DESIGN AND EVALUATION OF A PDA-BASED RESEARCH PLATFORM FOR COCHLEAR IMPLANT

by

Hussnain Ali

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I dedicate this work to hearing impaired people around the world,

to those whose ear shout endless silence to their brains,

to the hope that one day cochlear implants will penetrate the inhuman silence of all deaf people,

to all children and adults who have benefitted from implants and

to the doctors and researchers bringing sounds and communication to their lives

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This thesis presents the design, development, features and clinical evaluation of a PDA-based research platform for cochlear implant research. The research platform as a whole comprises of a portable processor for implementing and evaluating novel speech processing algorithms, a stimulator unit for electrical stimulation, and a recording unit for collecting evoked potentials. The focus of the presented work is towards developing a software driven solution for researchers working in this domain and provide them with a comprehensive infrastructure and versatile set of tools to design and conduct simple to complex experiments for cochlear studies with great ease and flexibility. Design of the platform for real-time and offline stimulation is discussed for electric-only and electric plus acoustic stimulation followed by evaluation with CI users for speech intelligibility task in quiet and different noise conditions. The results are comparable with users' clinical processor and very promising for undertaking long-term chronic studies.

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CHAPTER 1

INTRODUCTION

Cochlear implants (CI) serve as a benchmark technology in neural prosthesis for their high success rate in restoring hearing to the deaf and their growing and widespread use. According to the U.S. Food and Drug Administration (FDA), as of December 2010, approximately 219,000 people worldwide have received implants (National Institute on Deafness and Other Communication Disorders 2011). In the United States, roughly 42,600 adults and 28,400 children have received them. Comparison of these statistics to the year 2005 when there were about 110,000 implant recipients (National Institute on Deafness and Other Communication Disorders 2005) and the year 1995 when there were only 12,000 (National Institutes of Health 1995) implant recipients, indicates the growing demand and satisfaction with the implant performance. This growth is driven by extensive research both in academia and industry in developing better sound processing algorithms for sound representation, and novel design of stimulators and electrode arrays for improved stimulation patterns.

This research is largely dependent upon having access to a research platform which could be used to design new experiments and evaluate user performance over time. While most implant manufacturers provide research speech processors for use in human subjects that allow researchers to develop and test new signal processing algorithms, most labs are unable to use them due to limited technical resources or due to constrained framework of the interface provided by the manufacturer. These limitations include flexibility, portability, wearability, easy of programmability, long-term evaluation and features to design intricate experiments. One of the important factors which hinder their use for speech processing research is that a skilled programmer is required to implement the algorithms in high level or low-level language (Ahmad, et al. 2009). To address these needs (Lobo, et al. 2007) presented a highly versatile solution based on a portable computing platform, such as a smartphone or a PDA, which in addition to portability and wearability could provide all the computing horse-power required to drive complex algorithms/tasks, thus allowing long-term chronic evaluation with both human and animal subjects not confined only to laboratory environment but also in practical day to day environments and activities.

The PDA platform has since undergone numerous hardware and software upgrades and this proof-of-concept has emerged as a successful tool in the research arena with many researchers utilizing it to develop new algorithms and to establish its use in clinical studies. In (Lobo, et al. 2007), our group demonstrated real-time implementation of a 16-channel noise-band vocoder algorithm in C and LabVIEW, which is a similar algorithm used in commercially available implants. In the same paper recording of EEG signals on the PDA acquired through a compact flash data acquisition card was also reported. (Peddigari, Kehtarnavaz and Loizou 2007) presented real-time implementation of cochlear implant signal processing system on the PDA platform in LabVIEW using dynamic link libraries. A recursive real-time DFT-based ACE implementation for high analysis rate was reported in (Gopalakrishna, Kehtarnavaz and Loizou 2009). (Gopalakrishna, Kehtarnavaz and Loizou 2010) presented real-time implementation of wavelet-based Advanced Combination Encoder and a novel wavelet packet based speech coding strategy on the PDA platform in (Gopalakrishna, Kehtarnavaz and Loizou 2010). Successful implementation of these algorithms on the PDA platform justifies its potential for implementing wide variety of algorithms with varying complexity for real-time applications.

The work presented in this thesis focuses on extending the ground work laid by our research group and use the platform to implement and streamline speech processing algorithms for real-time and offline stimulation suitable for testing with human subjects. The functionality of the platform is extended to bimodal (combining electrical stimulation via the implant with acoustic stimulation via hearing aids) capabilities. In addition to this, design of various software applications and experiments for clinical studies with the platform is presented. Most importantly, evaluation of the platform with CI users in speech intelligibility tasks in quiet and different noise conditions is reported.

This thesis is organized as follows. Chapter 2 presents a literature review of existing research interfaces, their capabilities and limitations. Chapter 3 describes the platform in detail, starting from hardware components to software architecture in real-time and offline modes. This is followed by architecture description of the platform in bimodal mode. Chapter 4 presents the evaluation of the platform with unilateral, bilateral and bimodal CI subjects. The results are compared with the clinical processor. Finally Chapter 5 presents conclusion of the research work.

CHAPTER 2

BACKGROUND

The success and growth of cochlear implant use is largely driven by extensive research both in academia and industry. Experimental research platforms for cochlear implant research have allowed researchers from different disciplines to design various experiments by manipulating stimulation waveforms and parameters by using existing or designing novel sound processing algorithms and adjustments to users MAPs. Researchers have used these interfaces to gain a better understanding of the behavior of the cochlea in response to electrical stimulation, which has in turn led to the development of new speech-processing strategies that have resulted in improvements in speech recognition (Stohl, Throckmorton and Collins 2008). These interfaces have been essential to the improvement of speech-processing strategies, but researchers were constrained by the framework of the interface provided by the manufacturer e.g. ability to modify programs and stimulation parameters within speech processor, ability to control stimuli at individual electrodes and ease of programmability to name a few. Following section provides a brief overview of the past and existing research interfaces and tools along with their features and limitations.

BTNI was one of the earliest generations of research interfaces which allowed undertaking simple tasks such as experimental control of electrode number, pulse amplitude, duration, rate and pulse phase duration in patients implanted with the Nucleus device from Cochlear Corporation (Shannon, et al. 1990). It also provided some assembly routines to help implement speech processing strategies; however the design of BTNI was more suitable for performing psychophysics experiments within a limited framework. Another major limitation of the BTNI was that it only allowed non-simultaneous stimulation (which could be attributed to the implanted receiver as well) hence making it unsuitable to study channel interaction and perform similar experiments.

Nucleus cochlear Implant Communicator (NIC) is a set of software modules and libraries which allows communicating with two Nucleus (CI22 and CI24) implants (Goorevich, Irwin and Sawnson 2002). The stimuli can be created in any software on an IBM-compatible personal computer. NIC interface helps to encode these stimuli to sequences of frames and send instructions to the SPrint processor which acts as a hardware interface between the PC and the implant and controls transmission of radio frequency pulses to subject's implanted receiver/stimulator unit. In this way, NIC provides a software-based interface to communicate with the implant and lets researchers to design stimuli of their choice without having to worry about communication protocols. Nucleus MATLAB Toolbox (NMT) is a set of MATLAB routines/scripts provided by Cochlear Corporation of sample speech processing codes, such as ACE speech processing strategy, to help researchers write/modify speech processing strategies in MATLAB environment while employing NIC backbone to communicate with the Nucleus devices. While NIC is useful for experiments involving speech, the main drawback is that it is confined to the laboratory use because it involves a PC to create stimuli (Stohl, Throckmorton and Collins 2008). NIC-NMT cannot be used in real-time mode like a commercial processor which processes acoustic signal and provides electrical stimulation in real-time. Another limitation for stimuli longer than 15,000 pulses with NMT is their restriction only to monopolar mode with constant pulse rate and constant pulse shape (Laneau, et al. 2005).

SPEAR3 research platform from Hearworks Pty Ltd. is a portable, wearable research sound processor for electrical and acoustic research which is compatible with Cochlear Corporation's Nucleus CI22 and CI24 implants (Stohl, Throckmorton and Collins 2008). Also known as SHARP/SPEAR programming system (SPS), it allows allows either a SHARP or SPEAR3 sound processor to be connected directly to a personal computer for programming (HearWorks Pty Ltd. 2003), (CRC and HearWorks 2003).

The SPS/SPEAR3 package had two real advantages over previous research interfaces:

1. It provided developers access to the digital signal processor program file, and the ability to upload modified assembly code which could be used to implement original psychophysical experiments and completely new sound processing algorithms.

2. Ability to upload new strategies and conduct take home experiments with the portable processor, rather than being confined to an experimental environment.

Daily use of a new strategy in familiar environments provided an opportunity to observe the possible effects of adaptation on a user's performance for the first time. In addition to access to the programs within the digital signal processor, the SPEAR3 had the ability to drive two Nucleus CI22 or CI24 implants for bilateral stimulation and could also be used in a multi-modal fashion, where an acoustic stimulus is also presented to take advantage of some remaining residual hearing.

Complete SPS/SPEAR3 software package includes a configurable program file for the SPEAR3 and a graphical user interface (GUI) called Seed-Speak that enables researchers to manipulate the parameters of the SPEAK speech-processing strategy. Seed-Speak could also be used to obtain psychophysical data that is typically collected clinically via tasks including estimation, ranking, and loudness balancing. HearWorks also provided software development tools to researchers willing to build a platform with additional features. Thus, it allowed extension of the software and implementation of more complex psychophysics experiments. The main drawback of SPS/SPEAR3 system was that it was difficult to implement new experiments and sound coding strategies as it required programming in assembly language which is a cumbersome task even for experienced programmers. In addition to this, the platform lacked the ability to update stimulus information/parameters in runtime. The example Visual Basic codes and Graphical User Interface (GUI) provided was not compatible with provided assembly code. Therefore, even with interesting features like ability to implement your own strategies/experiments in bilateral and/bimodal mode, SPS/SPEAR3 lacked the flexibility and features which researchers would want in an ideal research interface to develop variety of experiments easily and quickly.

The Clarion Research Interface (CRI) developed in late 1997 and 1998 was a result of a joint collaboration between House Ear Institute and Advanced Bionics Corporation intended for use by technically sophisticated research groups who desired to work with Clarion cochlear implant subjects (Shannon, et al. 1999). The hardware consists of a host PC capable of running Clarion SCLIN software, a DSP Development Board (EVM), and the Clarion Speech Processor (SP)/Headpiece and Implantable Cochlear Stimulator (ICS). The software on the CRI consists of RSP software running on the SP and DSP software running on the EVM. Applications that required custom software on the PC could be interacted with EVM through the Host Port Interface. The Clarion Research Interface allowed:

i. The presentation of preprocessed stimuli, or
ii. Presentation of speech file from a host PC, and
iii. The implementation of real-time speech processor

CRI made offline implementation of complex algorithms significantly easier. Its main strength however was the ability to program it in C language. On the other hand, CRI was not capable of bilateral stimulation and it was also not portable making it unsuitable for experimentation outside the lab environment.

Research Interface Box (RIB) provided by University of Innsbruck worked with the MED-EL cochlear implants. The interface was controlled using a personal computer by a serial communication port (RS-232) (Nie, Barco and Zeng 2006). The computer processes a sound file (.wav) offline and generates a data file that contains all parameters describing the electric stimulus, including the electrode number, current amplitude, pulse duration, inter-pulse interval, and stimulation rate. During the test, the RIB would download the data file, generate its equivalent signal and send this signal to the internal receiver through radio frequency link coil. MATLAB could be used for sound processing. While RIB was a satisfactory platform for offline implementation of complex algorithms, it was not suitable for real-time processing. Also, it is not portable and is not suitable for bilateral studies.

APEX (acronym for computer Application for Psycho-Electrical eXperiments) is a versatile software test platform for auditory behavioral experiments (Laneau, et al. 2005). It provides a generic frame-work for setting up behavioral and psychophysical experiments without any programming by exploiting the fact that most experiments have many parts in common. Strength of APEX lies in its versatility to run both in Windows and Linux environments and its ability to support cochlear implants from Cochlear Corporation and Advanced Bionics. This is done by utilizing NIC interface for Cochlear Corporation devices and Clarion Device interface for Advanced Bionics devices. In conjunction with electric stimuli, APEX supports bimodal capability and can provide both electric and acoustic stimulation. However, APEX is a software framework which employs NIC and Clarion research interface (CRI) as its backbone; thus it inherits the same limitation as those with NIC and Clarion interface such as inability to provide stimulation in real-time mode, portability and wearability for long-term assessment of algorithms.

A brief review of the existing research platforms with their features and limitations was presented above. Limitations of these interfaces can be summarized as:

i. Limited access to the programs within the speech processor,
ii. Ease of programmability, with most platforms require programming in assembly language,
iii. Limited feature space to be able to define stimulation modes, stimulation parameters and/or stimuli pulses of individual electrodes to be able deliver complex stimuli, e.g., in psychophysics experiments.
iv. Limited flexibility to allow quick development and evaluation of new research ideas
v. Portability and wearability for realistic assessment of new algorithms after long-term evaluation.

These limitations served as a rationale for our group to design a research platform which could provide all software flexibility and hardware portability for long-term assessment of novel research ideas in daily lives of cochlear implant users. A portable platform such as a smart-phone or a PDA which could allow implementation of speech processing strategies along with a user interface in the form of a touch-screen which could allow real-time feedback and response from the users. Such a platform would provide control of the speech processing parameters to the users and allow them to tune and optimize their listening experience according to the physical environment they are in.

CHAPTER 3

RESEARCH PLATFORM

Current CI users carry either a body-worn speech processor or a Behind-The-Ear (BTE) processor. The headpiece or BTE contains the microphone and RF transmitter, and is connected to the body-worn processor by a custom cable. Sound is picked up by the microphone and sent to the processor, which processes the signal in a way that mimics the auditory signal processing performed by the inner ear. The processor sends electrical stimulation information (e.g., pulse width, envelope amplitudes, etc.) back to the RF transmitter through the same cable. The electrode and amplitude information (reflecting current amplitude levels) is transmitted via RF through the skin to the implanted RF receiver, which in turn decodes the information and sends electrical stimulation to the electrode array implanted in the inner ear.

The main difference between what is currently available in the market and the developed speech processor is the replacement of the body-worn or BTE processor with the PDA and SDIO-interface board. The FDA-approved RF transmitter, containing the transmitting coil, is the same in both cases. This interface board was custom designed in our lab.

The main function of the PDA is to process the acoustic signal picked up by the microphone, which is located in the BTE. There are a number of signal processing algorithms that can be implemented on the PDA, see review in (Loizou 1998). Typically, the CI signal processing involves filtering the signal into a number of bands (12-22), and estimating the envelope (energy) in each band. The electrode and amplitude information of the processed signal together with stimulation parameters is then sent to the implant via the interface board.

Details of the platform are presented in the following sections. A brief overview of the hardware modules of the platform is presented first. This is followed by the software architecture for real-time and offline speech processer. Finally framework for bimodal (electric + acoustic) stimulation is described.

3.1 Hardware Overview

The research platform as a whole comprises of:

i. A portable processor in the form a of a smart-phone or a PDA for implementing and evaluating novel speech processing algorithms after long-term use (Lobo, et al. 2007),

ii. an interface board to connect the PDA with Freedom cochlear implant coil using secure digital input output (SDIO) port of the PDA (Lobo, Lee, et al. 2007), (Kim, et al. 2008),

iii. a bench-top and a portable stimulator (monopolar and bipolar) designed for electrical stimulation and neurophysiologic studies with animals (Kim, Gopalakrishna, et al. 2009) and

iv. a recording unit for collecting evoked potentials from the human subjects (Lobo, Loizou, et al. 2007).

Following sections provide details on each of the hardware components and their usage.

3.1.1 PDA

PDA is used as a portable processor for implementing signal processing algorithms. There are a number of reasons for choosing the PDA as the computing platform. First, it is light weight (4-6 oz), small in size and therefore portable. Second, it uses a powerful microprocessor, with majority of the PDAs running on high clock speeds and using Intel's PXA27x processor which are ARM-based processors and hence allow very efficient programming. Third, the software running on the PDA operates under the Windows Mobile environment which allows researchers to program novel sound processing strategies using high-level languages such as C, C++ and C#. LabVIEW also provides a toolbox to program using their conventional graphical environment. We have developed libraries which allow PDA platform to be interfaced with MATLAB for offline stimulation tasks. This is important as it offers flexibility and ease in terms of implementing and testing new algorithms for cochlear implants at a relatively short time, as opposed to doing the implementation in assembly language. Fourth, the PDA platform is easily "adaptable" to new and emerging technologies as they become available. That is, platform is easy portable to newer generation of PDAs and smart phones as more powerful and more energy efficient chips become available in the market.

