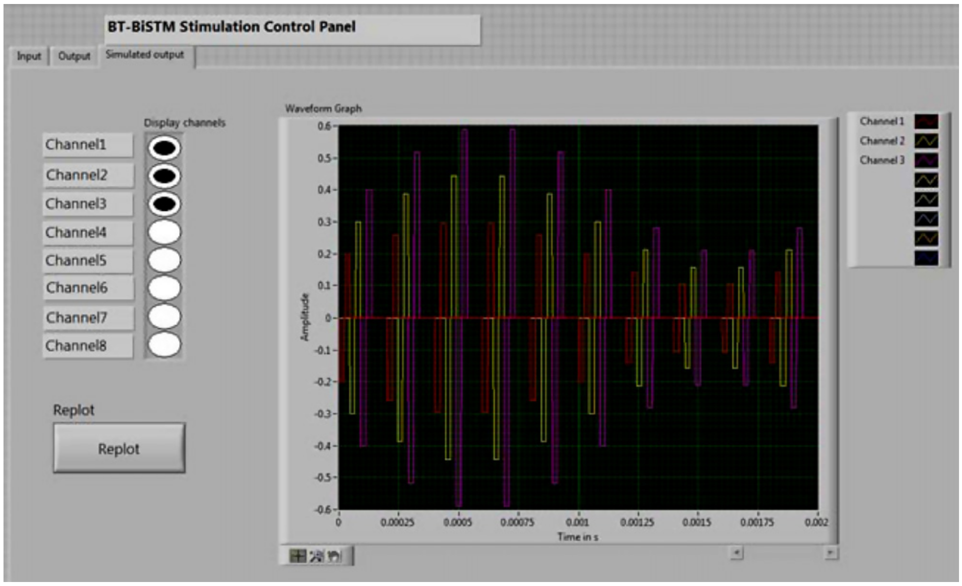

Fig. 5. LabVIEW GUI screen shot of simulated output for interleaved AM stimulation.

tom applications to control the board in order to better integrate it into their existing test setups.

For users who prefer Matlab, .m file scripts that control the PCI-6534 are also available. These scripts like those of the LabVIEW GUI eliminate the complexities associated with programming the PCI-6534 in order to generate a specified set of current stimuli. Users are simply required to specify stimulation parameters familiar to them such as current amplitude, pulse width, interphase gap, pulse rate, etc. The .m file script then automatically translates these parameters into the appropriate set of digital control signals generated by the PCI-6534 that in turn drives the current outputs of the BT-BiSTM board. The PCI-6534 Matlab scripts can also be easily integrated into existing test control software also written in the Matlab programming environment. Therefore, coordination of a whole host of devices needed in an experiment, ranging from recording instruments and scopes to current stimulators and audio outputs can be realized in a single programming environment.

Lastly, by taking advantage of the BT-BiSTM open control interface, digital output boards other than the PCI-6534 may also be used to control the BT-BiSTM given that these boards have a minimum of 30 outputs, each 5 V TTL compatible, and are capable of meeting the timing requirements of a particular application.

5. Evaluation

The following sample waveforms are provided as examples of the BT-BiSTM capabilities. Beginning with the two simultaneous charge-balanced bipolar signals shown in Fig. 7 where channels 1 and 2 generate identical signals locked in phase with pulse widths of 50 μs, interphase gaps of 10 μs, inter-stimulus interval of 140 μs,