The PDA used in the current study is HP iPAQ model hx2790 which houses ARM920T processor based on ARM9 processor core and allows ARMv4 Thumb Instruction Set Architecture (ISA).

3.1.2 SDIO Interface Board

The SDIO interface board is a custom developed board used to communicate and interface the PDA with the Cochlear Corporation's cochlear implants (including both CI22 and CI24 generations). The board plugs into the Secure Digital Input Output (SDIO) port of the PDA and enables PDA to stimulate the Cochlear Corporation's CI24 and CI22 implants. Very briefly, the PDA sends stimulus amplitude packets to the SDIO card using the SDIO 4-bit communication protocol. The amplitudes are converted by the FPGA to the embedded protocol (Daly and McDermott 1998) for the CI24 implant or the expanded protocol (Crosby, et al. 1985) for the CI22 implant, and finally sent to stimulate the implant via the Freedom Coil. Figure 3.1 shows the functional diagram of the SDIO board. It has two cochlear headset sockets which connect with the right and left ear BTEs (Behind the Ear units) via cochlear cables, thus allowing bilateral stimulation. To ensure that patients will not plug into the SDIO board a commercially available cable, we use a different size socket that does not mate with the commercial cable used previously in the body-worn processors. This cable is not commercially available.

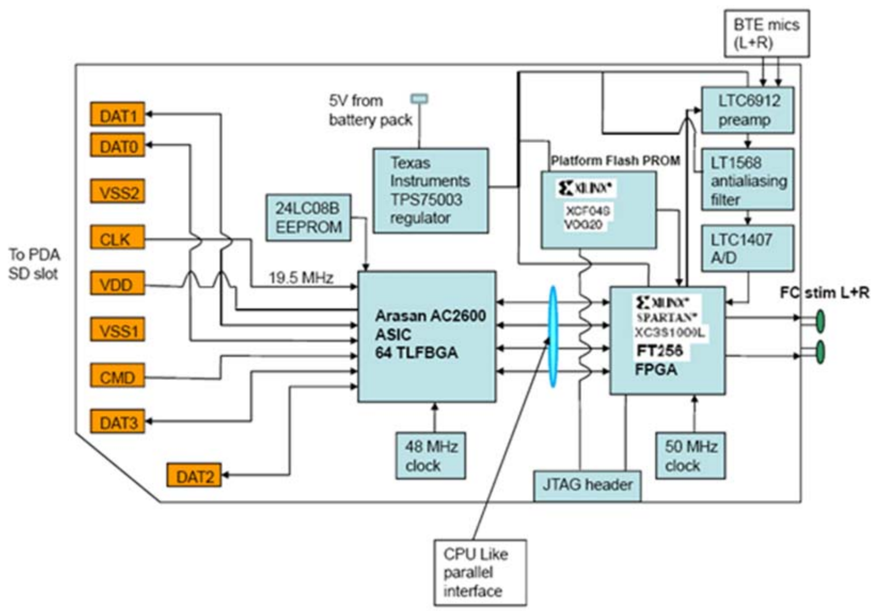


Figure 3.1. Functional Block diagram of the SDIO Board.

The board is equipped with a Xilinx Spartan 3 FPGA, Arasan SDIO interface controller, PROM, and power management circuitry to implement the communication interface between the PDA and the implant. The board is powered using a 5V battery source. An overview of the primary board components is given below:

The Arasan ASIC AC2600 is a SDIO card controller and implements the SDIO standard 1.2 and SD Physical Layer specification 1.10. It communicates with the SDIO host controller on the PXA270 processor in the PDA via a command response interface.

The 24LC08B EEPROM stores the initialization parameters for the ASIC to startup in CPU Like interface mode. The EEPROM communicates with the ASIC via an I2C bus.

The Xilinx FPGA (XC3S1000L) receives the amplitude packets from the Arasan ASIC and converts them to the Embedded protocol. The Embedded protocol bit stream is sent to the Freedom coil using a 5 MHz data signaling clock. The FPGA is clocked by a 50 MHz crystal. The FPGA logic implements a receive and transmit state machine and can support the 0.94 Mbps data link to the Freedom coil in the low rate stimulation mode using 5 cycles per cell. The peak stimulation rate is 15,151 pulses/sec. Using 4 cycles per cell the peak stimulation rate can be increased to19,608 pulses per second. The SDIO board can be used for bilateral or unilateral cochlear implant studies.

The Xilinx XCF04S is a Platform Flash PROM which stores the synthesized logic from which the FPGA boots off during power on.

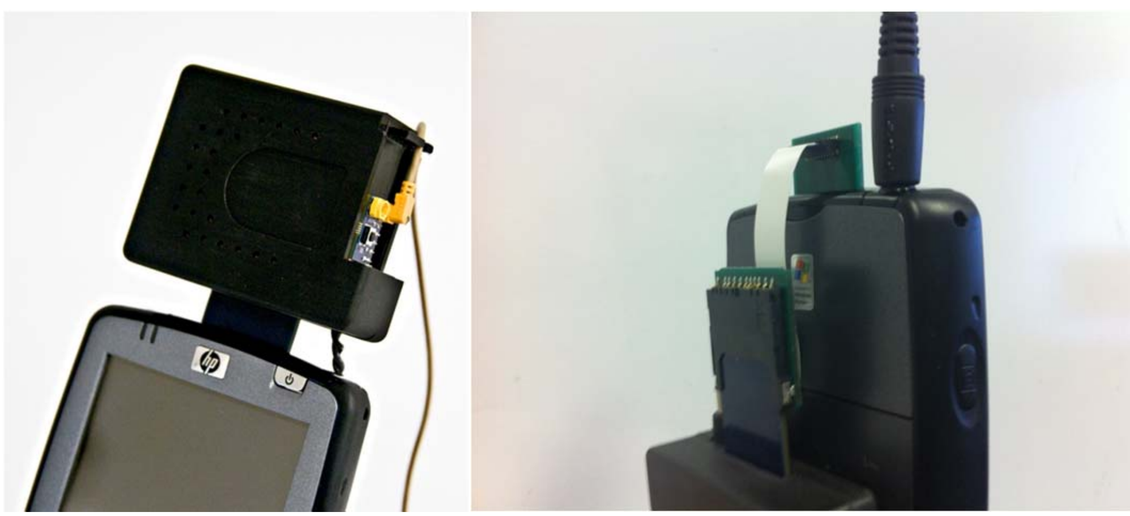
Preamplifier from Linear Technologies (LTC6912) provides two independent inverting amplifiers with programmable gain. The signal in each channel can be amplified with one of 8 gain settings 0, 1, 2, 5, 10, 20, 50 and 100 corresponding to -120 dB, 0 dB, 6 dB, 14 dB, 20 dB, 26 dB, 34 dB and 40 dB. The LTC6912 is programmed by the FPGA using the Synchronous Peripheral Interface (SPI) bus

The output of the preamplifier is filtered by the LT1568 anti-aliasing filter IC configured as a dual second-order Bessel filter with a cutoff frequency of 11,025 Hz.

The LTC1407 is a stereo A/D converter (ADC) and samples the microphone outputs from the bilateral BTE connected to their respective Freedom coils. The ADC has 1.5 Msps throughput per channel, operates at a sampling frequency of 22,050 Hz and presents a 14-bit two's complement digital output (interleaved left and right channels) to the FPGA. The samples are received by the FPGA over the SPI interface.

Texas Instruments' triple-supply power management IC (TPS75003) supplies power to the FPGA and platform Flash PROM. The TPS75003 takes a 5 V input from the external battery pack and generates 1.2 V for VCCINT (core voltage), 3.3 V for VCCO (I/O voltage) and 2.5 V for VCCAUX (JTAG, Digital Clock Manager and other circuitry). The TPS75003 is a switching regulator (Pulse Width Modulation control) type and is "on" only when power is needed.

The SDIO board also has a Lemo mini-coax connector providing access to an output trigger signal. A 5V trigger signal can be generated from the FPGA. This trigger signal can be used for synchronization purposes in external recording systems of evoked potentials. It is used



(a) (b)

Figure 3.2. SDIO Board with Enclosure and extender card.

as input to external neural-recording systems (e.g., Neuroscan, Compumedics Ltd), which have been approved by FDA for use with human subjects.

SDIO board is housed in an enclosure as shown in Figure 3.2 (a) for protection and safety purposes. An SDIO extender cable as shown in Figure 3.2 (b) may also be used for placing the board adjacent to the PDA rather than on the top for ease of use and to get access to the other PDA ports, e.g. an audio port in bimodal studies.

3.1.3 Stimulator Unit

Bipolar stimulator or BiSTM is a multichannel bipolar current source designed for acute experiments on percutaneous, animal cochlear implant systems. The BiSTM offers researchers the ability to study the effects of channel interactions on speech recognition particularly as a function of the electrode array configuration. While the 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.

The portable 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. At the core of the board is the 9-bit configurable current source chip, simply referred to as the BiSTM chip designed in our lab (Kim, Gopalakrishna, et al. 2009), (Loizou, Lobo, et al. 2011). The BiSTM chip is designed to provide programmable anodic and cathodic current pulses for stimulation. 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.

BT-BiSTM platform possesses the following specifications:

- 8 independently controlled bipolar channels or up to 8 independently controlled time interleaved monopolar channels, each electrically isolated and charge-balanced
- 5V compliance voltage
- 1mA maximum current amplitude per channel
- 9-bit current amplitude resolution per channel (1.95 μA)
- 4μs minimum pulse width per channel (1 sec maximum pulse width)
- 0μs minimum interphase gap per channel (maximum allowed interphase gap depends on maximum pulse width)
- 4μs minimum inter-stimulus interval per channel (maximum depends on desired pulse rate)
- 83.3 kHz maximum pulse rate per channel
- >50MΩ output resistance per channel

With these features, 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, inter-stimulus interval (ISI) and pulse rate, a multitude of stimulation patterns can be created both in phase (simultaneous) or interleaved across multiple channels.

3.1.4 Recording Unit

The PDA platform has the capability to record EEG and cortical auditory evoked potentials (CAEPs) via data acquisition cards. Off the shelf data acquisition cards (such as Dataq-CF2 and CF-6004) plug into the compact flash slot of the PDA and can be programmed in C or in LabVIEW. A major limitation of the commercial data acquisition cards is limited number of recording channels. In order to compensate this, the SDIO board was equipped with a Lemo mino-coax connector which provides access to an output trigger signal. This trigger signal can be used for synchronization purposes in external recording systems of evoked potentials. The trigger output signal is not connected directly to the patient; hence it poses no safety concerns. Rather it is used as input to external neural-recording systems (e.g., Neuroscan, Compumedics Ltd), which have been approved by FDA for use with human subjects.

The above sections provided an overview of the hardware modules involved in the PDA platform. However, not all hardware modules are required for individual experiments. For example, hardware required for the real-time stimulation of speech processing is a PDA and an SDIO board connected with Freedom BTEs. For lab testing purposes, an implant emulator is used.

3.2 Software Architecture

PDA-based speech processor has two modes of operation:

i. Real-time Speech Processor which allows both electric only and electric plus acoustic stimulation (EAS) in real-time and

ii. Offline Speech Processor which allows speech processing in offline mode through MATLAB running on a PC. In addition to the bimodal stimulation capability, it also supports psychophysics.

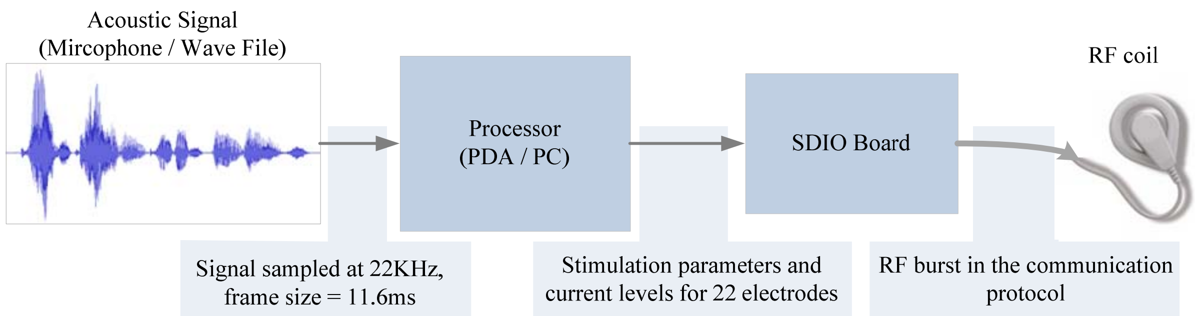


Figure 3.3. Generic Signal Flow for speech processing through real-time or offline speech processors.

A simplified generic signal flow for both modes is depicted in Figure 3.3. For real-time mode, acoustic signal is acquired from the microphone and processed within the PDA. Alternatively, for offline mode, a wave file is read and processed by the software running on the PC. The processing results in a set of n amplitudes which represent energy levels in each of *n* bands. Note that *n=22* in our case corresponding to the total number of electrodes available in the Cochlear Corporation's implant. The set of amplitude levels and electrode information are sent to the interface board via the SD slot. An FPGA (Xilinx Spartan XC3S1000L) on the interface board receives the envelope amplitudes and prepares them for transmission using an RF data communication protocol (Daly and McDermott 1998). The FPGA sends, via the cable, a stream of RF bursts containing information about the current levels (amplitudes) to be used to stimulate each electrode along with a set of stimulation parameters, such as pulse duration and

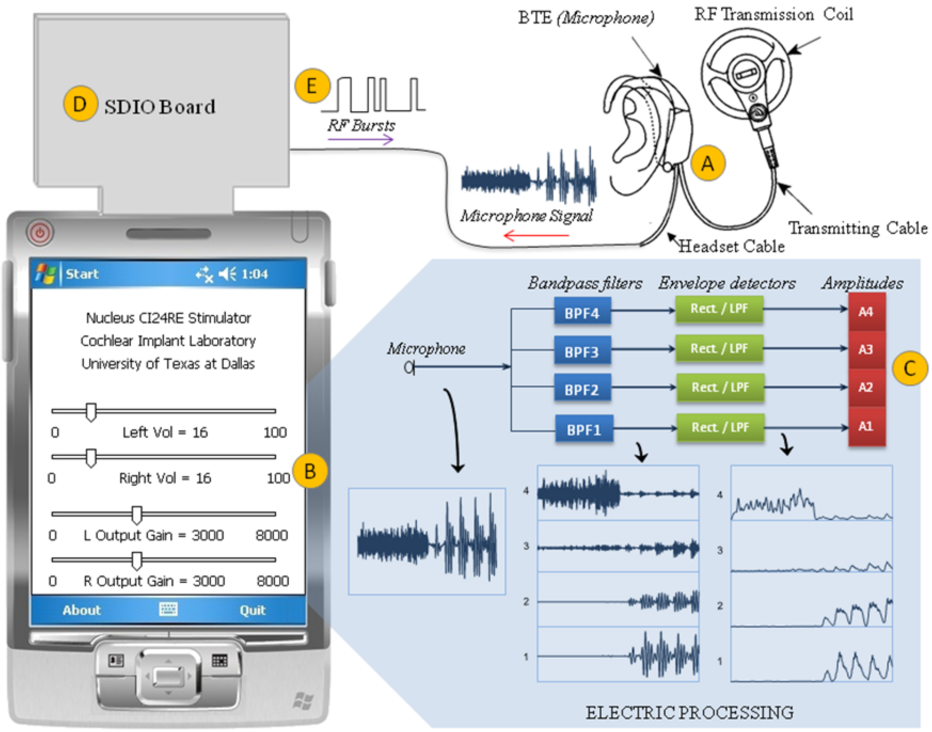


Figure 3.4. Signal Flow in the PDA Speech Processor in real-time mode. The acoustic signal is picked up by the microphone (A), sent (via the headset cable) to the SDIO interface board (D), which is then sent to the PDA. The PDA (B) processes the signal and generates a set (one for each channel of stimulation) of amplitudes (C). The amplitudes are sent to the FPGA interface board (D), which are then prepared for transmission to the cochlear implant in the form of RF bursts (E).

mode of stimulation (bipolar vs. monopolar). The latter set of parameters is used by the implanted RF decoder for constructing biphasic pulses.

Following sections describe both real-time and offline software modes in greater detail.

3.2.1 Real-time Speech Processor

Real-time speech processor mimics a commercial processor such that all speech processing and stimulation is done in real-time. Figure 3.4 provides a general overview of the signal flow involved in the PDA-based real-time speech processor. The acoustic signal is picked up by the microphone located in the BTE and sent to the FPGA interface board via the headset cable. The interface board samples the signal binaurally at a rate of 22 kHz/channel and sends the sampled (digital) signal to the PDA via the SD slot. The PDA processes the digital signal via a speech coding algorithm (e.g., CIS or ACE) and produces a set of amplitude levels for *n* bands representing the energy levels in respective bands. These amplitude levels and electrode number information together with stimulation parameters is sent to the SDIO board which transmits them to the implant using RF protocols specific to the implant.

In a nutshell, PDA acts a body-worn processor more like the olden day processors but with additional features, programming flexibility and user-interface to interact with speech processing parameters. Following section describes the software architecture for real-time mode.

Software Architecture

The software running on the PDA performs the signal processing (i.e., implements the speech coding strategy) while the firmware running on the FPGA implements the communication protocols needed for proper communication with the Freedom transmission coil. Figure 3.5 shows the organization of the software running on the PDA and FPGA. There are two inputs to the PDA software: i) Acoustic Signal from the microphone and ii) Patient MAP. The patient's MAP file is an ASCII text file in a custom format stored on the local PDA directory. Routine *Read_Patient_File()* reads the contents of the MAP file and returns a map structure containing stimulation parameters specific to the patient. Some of these parameters are shown in Table 3.1.

The researchers have the freedom to program any strategy they find appropriate for their research. A distinction between CIS–like and ACE-like strategies is important to make because different limits are imposed on the stimulation parameters of each depending on whether the same set of electrodes are stimulated in each cycle (e.g., CIS-like) or a sub-set of electrodes are stimulated in each cycle (e.g., ACE-like). The limits on the stimulation parameters were obtained from Cochlear Corporation. The CIS-like and ACE-like options taken together cover all possibilities in terms of the investigators programming the PDA processor with new signal processing strategies.

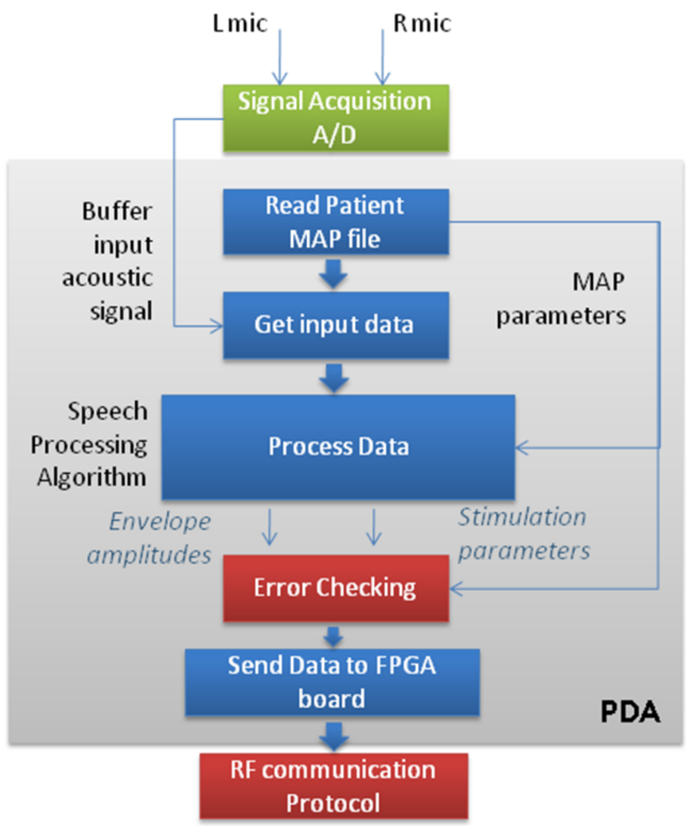


Figure 3.5. Organization of the software running on the PDA and FPGA.

The stimulation parameters saved in the patient's file are checked in every cycle prior to stimulation. The error checking is performed on both the PDA and FPGA. More details on error checking are given in the following section.

The *Get_Input_data block* (Figure 3.5) captures and buffers the acoustic signal from the microphones at 22050 Hz sampling rate in frames of 11.6ms binaurally. The *Process_Data* block takes as input the patient's MAP file and the acquired signal buffer and returns the amplitudes to be transmitted to the cochlear implant via the SDIO board. The amplitudes can be obtained either by band-pass filtering the signal into a finite number of bands (e.g., 12, 22) and detecting the envelope in each band, or by computing the FFT spectrum of the signal and estimating the power in each band depending upon the speech strategy. The researchers have the freedom to implement their own sound processing strategy. In the current study, ACE strategy as reported by (Vandali 2000) was implemented in fixed-point C. The signal flow in ACE strategy is illustrated in Figure 3.6.

Table 3.1 Patient File MAP parameters.

Parameter	Options
Implant/Electrode type	CI24RE(CS/CA) / CI24M /CI24R (CS/CA) / CI22M / ST
Left/Right ear Implant Strategy	ACE / CIS / SPEAK
Number of Implants	Unilateral / Bilateral
Electrode Configuration	MP1 / MP2 / MP1+2
Number of active electrodes	(1 to 22) depending upon the strategy
Left/Right Stimulation Rate	250Hz – 3500 Hz
Left/Right Pulse Width	9.6µs-400µs depending upon rate
Left/Right THR	0 – 255 clinical units
Left/Right MCL	0 – 255 clinical units

Very briefly, acoustic buffer of 11.6ms for each left and right is first pre-emphasized using a pre-emphasis filter. The signal is then buffered into analysis windows of 5.8ms and multiplied with a Blackman window. Overlapping of window depends upon the stimulation rate. For each window, 256 point FFT is computed. Magnitude squared FFT is passed through the triangular filters which are weighted according to the 22 band gains. Next, a shell sort routine sorts magnitudes of all bands and selects *n-maxima* (bands with highest magnitudes). In order to convert acoustic amplitudes to electrical dynamic range, magnitudes are compressed using a logarithmic compression function which is implemented as a look-up table. Intel IPP routines and various signal processing tricks have been used wherever possible to lower the computation cost and keeping all computations in real-time.

Electrical amplitude levels are passed through the *Error_Checking* block (described later) and finally the *Send_Data* block transmits the data (envelope amplitudes and stimulation parameters) to the SDIO board. The firmware running on the FPGA prepares the received data for RF transmission using the expanded protocol (Crosby, et al. 1985) for the CI22 system and the embedded data protocol (Daly and McDermott 1998) for the CI24 system.

Error Checking

A number of mechanisms have been set in place to ensure the safety of the patients. Foremost among those mechanisms is to keep the firmware running on the SDIO interface board unreadable and un-modifiable. This is done to ensure that the cochlear implant patients will not be overly stimulated. Secondly, software safety checks are set in place on the PDA side for checking: (1) the range of envelope amplitudes and (2) the range of stimulation parameters (e.g., pulse width) to ensure that they fall within the permissible and safe range. All safety checking

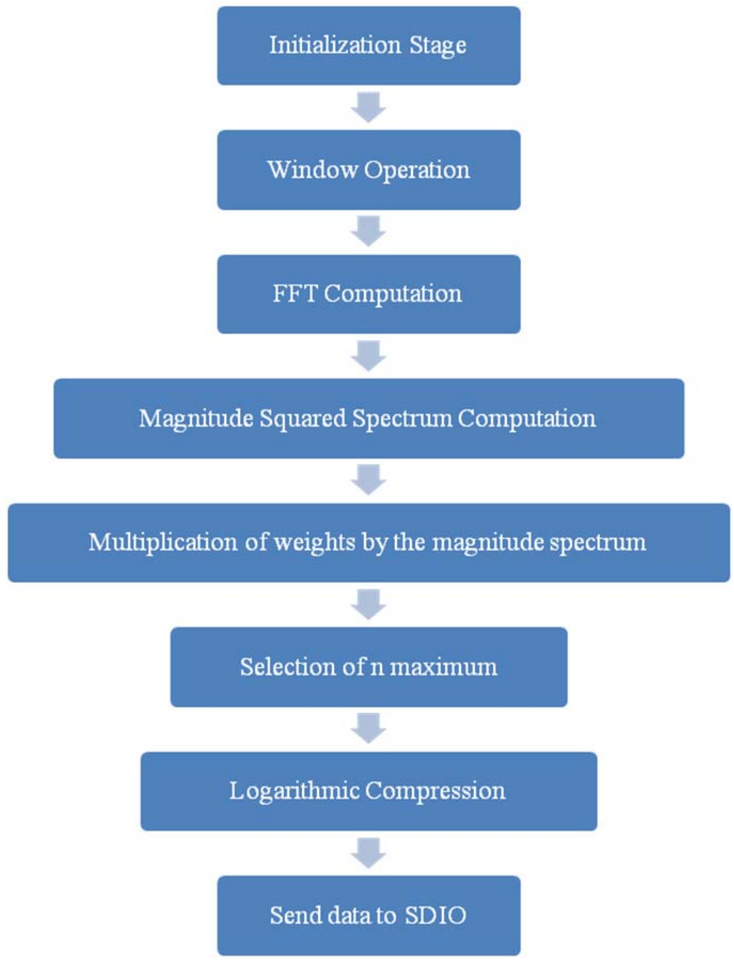


Figure 3.6. ACE Algorithm.

routines are hardcoded and are thus not accessible in the code. Only stimulation parameters that have already been approved by the FDA and are currently in use by patients are allowed.

The error checking software routine as shown in Fig. 5 is the gateway routine to the electrical stimulation. This routine takes as input the stimulation parameters from the MAP along with the envelope amplitudes. There is a limit on range of the parameters set by the manufacturer for safe operations. For instance, the biphasic pulse width cannot exceed 400µsecs/phase (in general, the maximum allowable pulse width depends on the stimulation rate). This 400 μsecs/phase upper limit is based on evidence from physiological studies published by (McCreery, et al. 1988) and (R. Shannon 1992). The stimulation parameters are checked in each cycle to ensure that they fall within the acceptable and safe limits. The permissible range of stimulation parameters was taken from Cochlear Corporation's documents. Validation tests were conducted to verify this. If any stimulation parameters are found to fall outside the permissible range, they are saturated to the maximum allowable value.

Checking valid range of envelope amplitudes

The envelope amplitudes of each electrode need to be limited within the range of threshold (THR or T) to the most comfortable (MCL or M) levels (the T and M levels are expressed in clinical units and can range from 0 to 255). Most importantly, the envelope amplitude of each active electrode is checked to ensure that it is smaller than the M level of each electrode. If any of the amplitudes falls outside this range, the program saturates the amplitude to the corresponding M level. This is done to avoid overstimulation. The M levels can be obtained using the clinical fitting software, Custom Sound (v.2), and are subsequently entered into the patient's file. Rechecking stimuli amplitudes in software is done as an additional measure to ensure that any change in M levels would not cause overstimulation.

For the CI24 and Freedom implants (CI24RE) that use the contour electrode array, further safety checks are set in place to limit the charge density. More specifically, the maximum stimulation level allowed depends on the pulse width, from which a maximum charge-density level (CDM) is computed. The envelope amplitude is thus not allowed to exceed the minimum of the M and CDM levels.

Checking for valid range of stimulation parameters

The relationship between the various stimulation parameters available is complex and it depends among other things on: (1) the generation of the Nucleus device (e.g., CI22, CI24), (2) the electrode array used and (3) the stimulation strategy used (CIS-like vs. ACE-like). For instance, the allowable pulse width depends on both the stimulation strategy used and the generation of the Nucleus device. These relationships and dependencies among the stimulation parameters were taken into account when writing and testing the error checking routines. Provisions were made for the above dependencies and more precisely about:

i. Valid Stimulation Modes
ii. Pulse-rate and Pulse-width dependency
iii. Charge density limitations for different electrode arrays

Validation tests were conducted to verify that the stimulation parameters and stimuli pulses are always within the safe limits.

3.2.2 Offline Speech Processor

Offline version of the PDA platform is based on a PC running MATLAB where all processing takes place while the PDA acts as an interface to the implant. The software architecture is designed such that PDA acts a server which accepts the incoming connections and the PC acts as a client with MATLAB as a front-end as shown in Figure 3.7. Therefore, overall design can be broken down into three main software components:

i. Server running on the PDA,
ii. MATLAB client (.mexw32 or .mexw64 dll) called from the MATLAB front-end, and
iii. MATLAB front-end running on PC.

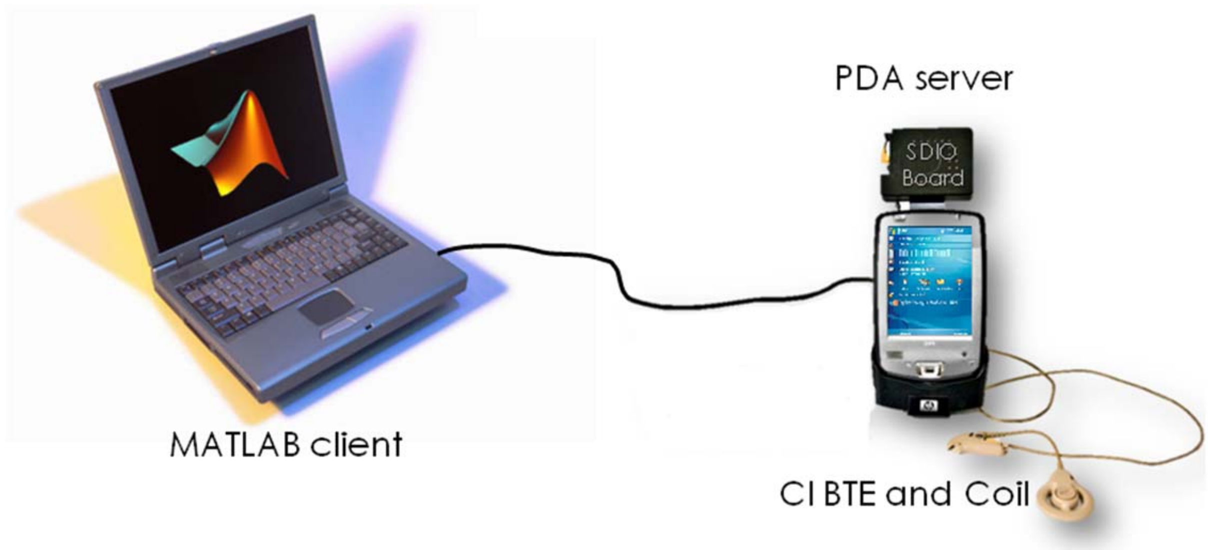


Figure 3.7. High-level diagram for offline-mode set-up.

Server client interface is based on Winsock (Windows Sockets API) which is a technical specification that defines how Windows network software should access network services, especially TCP/IP. It defines a standard interface between a Windows TCP/IP client application (such as an FTP client or a Gopher client) and the underlying TCP/IP protocol stack (Makofsky 2003). Figure 3.8 shows the transfer of parameters and amplitudes from MATLAB to the PDA and status returned from the PDA to MATLAB.

Figure 3.8. MATLAB-PDA interface using Windows Sockets.

MATLAB front-end

MATLAB front-end, as the name suggests, is the application layer of the system around which most researchers work. It can either be a simple command script to create synthetic stimuli and stream them to the PDA by calling the client dll (dynamic link library - responsible for invoking client-server communication protocol) or it could be an elaborate GUI or application which implements speech coding algorithms and uses client dll as a backbone. A variety of applications can be created at the front-end suitable for different experiments using the same client dll. An example of such an application is shown in Figure 3.9 which shoes complete software suite to read and modify map files and perform psychophysics experiments similar to the Custom Sound. Other applications for more specific experiments will be presented in the next section.

MATLAB-to-PDA interface is designed with the goal to be simple, user-friendly and flexible so that it provides researchers with enough feature space to design experiments they could not with conventional researcher interfaces. This is achieved by limiting all the overhead code and communication protocols to the PDA component or the server hence allowing the researchers to focus more on experiments than on coding while saving a lot of time. The MATLAB front-end has following important functions:

1) Load patient MAP: First, the patient map is loaded or created either from an existing patient map or an application to load stimulation parameters specific to a patient. The file format remains the same as the one used in the real-time version.