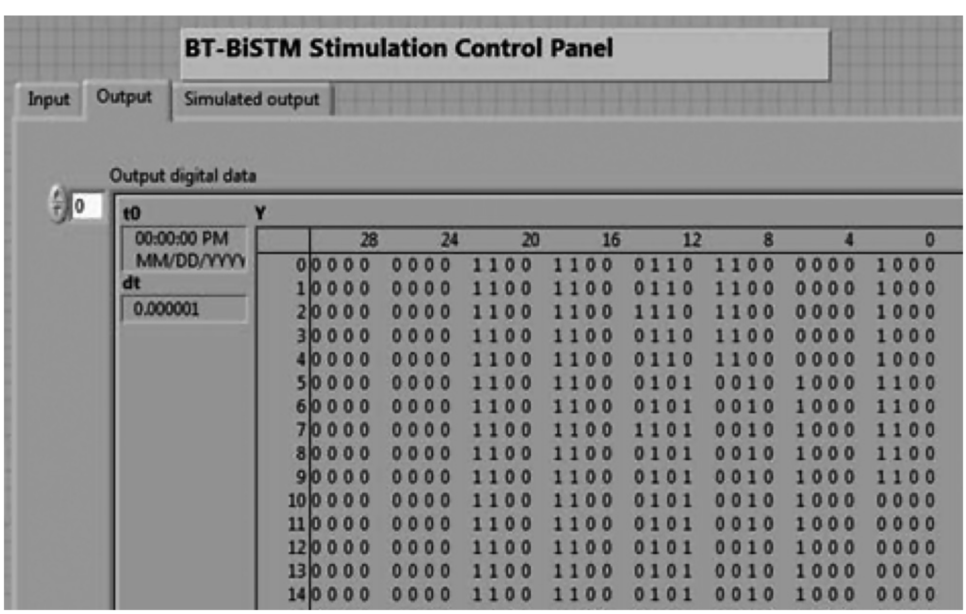

Fig. 6. LabVIEW GUI screen shot of simulated output for interleaved AM stimulation.

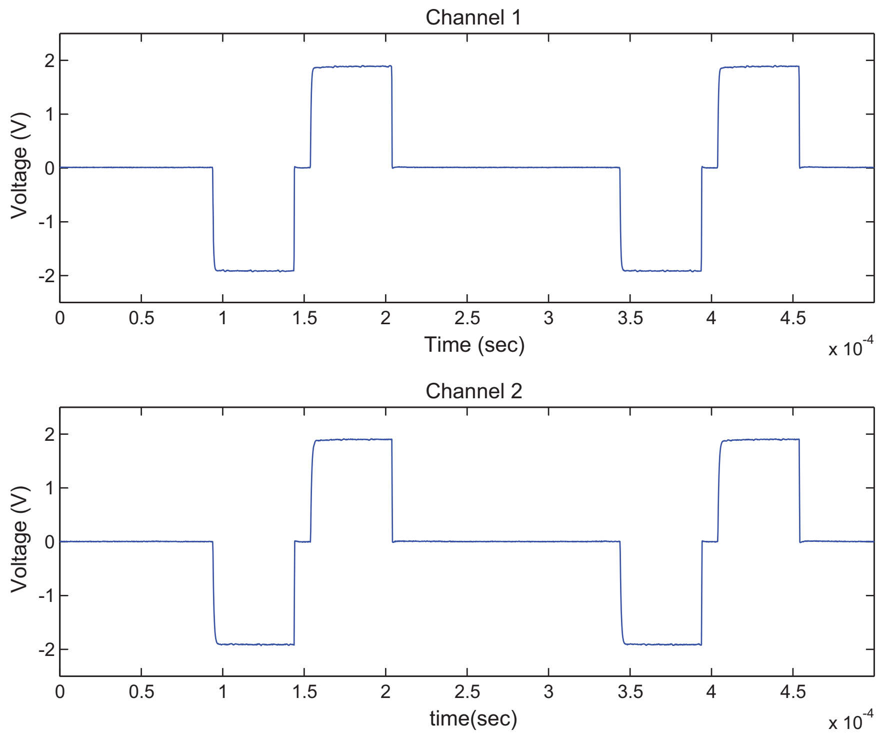

Fig. 7. Pulsatile simultaneous stimulation measured over 2 kΩ loads.

and current amplitudes of 965 μA. Measurements are taken over 2 kΩ loads. Note, that although only 2 channels are displayed in this example, the BT-BiSTM is capable of generating simultaneous signals over all 8 channels.

To further demonstrate the BT-BiSTM's ability to generate nearly any arbitrary waveform pattern, two additional phase locked signals are shown in Fig. 8 where channel 2 generates a *symmetric* bipolar signal and channel 1 generates an *asymmetric* bipolar signal. For both outputs, the total charge[4] over the cathodic pulses are equal in voltage (current) and pulse width where 765 μA is applied to both channels for 10 μs. However, for the anodic pulses, channel 2 (as it does for its cathodic pulses) outputs 765 μA for 10 μs whereas channel 1 outputs 191 μA for 40 μs. Measurements are taken over 2 kΩ loads.