2) Check Stimulation Parameters: Parameters loaded from the MAP file are checked for the safe implant operation within the implant limits. These safety checking routines are

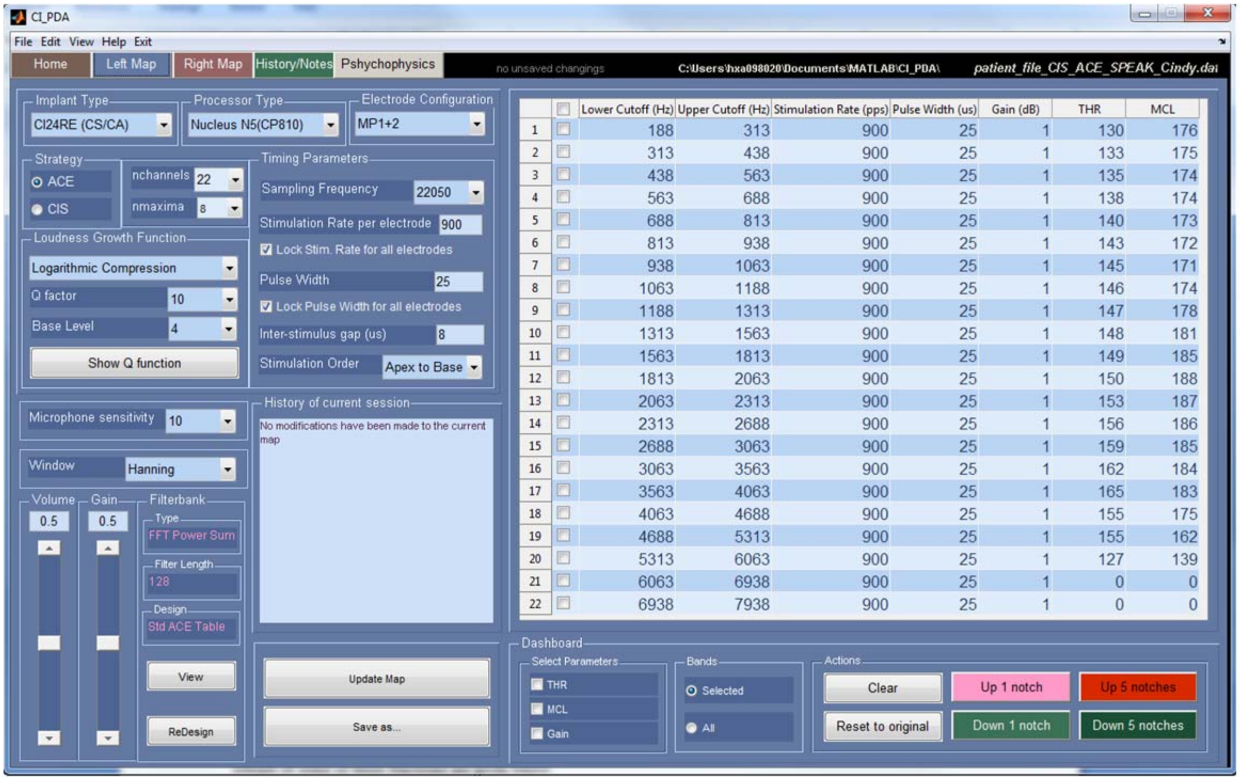


Figure 3.9. Software suite for Offline mode. It is designed to load and create patient maps and performs psychophysics.

similar to the ones used in the real-time version but in the offline version, they are embedded as MATLAB pcodes for additional safety. Parameters which are checked for the safe operation are stimulation rate and pulse width for both left and right ear implants depending upon the implant type. In addition to this, rate-centric and pulse-width centric parameter checking routines are also hardcoded. In a rate-centric routine, for example, if a user specifies stimulation rate and pulse width that are not realizable, the pulse width will be adjusted to fit the pulse rate. These routines are also hard-coded at the server end for further safety purposes.

3) Create Stimulation Data: Stimulation data essentially comprises of two data arrays containing magnitude and electrode information. Electrode information comprises of an array

containing the stimulation sequence of the active electrodes while magnitude information corresponds to the amplitude levels of the stimuli for the respective electrodes. Stimulation data may be loaded from a pre-processed file or it may be created by implementing any speech coding strategy. Alternatively, a waveform of synthetic stimuli may be created for psychophysical experiments as elaborated in the next section.

4) Error Checking: Before streaming stimuli to the implant, a final check on the stimuli amplitude levels is done to make sure that the amplitudes fall within the safe range specified by the patient's map. This is done by comparing the amplitude levels of each electrode with the MCL and THR levels of that electrode.

5) Call to client: Finally, the client dll is invoked which transfers stimulation parameters and data to the implant via the PDA. The client dll initializes Winsock, creates a socket, connects to the server and transmits stimulation parameters and electrode/amplitude buffer created in the MATLAB front-end application. It does this in two steps to match the receive function on the server: first the number of 11 ms frames, nframes, and the number of pulses per frame are transmitted. Second, nframes frames are transmitted continuously with the time interval between frames set to 11ms. The dll is compiled from the C source using the MATLAB MEX compiler.

PDA component – Server

The PDA component initializes Winsock, creates a socket, binds the socket, '*listens*' on the socket, accepts incoming connections, and performs blocking receives to receive the parameter and stimuli data from the client. The *'receive'* is performed within a thread in two steps. In the first step, information about the total number of frames (*nframes*) is received. The server then performs *nframes* receives, each time sending the data to the SDIO board verified by the error-checking routine. After *nframes* are sent and received, the server closes the socket and the connection. This process is illustrated in Figure 3.10.

The PDA server which runs continuously in a Windows thread automatically initializes a new connection for the next incoming stimulus and waits for the client to transfer the corresponding next set of parameter and amplitude frames. In this way, the transfer of amplitudes takes place on demand i.e. the transfer is made under the complete control of the user.

The PDA component is built as a Windows Mobile 5.0 executable using Visual Studio

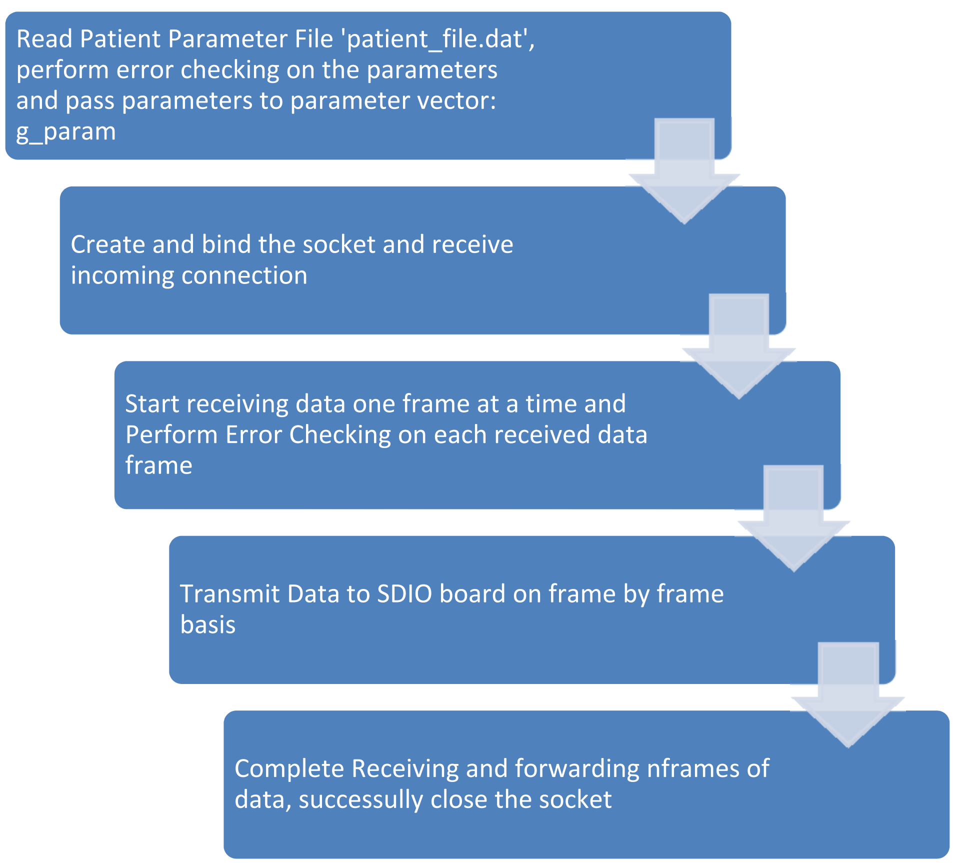


Figure 3.10. Program flow of server routine.

2005/2008 Professional in C. The executable is deployed on the PDA and is run from the desktop remotely using the Windows Remote API (RAPI) application prun.

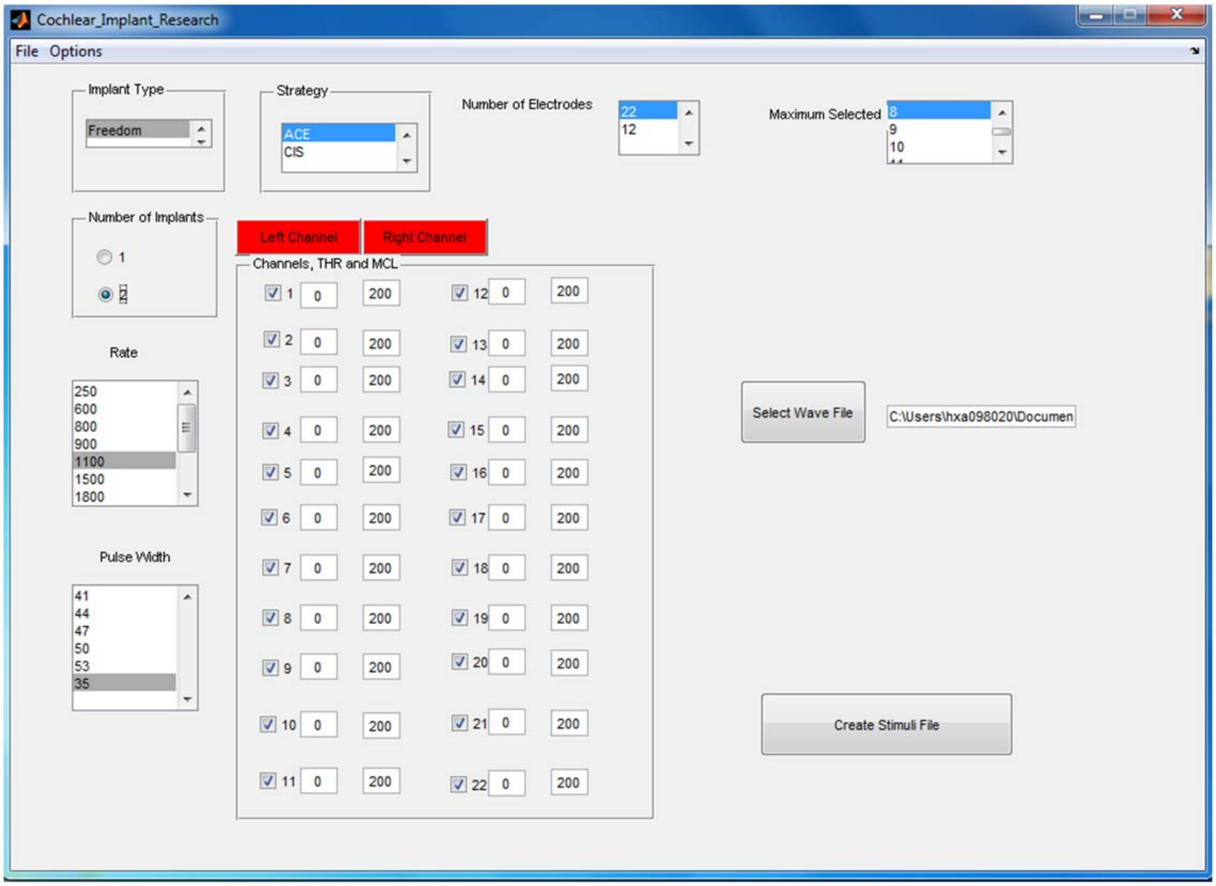


Figure 3.11. Offline Mode Application – Process and Stream. Process and stream application for converting an audio signal to stimuli pulses using a set of parameters (provided by the user) and streaming them to the CI using the PDA platform.

Applications

Flexibility to program in MATLAB in offline mode provided great opportunity to design applications targeted to address various behavioral and psychophysical experiments which were not possible with conventional platforms.

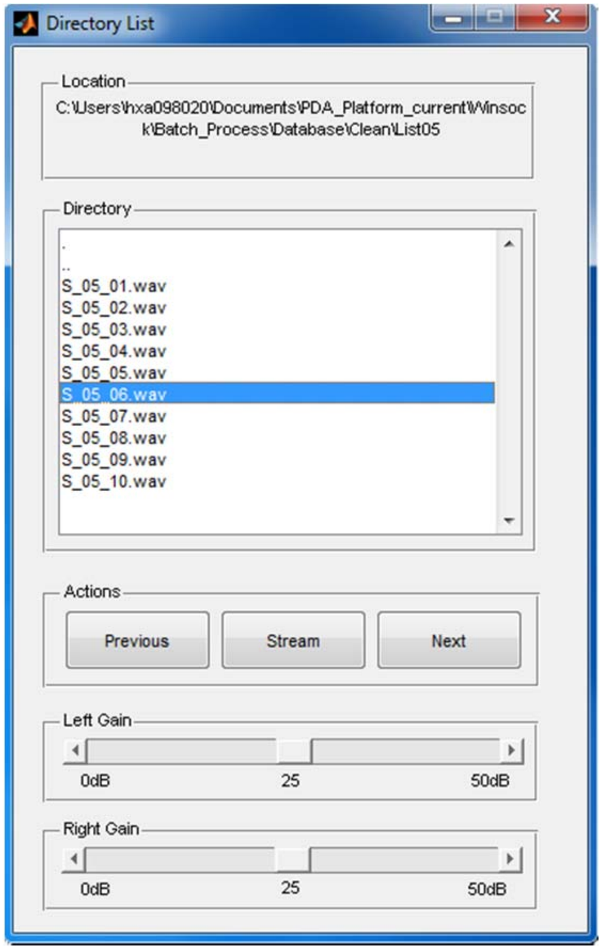


Figure 3.12. Offline Mode Application – Batch Process. This application allows user to select an audio (.wav) file and process and stream the stimuli to the implant via the PDA in offline mode in a playlist manner.

Figure 3.11 shows MATLAB application to to convert an audio signal/file to a stimulus file and then stream it to the patient using the platform while providing complete control over stimulation parameters, speech processing strategy and patient map. Complete speech processing algorithm is implemented in MATLAB which acts as a flexible environment to implement algorithms with great ease.

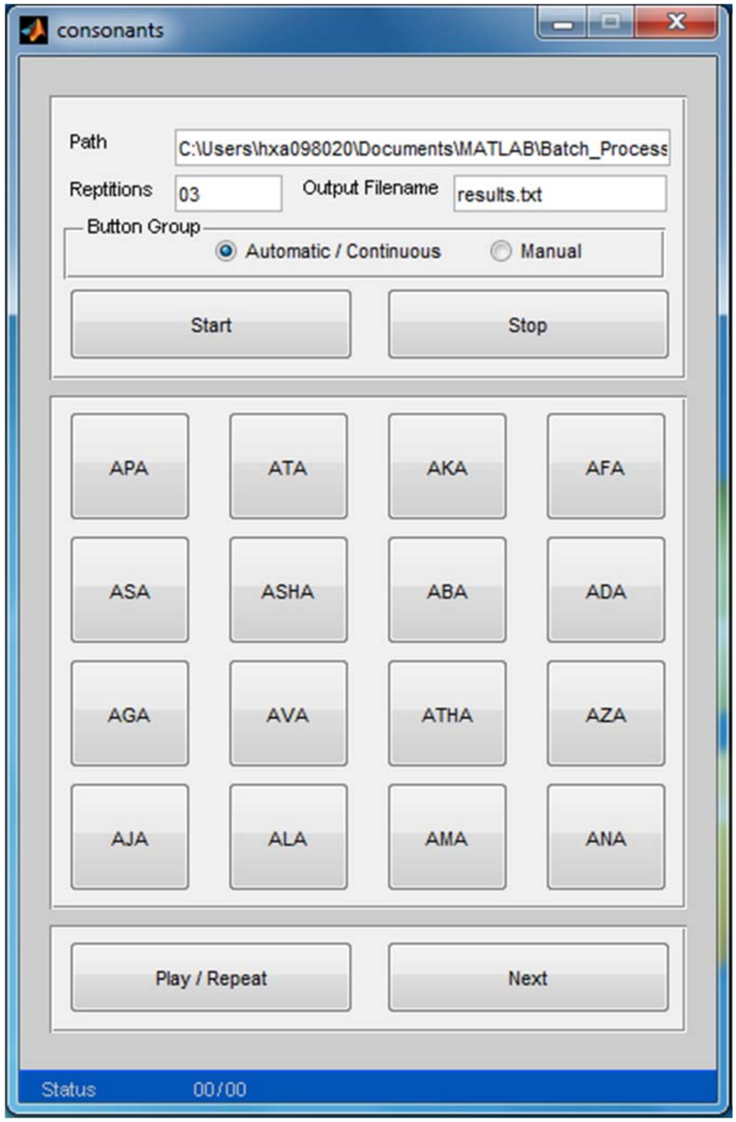


Figure 3.13. Offline Mode Application – Consonants Testing.