As a demonstration of the BT-BiSTM's ability to also generate amplitude modulated (AM) signals, displayed in Fig. 9 are three sets of phase locked AM signals for both channels 1 and 2. The AM current signal, $y(t)$, is defined as

$$y(t) = [A + M\cos(2\pi ft)]p(t) \quad (1)$$

$$y(t) = A[1 + \mu\cos(2\pi ft)]p(t) \quad (2)$$

where the modulation index, μ, is defined as

$$\mu = \frac{M}{A}. \quad (3)$$

From the above, the low-frequency envelope of the AM signal, $M\cos(2\pi ft)$, is offset by a DC signal, A, and modulated by a periodic bipolar carrier signal, $p(t)$ [14]. The modulation index of the signal, μ, defines the ratio of the amplitude of the low frequency envelope, M, to the DC offset, A.

In Fig. 9, the frequency, f, of the low-frequency envelope for channels 1 and 2 is equal to 400 Hz and 200 Hz, respectively. For both channels, the bipolar carrier signal has a pulse width of 50 μs, interphase gap of 10 μs, and inter-stimulus interval of 140μs. Plots are shown for modulation depths of 25%, 50% and 100%.

Presently in use by members of the Laboratory of Auditory Neurophysiology at Johns Hopkins University[5] as part of a research project that seeks to develop a non-human primate electrophysiological model for cochlear implant research [15,16], the BT-BiSTM board has been used in chronic cochlear implant neurophysiological experiments in which an array of cochlear implant electrodes surgically implanted within the cochlea of awake marmosets is electronically stimulated by current signals generated by the BT-BiSTM board in order to activate the auditory pathways in primates.

Sample waveforms taken from these chronic implant experiments are shown in Fig. 10. Unlike the waveforms shown in Figs. 7 and 8 which were applied across purely resistive loads, a noticeable transient response is present on both the measured cathodic and anodic pulses due to the capacitive effects inherent to cochlea physiology. Both waveforms output 100 μA and have equal pulse widths of 100 μs and varying interphase gaps of 50 μs and 8 μs.

6. Portable stimulator: SDIO-BiSTM system overview

Presently under development is a portable adaptation of the bipolar stimulation platform. Referred to as the SDIO-BiSTM, the portable stimulation platform is based upon the open-interface cochlear implant research platform reported in [11]. Similar to the open-interface research platform, the SDIO-BiSTM exploits the mobile processing capabilities of a personal digital assistant (PDA) and combines it with a custom made interface card that communicates with the PDA through a secure digital IO (SDIO) slot, hence the SDIO prefix in the name of the portable stimulation platform. In

[4] Here, charge is defined as the time integral of the absolute value of the pulse voltage while active.

[5] http://web1.johnshopkins.edu/xwang/HomePage.html.

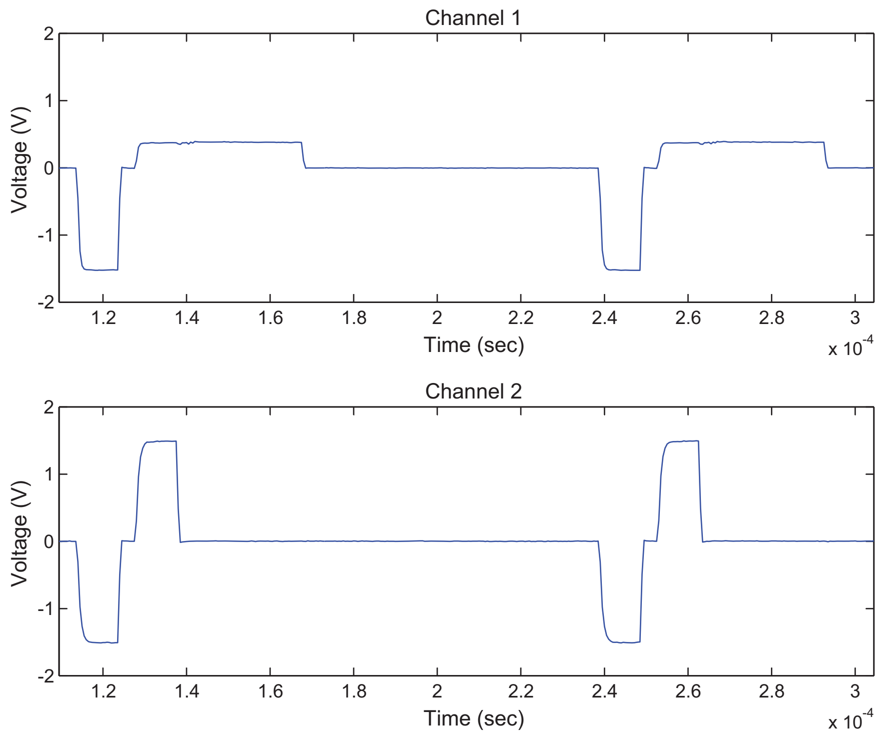

Fig. 8. Symmetric/asymmetric simultaneous stimulation measured over 2 kΩ loads.

the case of the SDIO-BiSTM, the interface board consists primarily of the BiSTM chip.

The SDIO-BiSTM is comprised of a main board and a daughter board as shown in Figs. 11 and 12, respectively. Listed below are the main circuit elements of the main board and their respective functions.

- *Xilinx Spartan FPGA*: Accepts desired output waveform parameters from a GUI application running on the PDA and controls the BiSTM chip accordingly in a way similar to that of the PCI-6534 in the case of the BT-BiSTM platform.
- *Arasan SDIO interface controller*: Controls communication between the PDA and FPGA.

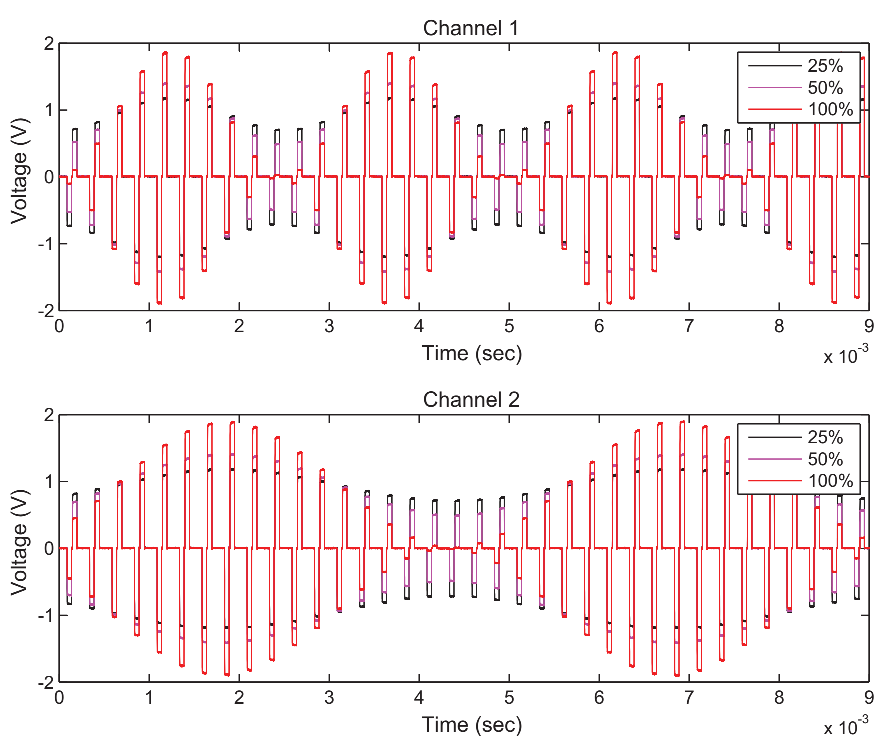

Fig. 9. Simultaneous amplitude modulation stimulation measured over 2 kΩ loads.

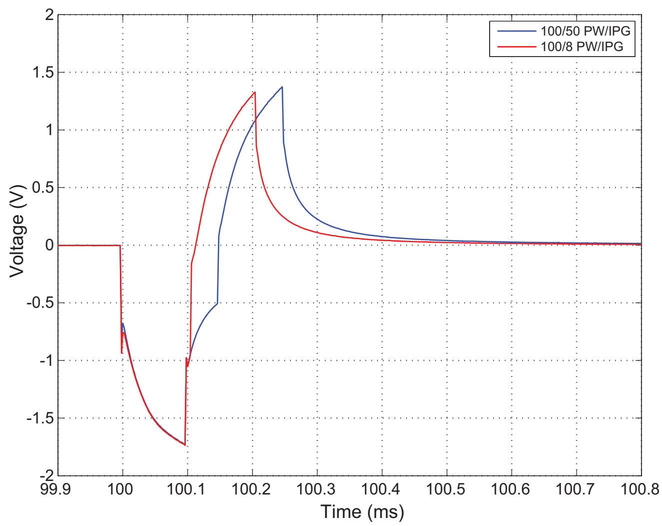

Fig. 10. Stimulation of chronically implanted cochlear implant electrode array at 100 μA.