Figure 3.12 shows an application to load a playlist of audio files intended to be processed and streamed to the implant via the PDA. User selects a MAP file and is then able to browse through computer directories to locate audio files. Once an audio file is selected, pressing the ‘stream’ button processes the audio file through ‘ACE’ strategy and streams it to the implant. This application allows bilateral implant configuration and also supports EAS. User is able to set overall gain for both left and right implants as well as able to tweak stimulation parameters for better performance.

Figure 3.13 shows an application to test the intelligibility of consonants through a MATLAB GUI which automatically scores the users performance and provides training to the users in different sound environments.

One of the most interesting capabilities of offline mode is to design psychophysics experiments. These are discussed in the next section.

Psychophysics

PDA platform has the ability to control stimulation of each individual electrode along with the stimulation parameters. We can stimulate one electrode only or all 22 electrodes (as long as stimulation parameters are within range to allow such stimulation) in any timing sequence possible. This allows conducting psychophysics experiments with great ease and flexibility. Hence offline software mode has been used to design various psychophysics experiments. Figure 3.14 shows view of the offline suite of applications which is capable of performing psychophysics within the same application. User can either start from an existing map or create patient map by stimulation individual electrodes at different amplitude levels. Single or multiple electrodes with different types of stimuli can be activated for each trial.

Another simple application to perform psychophysics is shown in Figure 3.15. which streams different types of stimuli waveforms e.g., tones and chirps with different parameters to the specified electrodes. User has the ability to control each individual electrode and change the gain, stimulation rate, pulse width and number of active electrodes.

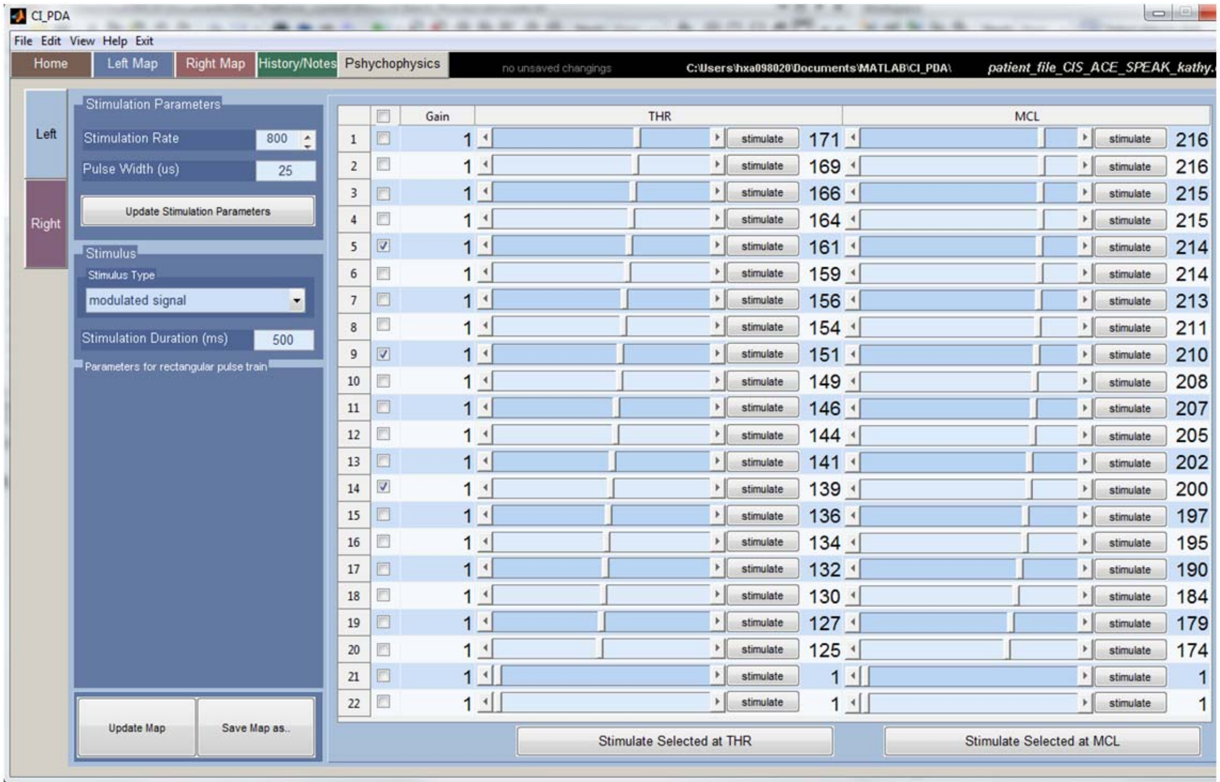


Figure 3.14. Psychophysics Application.

The platform was also successfully integrated with Percept (Goldsworthy 2011), which is a software developed by Sensimetrics Corporation to facilitate the design and assessment of sound processing strategies. Percept offers a wide range of psychophysics experiments which can easily be performed on human subjects in lab environment.

Figure 3.16 illustrates some examples of complex stimuli waveforms which can be generated to conduct psychophysics experiments.

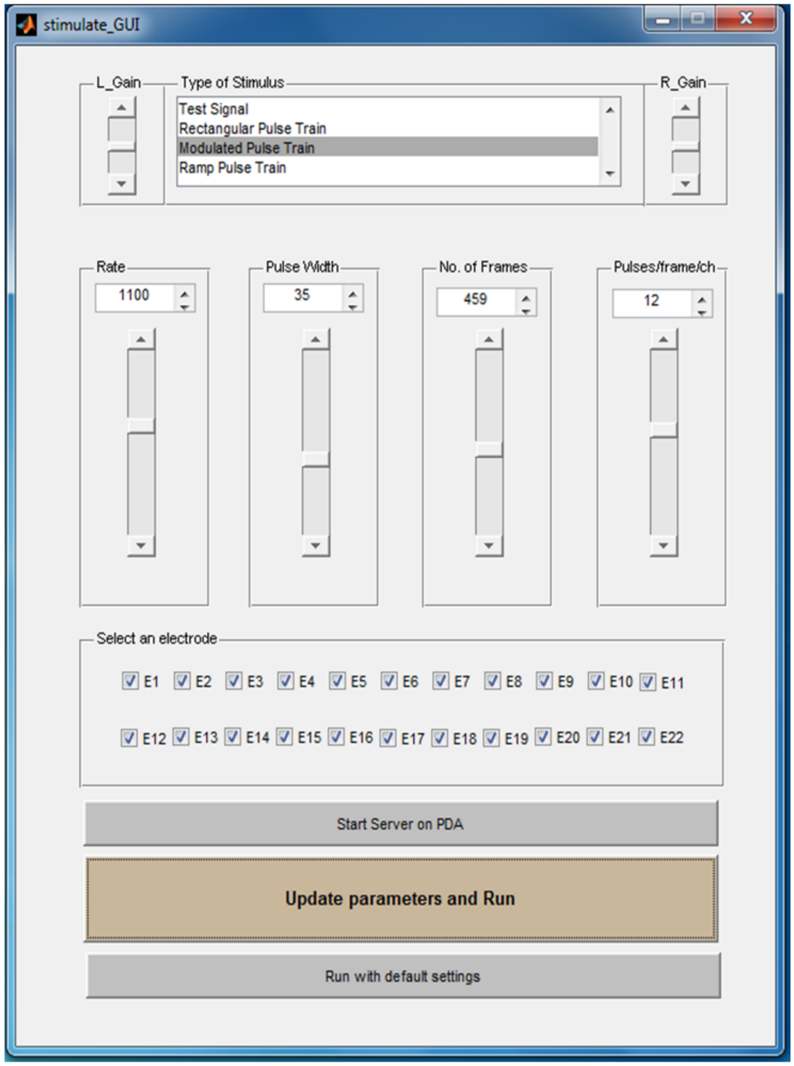


Figure 3.15. GUI of a simple psychophysics applications.

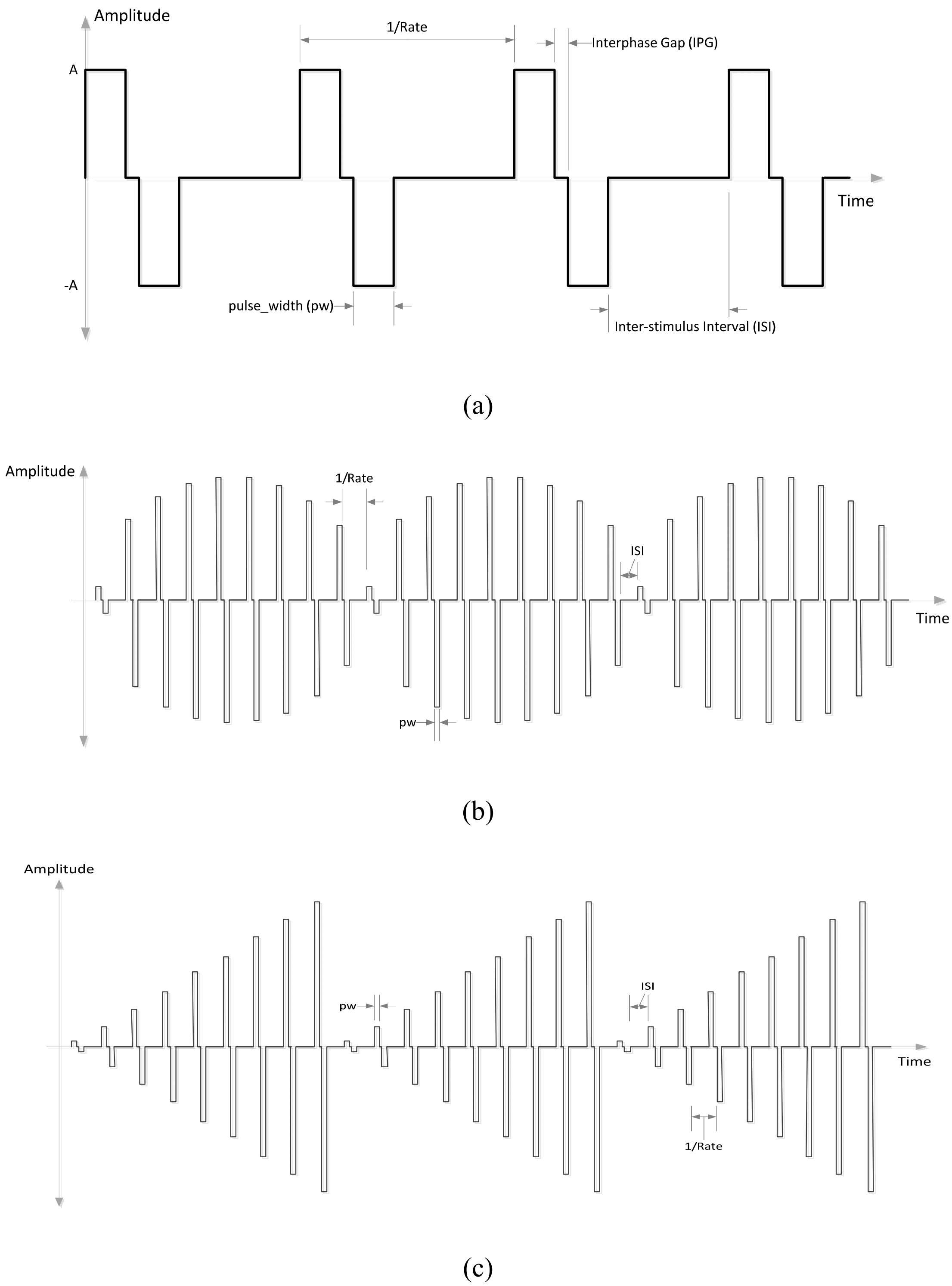


Figure 3.16. Examples of stimuli waveforms for psychophysics experiments. Rectangular pulse train; (b) Modulated pulse Train; (c) Saw-tooth pulse train.

3.2.3 Electric plus Acoustic Stimulation (EAS)

A number of studies recently have focused on the combined electric and acoustic stimulation (EAS) as a rehabilitative strategy for sensorineural hearing loss (HL) (Turner, Reiss and Gantz 2007), (Wilson and Dorman 2008), (Gantz, Turner and Gfeller 2006), (Gantz, Turner, et al. 2005), (Kiefer, et al. 2005), (Gifford, et al. 2007), (Ching, et al. 2001), (Ching, Incerti, et al. 2006). It is now well established that patients fitted with a cochlear implant (CI) and who have residual hearing in one or both ears and combine the use of hearing-aid with their implant, receive a larger benefit in speech understanding compared to electric-alone or acoustic-alone stimulation. That is to say, combined electric and acoustic (EAS) stimulation has a strong synergistic effect (Turner, Reiss and Gantz 2007) both when acoustic information is delivered ipsilaterally to the implant (e.g., hybrid implants with partially inserted (short) electrode arrays) or when delivered contralaterally (implant in one ear and hearing-aid in the other). We refer to the latter mode of stimulation as bimodal stimulation. This improvement is more evident in the noisy conditions as suggested in (Turner, et al. 2004) and (Qin and Oxenham 2003) and is primarily attributed to access to more reliable F0 cues in the acoustic portion.

The functionality of the platform is extended to include acoustic stimulation in addition to electrical stimulation for researchers interested in experimenting with bimodal CI users. Similar to electric-only stimulation, the platform can be operated in two modes for bimodal experiments, i) real-time mode and ii) offline-mode. In the real-time mode acoustic and electric stimuli are delivered to the user in real time just like their own clinical processor or hearing aid. All the processing is carried out in the PDA in real-time. The offline mode, on the other hand, is based on a PC running MATLAB. The user selects an audio file from the PC which is processed by a speech processing strategy in MATLAB and the stimuli are streamed to the implant and ear-piece.

In addition to the basic hardware requirement discussed earlier in this chapter, bimodal experimentation involves one additional hardware module, a high fidelity earphone system. For the current study we have used commercially available insert earphones form E.A.R. Tone Auditory System. Insert phones are based on a transducer which is connected to the PDA via an audio cable. The other end of the transducer connects with ear-tip through a tube which transmits acoustic stimuli to the ear. Figure 3.17 shows the insert phones. A portable audio amplifier is sometimes used for additional acoustic gain. In this way, acoustic stimulation can be provided as loud as 120dB.

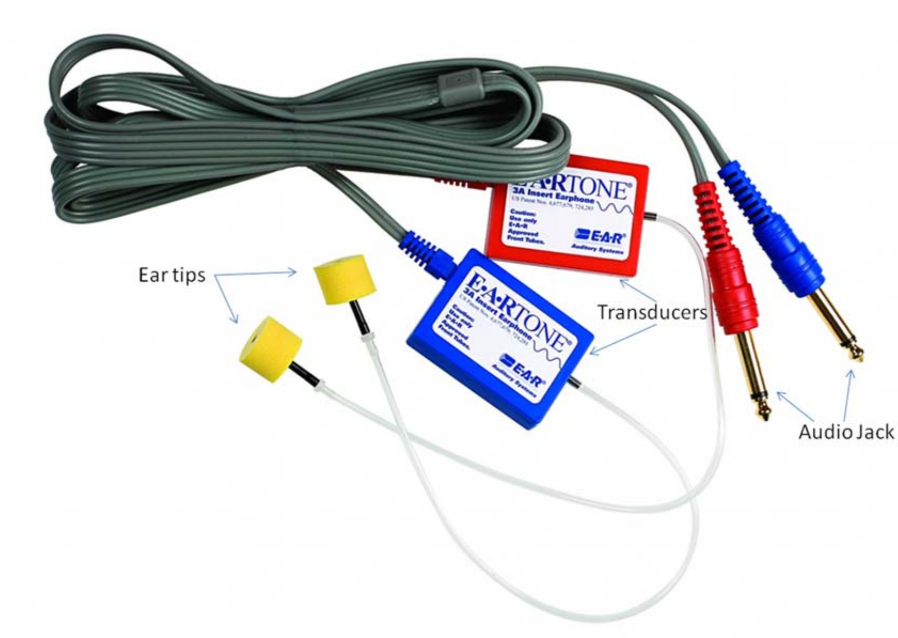


Figure 3.17. Insert phones for acoustic stimulation.