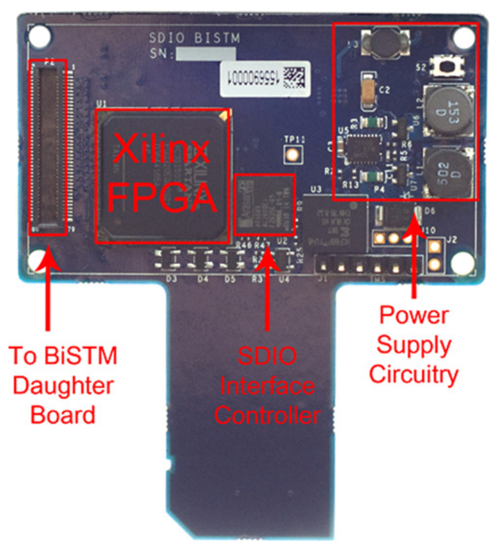

Fig. 11. SDIO-BiSTM main board prototype.

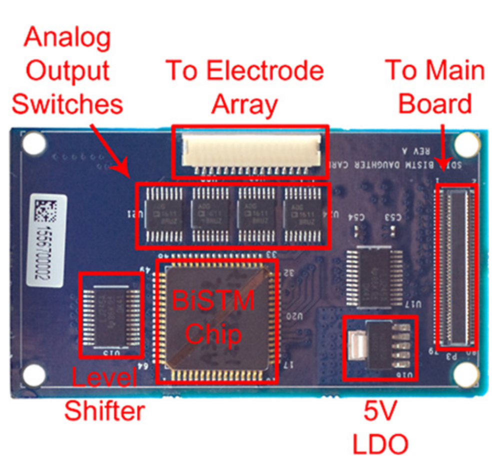

Fig. 12. SDIO-BiSTM daughter board prototype.

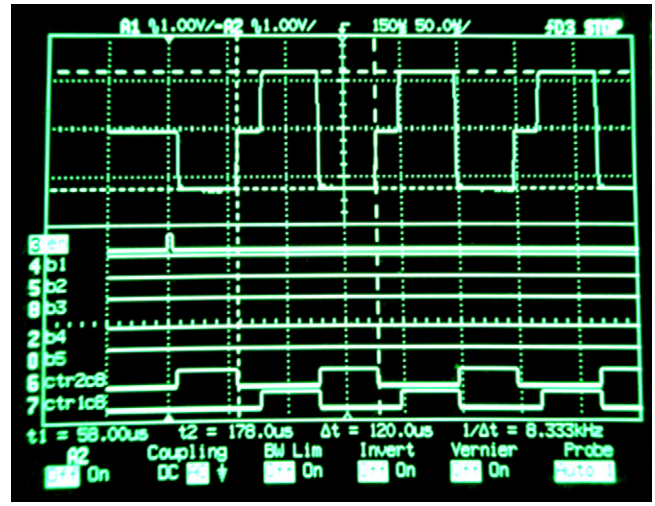

Fig. 13. SDIO-BiSTM sample output waveform generation.

- *80-pin board-to-board connector*: Routes various control/power signals to the BiSTM chip located on the daughter board.
- *Miscellaneous power circuitry*: Converts 6 V of battery supply power to the various voltage levels needed to power the board.

The daughter board circuit elements and respective functions are as follows:

- *BiSTM chip*: 8 channel configurable bipolar current source.
- *Analog output switches*: Disconnects the 5 V compliance voltage from the test subject when the BiSTM chip is in reset.
- *5 V low drop-out regulator (LDO)*: Converts/regulates 6 V battery power down to the 5 V BiSTM chip supply power.
- *Voltage level shifters*: Translates the FPGA's 3.3 V digital signal to the required 5 V level of the BiSTM chip.
- *16-pin output connector*: Routes the 8 bipolar signals to the implanted cochlear electrode array.

The SDIO-BiSTM board design was divided into two separate boards in order to minimize the overall size of the board. Combining both boards into a single board would have been many times larger and impractical. Also, in anticipation of the eventual release of the monopolar stimulator chip, the modular design of the daughter board system requires the design of just one other daughter board that can also be controlled by the same SDIO-BiSTM main board. This eliminates the need to redesign two additional circuit boards.

Also note that since the SDIO-BiSTM board operates on battery power to facilitate mobility; the 8 BiSTM bipolar outputs are electrically isolated from any other electrical devices that may be attached to a test animal.

Although still under development, preliminary tests of SDIO-BiSTM platform have been conducted in which biphasic pulses were generated to test the operation of the SDIO-BiSTM daughter card. The results of those tests are presented in this section. The sample output waveform generation procedure for channel 8 is summarized below in steps 1 through 8.[6]

1. Set signal SW8 to 0 to close the analog output switch for channel 8

[6] The control logic for the stimulation was coded in Verilog and synthesized for the XC3S1500 FPGA under Xilinx ISE 11.1 Project Navigator.

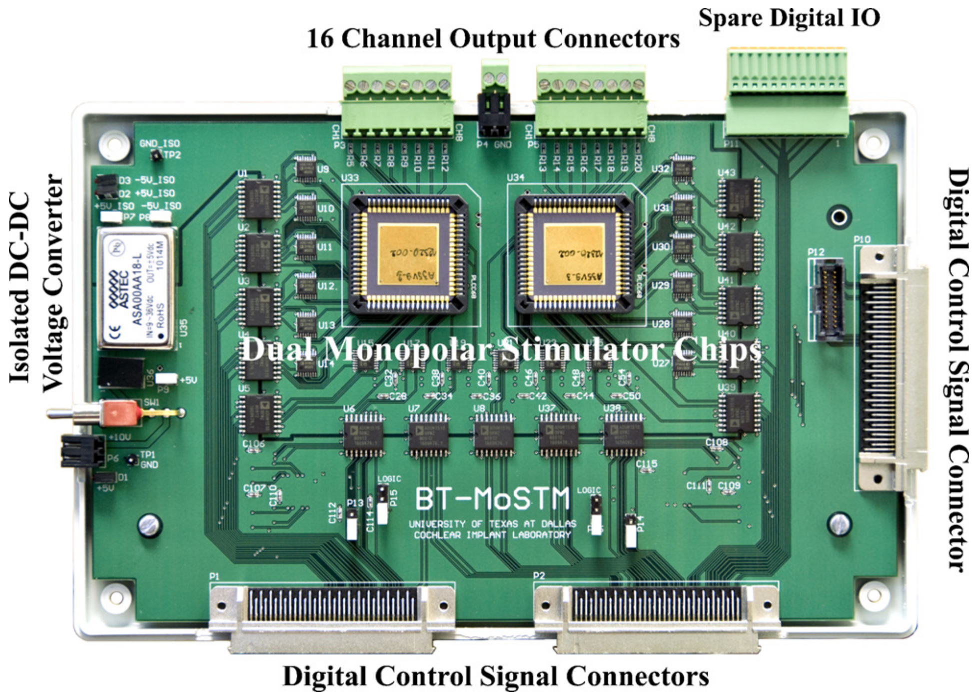

Fig. 14. SDIO-MoSTM benchtop stimulator.

2. Set the channel address bits ADDR3, ADDR2, ADDR1 to 111 to select channel 8
3. Set signal EN to 0
4. Set signals b1.b9 to define the desired output current level (set to 011111111 for 0.5 mA)
5. Wait 4 μs
6. Set EN to 1 in order to latch in the desired output current level for channel 8
7. Wait 4 μs
8. Set EN to 0
9. Continuously stimulate channel 8 as follows to generate a charge balanced biphasic pulse train at a rate of 8.3 kHz (time period = 120 μs):
   - CH8− → CH8+ for 50 μs to generate the negative phase. The right arrow indicates the current direction, i.e. from the negative to the positive terminal
   - CH8+ = CH8− = 0 for 20 μs to generate the interphase gap
   - CH8+ → CH8− for 50 μs to generate the positive phase. The right arrow indicates the current direction, i.e. from the positive to the negative terminal.