Software Architecture for Bimodal Stimulation

Figure 3.18 shows a generic signal flow when an acoustic signal is processed through hearing-aid processing routines and speech coding strategy simultaneously to deliver acoustic and electric stumili. Speech coding strategies like ACE and CIS and are dependent on a patient's electric map whereas hearing-aid processing routines are dependent on patient's audiogram

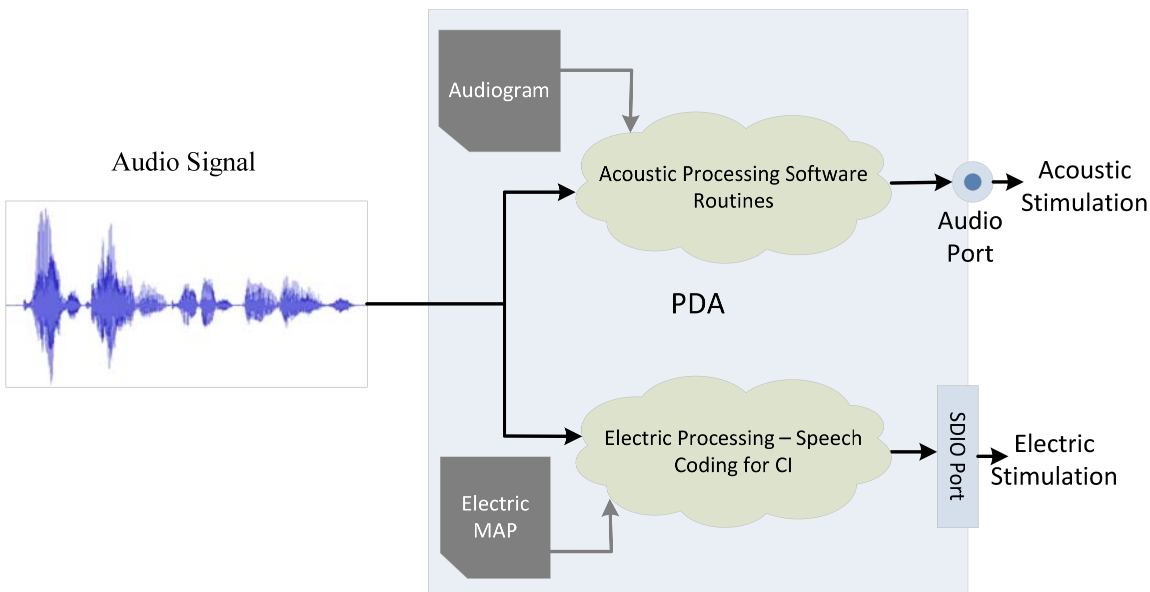


Figure 3.18. Signal Flow for electric plus acoustic stimulation.

Real-time mode

Figure 3.19 provides a general overview of the signal flow involved in the PDA-based real-time speech processor. The acoustic signal is picked up by a microphone located in the BTE and sent to the SDIO interface board via the headset cable. The interface board samples the signal binaurally at a rate of 22050 Hz/channel and sends frames of the sampled (digital) signal to the PDA via the SD slot. The PDA processes each frame of 11.2ms simultaneously through a speech coding algorithm (e.g., ACE or CIS) for electric stimulation and via an audio processing routine for acoustic stimulation. The electric processing routine requires a patient's clinical electrical map while the audio processing routine utilizes patient's audiogram for subjective processing. Electric processing produces a set of amplitudes representing the energy levels in each of 22 frequency bands, a subset (8-12) of which are used for stimulation. These amplitude levels are then sent to the SDIO board which transmits them to the implant using RF protocols specific to the implant. Concurrent with this, the processed acoustic buffer is streamed to the audio port of the PDA for acoustic stimulation. In this way, both electric and acoustic stimulations are perfectly synchronized. This is a remarkable feature of the PDA platform. Hearing aids and cochlear implants in practical use are completely independent of each other. They have their own, usually different, audio buffer sizes and audio processing delays. Hence, acoustic and electrical stimulation are not necessarily in perfect synchronization.

Offline Mode

Figure 3.20 depicts the software architecture for the offline processor in bimodal mode. MATLAB reads patient's map and processes the given audio file (.wav format) through a speech processing routine. Using Windows RAPI libraries the audio file and map file are copied to the PDA. The processed signal in the PC comprises of a set of amplitude levels and stimulation parameters which are streamed to the PDA server using Windows Sockets (Winsock) API, a technical specification that defines how Windows network software should access network services, e.g. TCP/IP (Makofsky 2003). After successful transfer, the PDA server performs error checking on the received data and buffers the electrical amplitudes and acoustic samples in frames of 11ms. These frames are then continuously and synchronously transmitted to the implant and the earphones respectively.

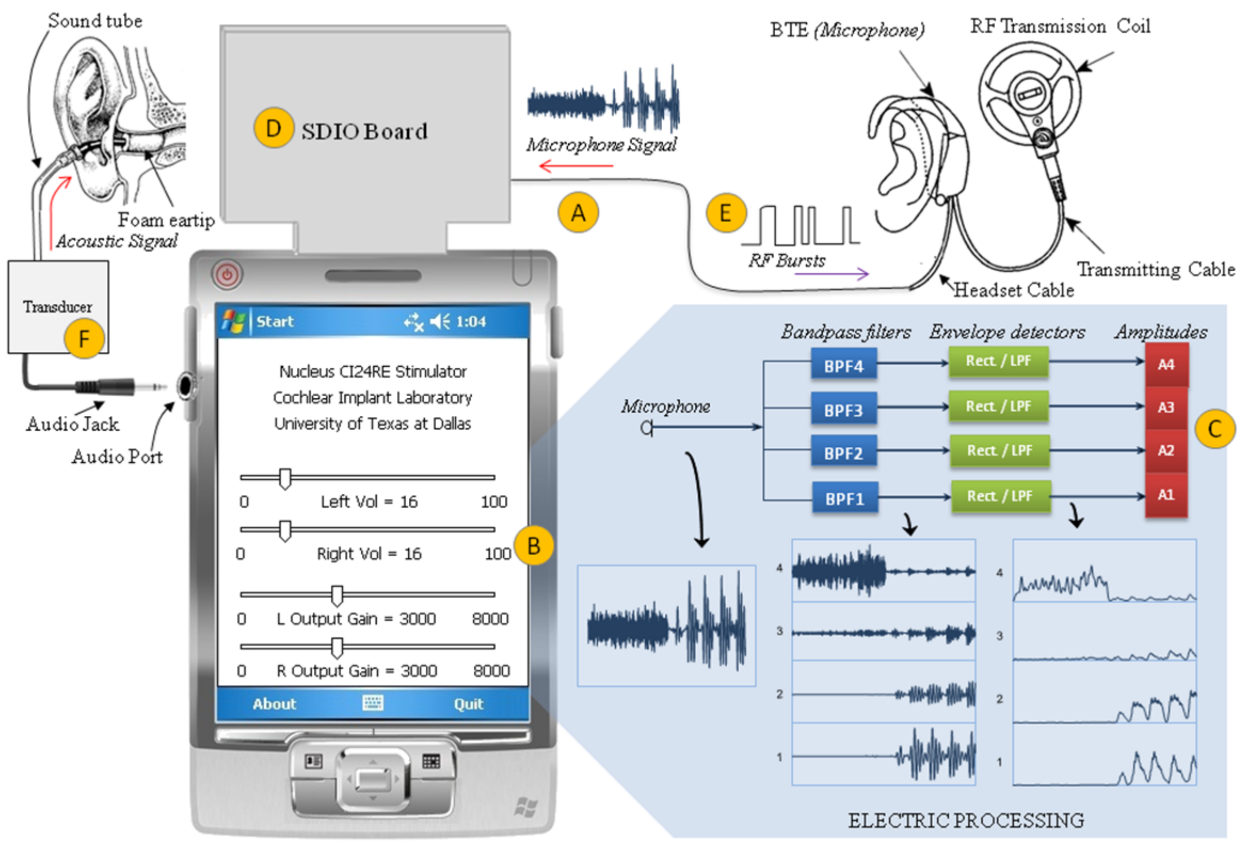


Figure 3.19. A schematic of the signal flow for the real-time speech processor. The acoustic signal is picked up by the microphone (A), sent (via the headset cable) to the SDIO interface board (D), which is then sampled and transmitted to the PDA. The PDA (B) processes the signal and generates a set (one for each channel of stimulation) of amplitudes (C). The example shows amplitudes generated for the CIS strategy while the platform supports both CIS and ACE strategies. These amplitudes are sent to the SDIO interface board (D), which are then coded for transmission to the cochlear implant in the form of RF bursts (E). At the same time, the processed audio buffer is sent to the tranducer (F) which presents the acoustic signal to the contralateral ear via the insert eartips. Both electric and acoustic stimuli are synchronized without any delay.

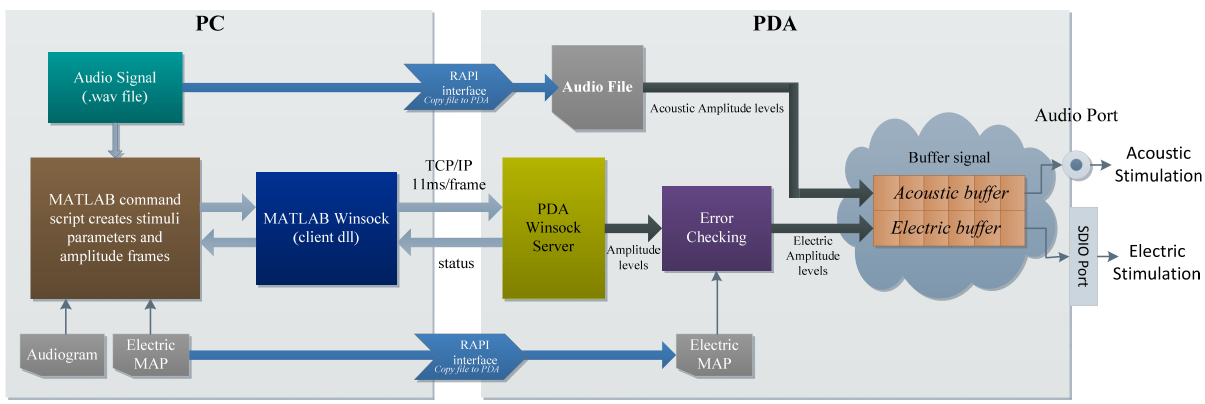


Figure 3.20. Software architecture of the offline speech processor for bimodal stimulation.

CHAPTER 4

EVALUATION WITH CI USERS

The PDA platform has undergone rigorous testing to ensure that it meets all safety criteria that of a clinical processor. In addition to this, the platform is FCC, IEC and FDA compliant. An IDE application for the PDA-based research processor was approved by the FDA in May 2011 for evaluation with human subjects. Since then, the platform has been tested with unilateral, bilateral and bimodal CI users. The results presented in this thesis are from an acute study (i.e., users were allowed to wear the processor for a few hours in the lab environment) on ten CI users. The aim of the current study was to evaluate the performance of the platform on a speech intelligibly task and compare the performance against the users' own clinical processor.

4.1 Subjects

A total of ten CI users participated in this chronic study. All participants were adults and native speakers of American English with post lingual deafness with a minimum of 1 year experience with cochlear implant(s) from Cochlear Ltd. There were five bimodal subjects (cochlear implant in one ear and hearing aid in the other). (one of the five bimodal subjects S5 was not available for some conditions). Of the remaining five subjects, one was unilateral (implant in one ear only) and rest four were bilateral CI.

4.2 Method

All subjects were tested with offline and real-time processors as well as their own clinical processor. The intelligibility scores from their own clinical processor were taken as benchmark scores for a fair comparison. Clinical processor and real-time processor evaluations were done in free-field in a sound booth at an average of 65dB SPL. (Speech stimuli for the offline processor are presented via audio files on the PC.) In all the cases, volume and gain adjustments were done on respective processors. For all the tests, subjects' everyday map was used. A short training with the PDA processor was carried out before each test.

In addition to electric-only (E), bimodal subjects were tested for acoustic-only (A) and electric+acoustic stimulation (EAS) with both types of processors. In the current study, no audio processing for acoustic stimulation was used.

4.3 Stimuli

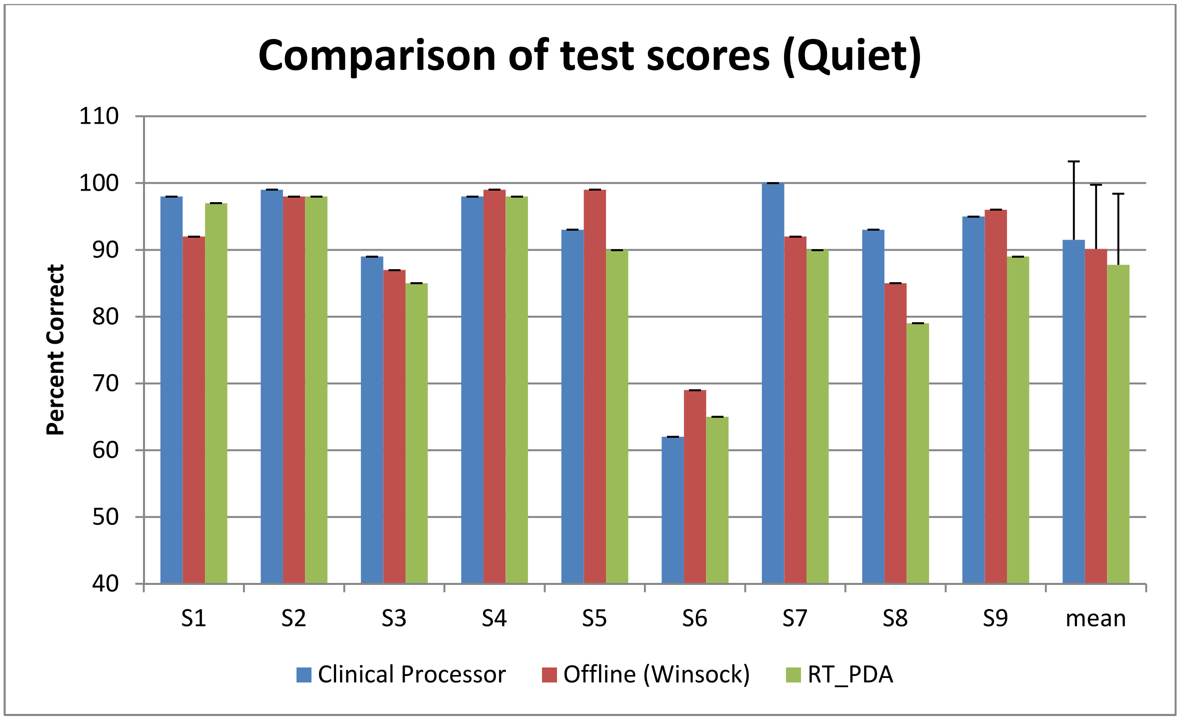
The speech stimuli used for testing were sentences from the Institute of Electrical and Electronics Engineers (IEEE) database (IEEE subcommittee 1969). Each sentence in the database is composed of approximately 7 – 12 words, and each list comprises of 10 sentences with an average of 80 words per list. Two lists for each test were used and the scores from the two were averaged. Three conditions were tested for each test, speech in quiet, speech in 10dB SNR and speech in 5dB SNR. Noise type used in all tests is speech shaped noise.