Fig. 13 shows the BiSTM chip control signals in the lower half of the oscilloscope display. Shown in the upper half of the oscilloscope display are the resulting biphasic pulses measured across the output of channel 8 (measurements taken across a 2.2 kΩ resistor). The current amplitude level bits, b1–b5, are also shown. The measurement was taken using two analog channel probes with one of the probes connected between the positive output terminal and ground and the second probe connected between the negative output and ground and taking the arithmetic difference between the two analog channels.

As can be seen, to generate the negative half of the pulse the CTRL1_CH8 control signal is set to 0 and the CTLR2_CH8 control signal is varied to either 0 or 1 in a pattern that resembles the desired output waveform where the output current is turned on when CTRL2_CH8 is in the ON state and turned off when CTRL2_CH8 is in the OFF state. To generate the positive half of the pulse the roles of CTRL1_CH8 and CTRL2_CH8 as described above is reversed. To generate the interphase gap both CTRL1_CH8 and CTRL2_CH8 are set to 0.

Because the SDIO-BiSTM was designed as a portable stimulator for chronic animal studies, the physical dimensions and weight of the overall system including the PDA, the current stimulation board, and battery pack have been minimized to the extent possible. Depending upon the animal to which the device will be used, the following physical specifications should facilitate portability.

- Dimensions of overall system: 6 × 3 × 0.6 in.
- Weight of overall system: 0.66 pounds

Furthermore, in order to minimize the frequency at which batteries must me replaced and recharged, preliminary tests have shown that the battery life of the overall system is approximately 4 h. Multiple spare batteries are available with the system. Depleted batteries can quickly be replaced with those fully charged and allowed to be recharged amongst a set of other spare batteries.

7. Monopolar bench-top stimulator

Currently under development is a bench-top version of our next generation monopolar stimulator, referred to as the BT-MoSTM. The new stimulator board is designed around two dual monopolar stimulator chips [17] that feature 8 monopolar channels per chip for a total of 16 charge balanced monopolar channels, each sharing a common reference and capable of sourcing a maximum of 1 mA. The 16 channels can be independently controlled, each varying in stimulation parameters including current amplitude, pulse width, interphase gap, pulse rate, etc. Much like the BT-BiSTM board, the new monopolar version of the board is also capable of generating a vast array of varying pulsitile and analogue-like stimulation patterns in either simultaneous or interleaved modes or combinations thereof. A user-friendly and intuitive LabVIEW GUI as well as Matlab scripts will also be provided with the BT-MoSTM board to simplify its use and control in order to minimize the time required by researchers to become familiar with the system. An image of the BT-MoSTM board is shown in Fig. 14.

8. Availability of developed animal stimulators

As to date, fully operational units of the BT-BiSTM benchtop bipolar stimulator are available for interested parties. Installation instructions and remote technical support are also available as part of the system. Once fully developed, the portable version of the bipolar stimulator, SDIO-BiSTM, will also be made available to researchers interested in its use; as will be the BT-MoSTM benchtop monopolar stimulator. For more information on acquiring any of these devices, please contact the project's principal investigator Dr. Philipos Loizou[7] at the Department of Electrical Engineering at the University of Texas at Dallas.

9. Conclusions

Presented in this paper is the design of a flexible and portable bipolar current stimulation system for chronic cochlear implant studies on animals. The proposed stimulation platform provides researchers with up to 8 independently controlled and charge balanced bipolar current outputs within a compact and light weight form factor that is powered by rechargeable batteries – well suited for portable applications.

Also presented in this paper, is the design of a bench top version of the bipolar current stimulation system for anaesthetized animal studies. Built around the same bipolar current stimulator chip used in the portable system, the bench top stimulator is also equipped with 8 independently controlled and charge balanced bipolar current outputs that – with the aid of the system's user friendly interface – can easily be programmed to generate a wide array of stimulation waveforms varying in parameters including pulse width, interphase gap, pulse rate, etc.

A preview of the monopolar version of the current stimulation system is also presented. Presently under development, the new system will deliver many of the same capabilities offered by its predecessors in a monopolar configuration and will support a higher channel count of up to 16 independently controlled current sources.

Acknowledgments

This work has been funded by contract N01-DC-6-0002 from NIDCD/NIH. We would like to thank Dr. Xiaoqin Wang and Luke Johnson of the Laboratory of Auditory Neurophysiology at Johns Hopkins University for providing us with the sample waveforms collected during their experiments for use in this paper.

References

[1] J. Laniau, B. Boets, M. Moonen, A. van Wieringen, J. Wouters, A flexible auditory research platform using acoustic or electric stimuli for adults and young children, Journal of Neuroscience Methods 142 (1) (2005) 131–136.
[2] L.V. Immerseel, S. Peeters, P. Dykmans, F. Vanpoucke, P. Bracke, Spaide: a real-time research platform for the Clarion CII/90K cochlear implant, EURASIP Journal on Applied Signal Processing 2005 (18) (2005) 3060–3068.
[3] K. Kasturi, P.C. Loizou, Use of s-shaped input-output functions for noise suppression in cochlear implants, Ear and Hearing 28 (3) (2007) 402–411.
[4] D.B. Grayden, S. Tari, R.D. Hollow, Differential-rate sound processing for cochlear implants, in: Proceedings of the Australian International Conference on Speech Science & Technology, New Zealand, 2006, pp. 323–328.
[5] D.D. Wong, K.A. Gordon, Beamformer suppression of cochlear implant artifacts in an electroencephalography dataset, IEEE Transactions on Biomedical Engineering 56 (12) (2009) 2851–2857.
[6] K. Kokkinakis, P.C. Loizou, Multi-microphone adaptive noise reduction strategies for coordinated stimulation in bilateral cochlear implant devices, Journal of the Acoustical Society of America 127 (5) (2010) 3136–3144.
[7] M. Ghovanloo, K. Najafi, A compact large voltage-compliance high output-impedance programmable current source for implantable microstimulators, IEEE Transactions on Biomedical Engineering 52 (1) (2005) 97–105.
[8] J. Sit, R. Sarpeshkar, A low-power blocking-capacitor-free charge-balanced electrode-stimulator chip with less than 6 nA DC error for 1-mA full-scale stimulation, IEEE Transactions on Biomedical Circuits and Systems 1 (3) (2007) 172–183.
[9] J. Georgiou, C. Toumazou, A 126-μW cochlear chip for a totally implantable system, IEEE Journal of Sold-State Circuits 40 (2) (2005) 430–443.
[10] P.T. Bhatti, K.D. Wise, A 32-site 4-channel high-density electrode array for a cochlear prosthesis, IEEE Journal of Sold-State Circuits 41 (12) (2006) 2965–2973.
[11] A. Lobo, P. Loizou, N. Kehtarnavaz, M. Torlak, H. Lee, A. Sharma, P. Gilley, V. Peddigari, L. Ramanna, A PDA-based research platform for cochlear implants, in: Proceedings of the 3rd International EMBS Conference on Neuroengineering, Hawaii, 2007, pp. 28–31.
[12] G. Stickney, P. Loizou, L. Mishra, P. Assmann, R. Shannon, J. Opie, Effects of electrode design and configuration on channel interactions, Hearing Research 211 (1–2) (2006) 33–45.
[13] S. Guo, H. Lee, P. Loizou, A 9-bit configureable current source with enhanced output resistance for cochlear stimulators, in: Proceedings of the IEEE Custom Integrated Circuits Conference, San Jose, CA, 2008, pp. 511–514.
[14] S.S. Haykin, Communication Systems, Wiley, New York, 2001.
[15] L.A. Johnson, C.C. Della Santina, X. Wang, Developing a new non-human primate model for cochlear implant research, in: Conference on Implantable Auditory Prosthesis (CIAP), Lake Tahoe, CA, 2009 (poster).
[16] L.A. Johnson, C.C. Della Santina, X. Wang, Developing a new vocalizing non-human primate model for cochlear implant research, in: Society of Neuroscience, Chicago, IL, 2009 (poster).
[17] S. Guo, H. Lee, Biphasic-current-pulse self-calibration techniques for monopolar current stimulation, in: Proceedings of the IEEE Biomedical Circuits and Systems Conference, Beijing, China, 2009, pp. 61–64.

[7] Email: loizou@utdallas.edu.