4.4 Electric-only Results

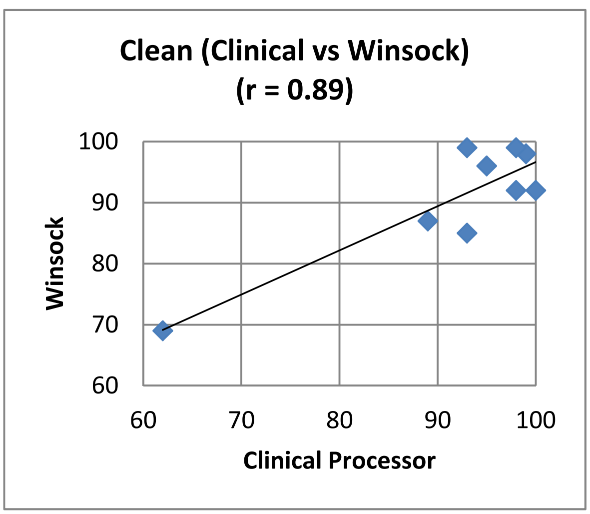
Figure 4.1, Figure 4.2 and Figure 4.3 show the percentage correct scores in: (i) quiet, (ii) 10dB and iii) 5dB SNR respectively. The results are reported for clinical processor, PDA-offline

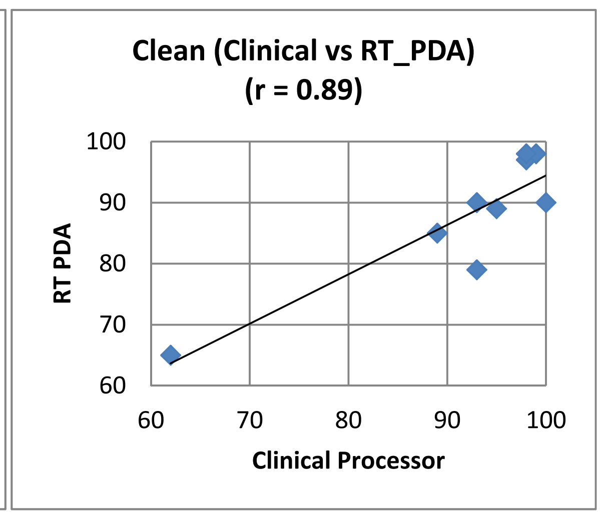
(Winsock) and PDA-Real Time (PDA-RT) processors. S1, S2, S3, and S4 are bilateral subjects while the rest (S5 – S9) are unilateral cochlear implant users. For easy comparison of test scores Figure 4.1 – 4.3 (b) and (c) shows scatter plots to depict the correlations between the PDA processors and the clinical processor in each of three test conditions (quiet, 10dB and 5dB SNR). Each scatter plot tries to fit in a best-fit line and gives the Pearson correlation coefficient 'r' for each test condition.

Clean and 10dB SNR conditions depict very strong correlations with both percentage correct scores within 10 percent window.



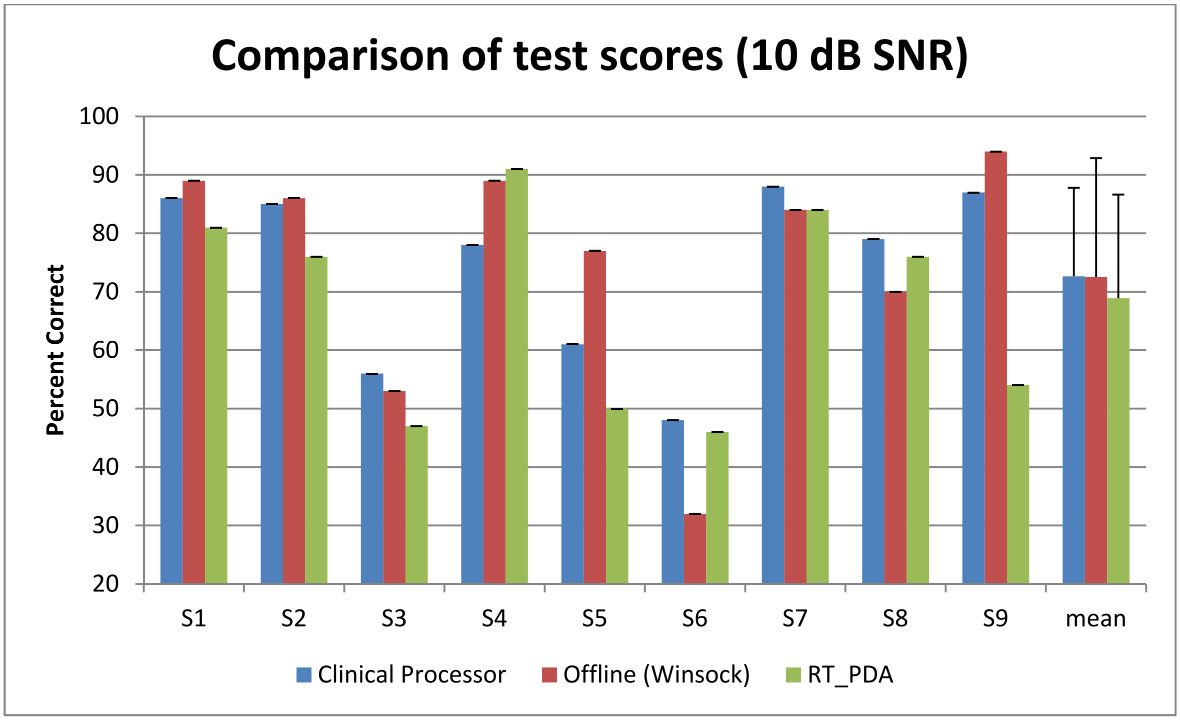
(a)



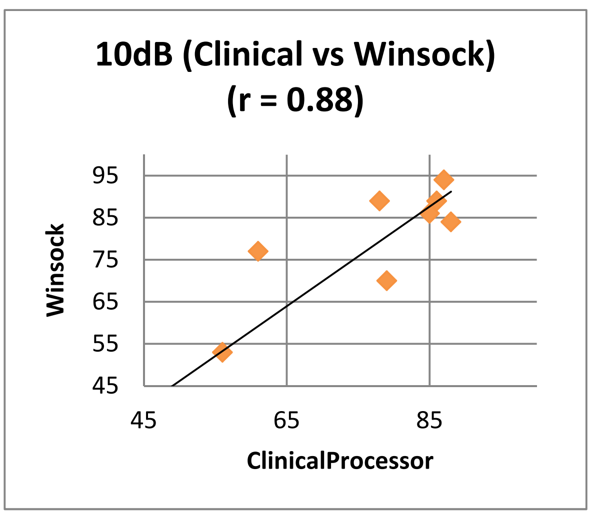


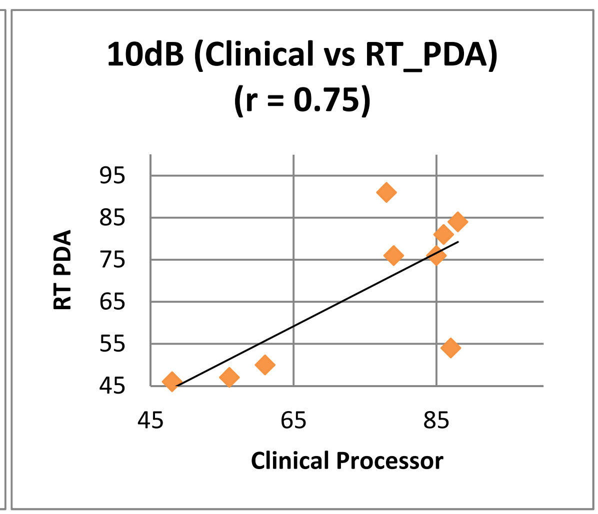
(b) (c)

Figure 4.1. Comparison of Electric Only Test Scores in Quiet.



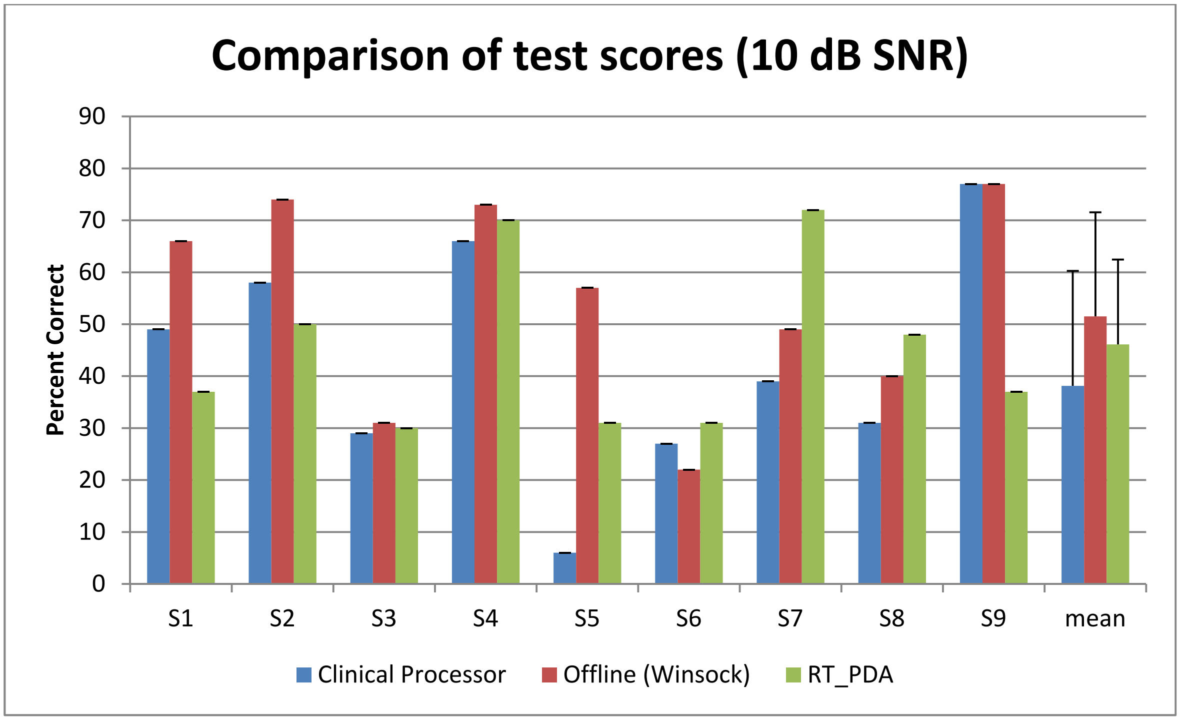
(a)



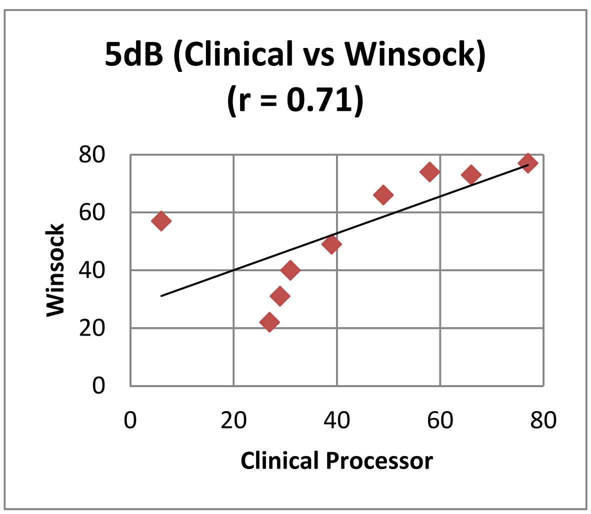


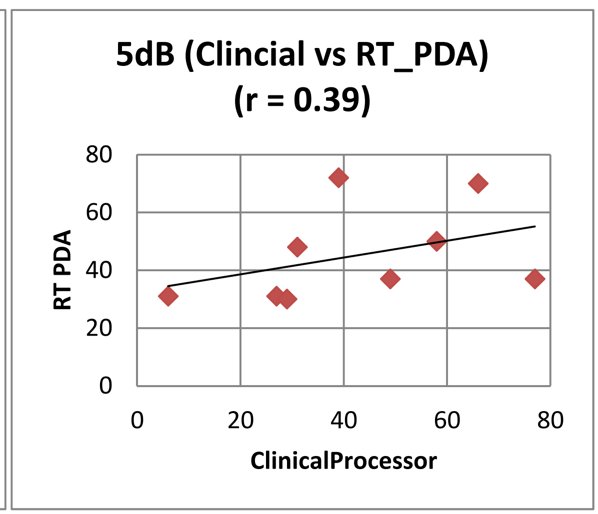
(b) (c)

Figure 4.2. Comparison of Electric Only Test Scores at 10dB SNR.



(a)





(b) (c)

Figure 4.3. Comparison of Electric Only Test Scores at 5dB SNR.

4.5 EAS Results

We studied a group of five subjects who had a fully inserted cochlear implant in one ear and who have low-frequency residual hearing in the other ear and thus wore hearing aid. In the non-implanted ear the mean thresholds at frequencies 500Hz or lower were 64dB HL and better. Thresholds at 1 KHz and above were 72 dB HL and poorer. Figure 4.4 displays the mean audiometric thresholds in the non-implanted ear. The mean thresholds at 0.25, 0.5, 0.75, 1.0, 2.0, 3.0, 4.0 and 6.0 KHz were 59, 64, 69, 72, 85, 85, 83, and 88 dB HL, respectively.

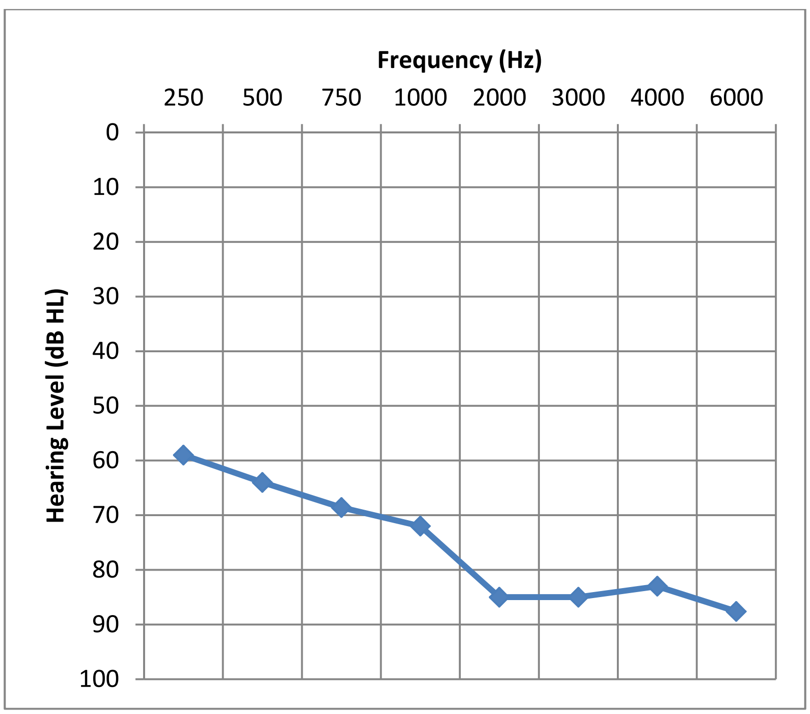
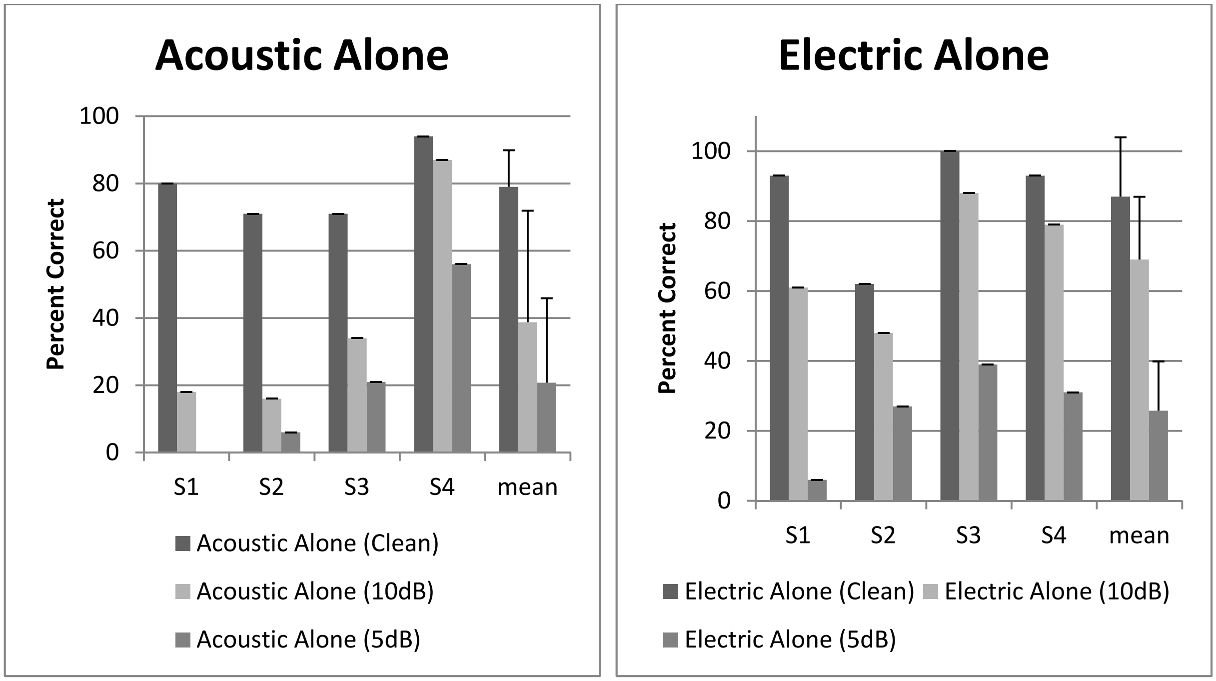


Figure 4.4. Mean Audiogram of five subjects.

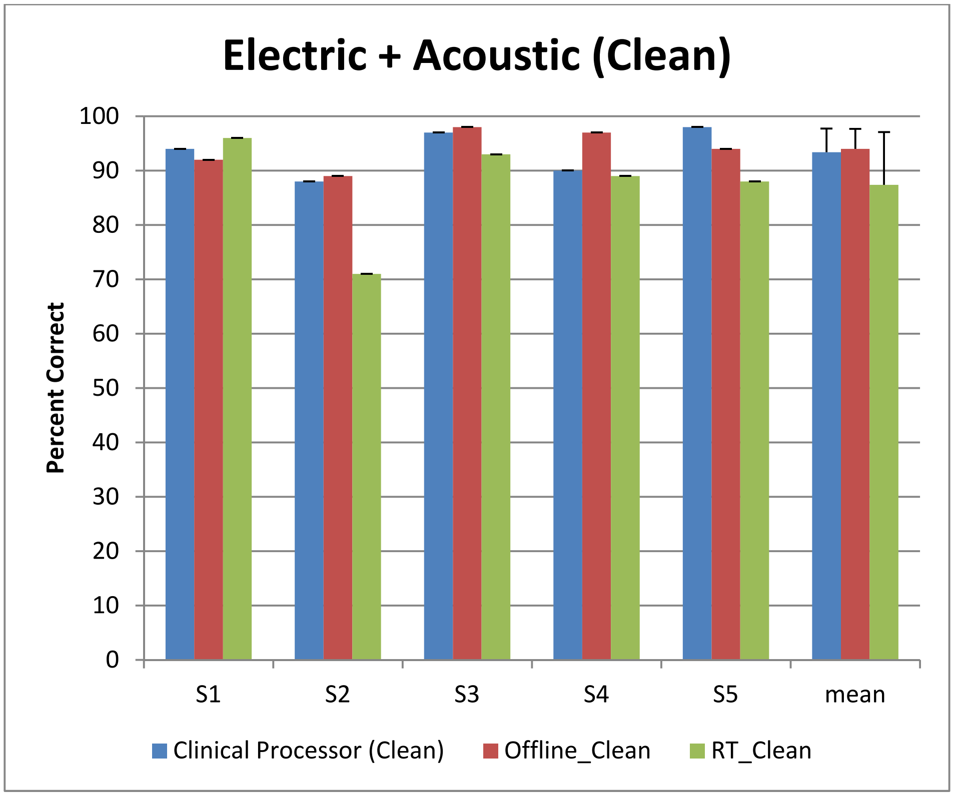
Figure 4.5 shows (a) acoustic-alone and (b) electric-alone scores for 4/5 subjects in quite, 10dB SNR and 5dB SNR. Subject S5 was not available for A and E.



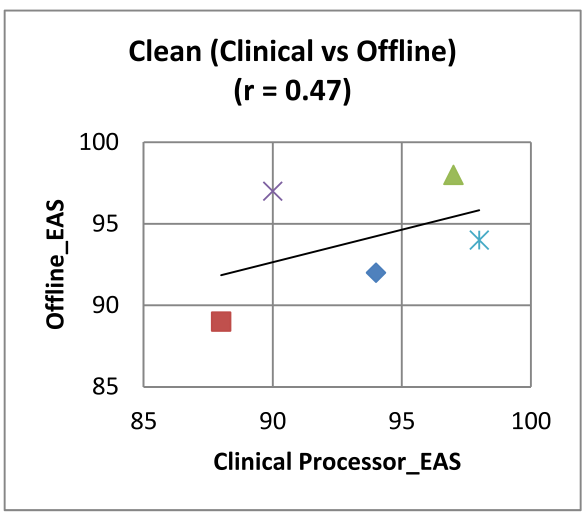
(a) (b)

Figure 4.5. Electric-alone (E) and acoustic-alone (A) scores of bimodal subjects.

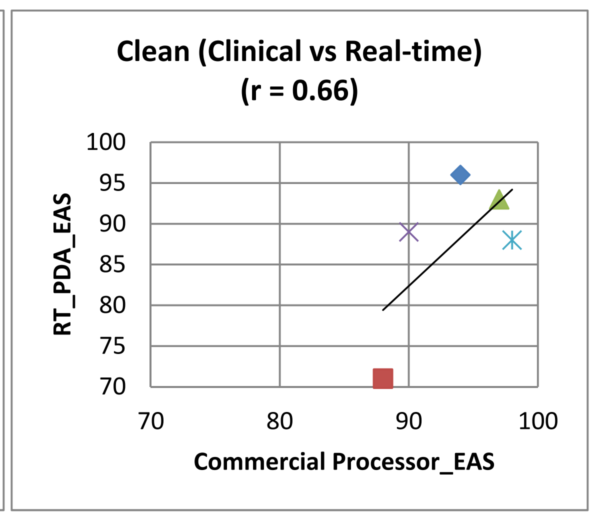
Figure 4.6, Figure 4.7, and Figure 4.8 show the percentage correct scores in: (i) quiet, ii) 10dB and (iii) 5dB SNR with combined electric and acoustic stimulation. The results are reported for clinical processor, PDA-offline and PDA-Real Time (PDA-RT) processors. Scatter plots to compare the performance of PDA processor versus clinical processor are also illustrated for each condition.



(a)

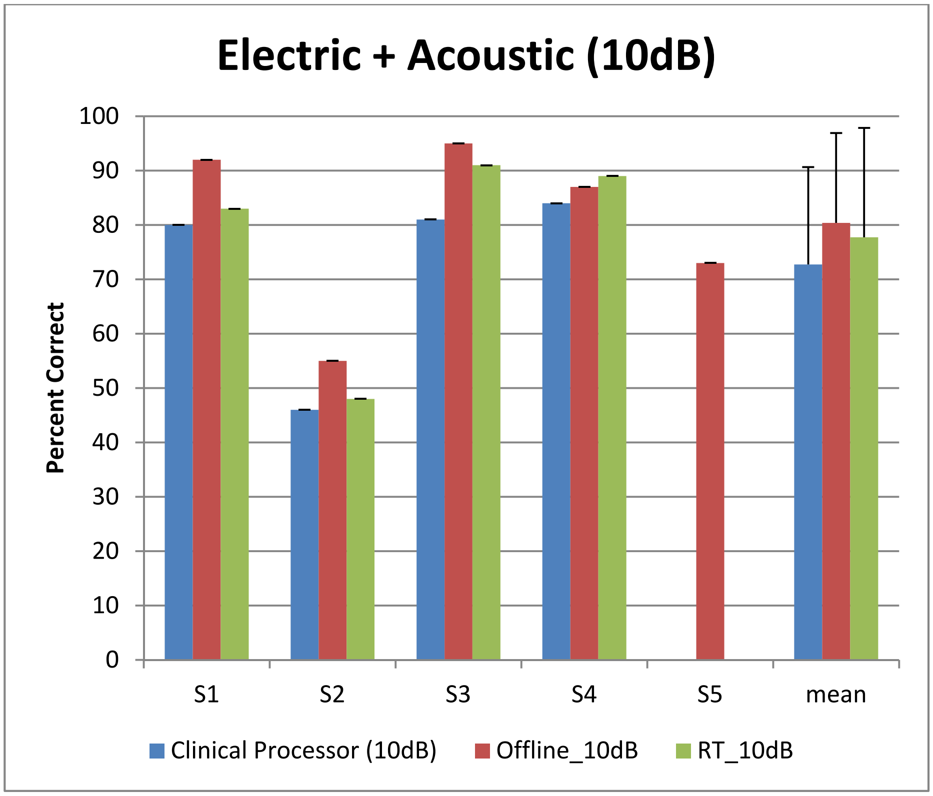


(b)

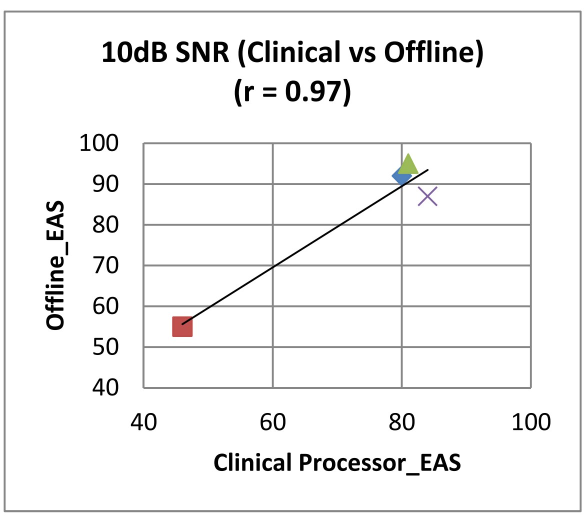


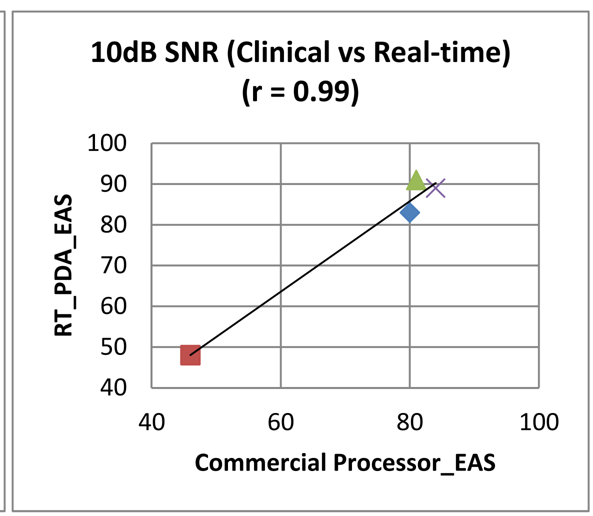
(c)

Figure 4.6. Comparison of EAS Test Scores in Quiet.



(a)





(b) (c)

Figure 4.7. Comparison of EAS Test Scores in 10dB SNR.

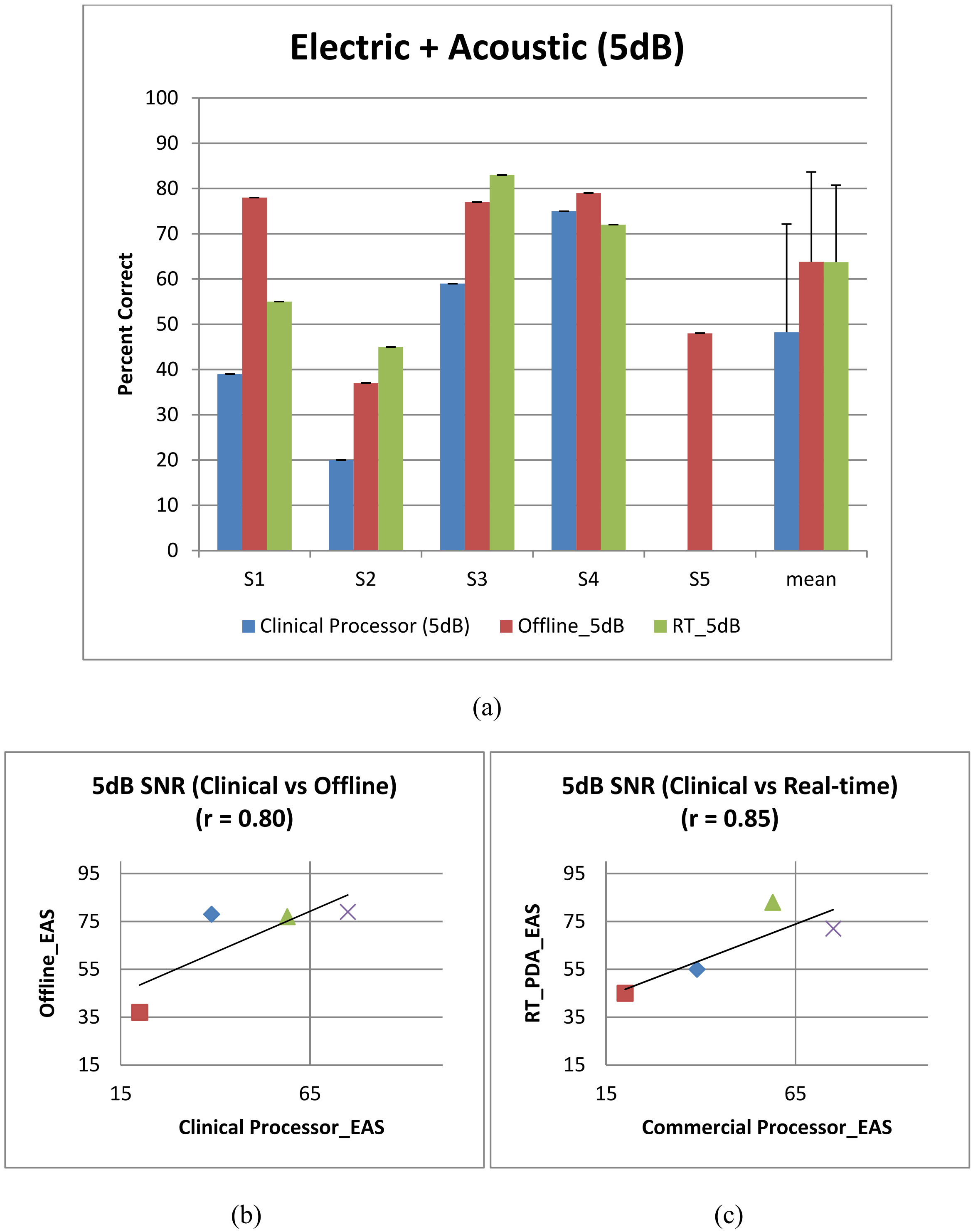


Figure 4.8. Comparison of EAS Test Scores in 5dB SNR.

EAS shows an improvement in scores as compared to A-only and E-only scores. This effect is more pronounced in noisy conditions. For example, percentage correct scores drastically improved from 21 percent with A-only to 60 percent with EAS. This is even greater than the sum of A and E alone. Significant improvement in noise is in line with studies published in (Turner, et al. 2004) and (Qin and Oxenham 2003). There is a strong correlation between all three processor types in all conditions. The Pearson's correlation coefficients for RT and clinical processor in 10dB and 5dB SNR were 0.99 and 0.85 respectively. For the offline processor in the same SNR, correlation coefficients were 0.97 and 0.80 respectively. These strong correlations suggest that the PDA platform delivers comparable performance with the commercial clinical processor.

CHAPTER 5

CONCLUSION

This thesis presented design of a PDA-based research platform which can be used to explore new ideas in cochlear research. Flexible design of the platform both in terms of hardware and software infrastructure allows for quick development and evaluation of new research ideas without having to learn advanced programming skills. The research processor works in two modes, real-time mode and offline mode. Real-time mode allows easy assessment of new algorithms and provides real-time feedback from the users. The programming environment for the real-time mode is C/C++ which is easy to program and hence saves considerable development time. Portability and wearability of the platform makes it possible for the real-time processor to be used outside the lab environment akin to a body-worn processor. This opens opportunities for long term evaluation of novel algorithms and strategies for chronic studies with human subjects, which was not possible with earlier generation of research interfaces.

For experiments which necessarily do not require real-time processing and stimulation, offline mode can provide an alternative by offering advantage of programming in much familiar MATLAB environment. MATLAB development environment is preferred by most researchers as it provides more robust and user-friendly control over the software routines. It is easier to program and make modifications in MATLAB code. Also debugging and testing code is extremely simple in MATLAB environment, thus, it saves plenty of development time which could be utilized in developing novel speech processing ideas.

The extension of the platform to include electric plus acoustic stimulation capabilities would allow researchers working in bimodal domain to take advantage of this innovative research interface. Bimodal mode can again be utilized in real-time or offline mode, thus giving same features and flexibility to researchers as with electric-only speech processor. EAS capability would help undertake research in emerging hybrid implants for patients with residual hearing in one or both their ears. Both electric and acoustic stimulations can be delivered binaurally which makes the platform generic and suitable for diverse applications.

The platform also supports capability to conduct psychophysics experiments in offline mode. The ability to control stimulation parameters and stimulation patterns of each electrode with time opens endless opportunities for researchers working in this domain to design and conduct novel experiments.

Finally, the successful evaluation of the platform with ten CI subjects with a population comprising of unilateral, bilateral and bimodal CI users proved the platform is comparable in performance in speech intelligibility task when compared against users' own clinical processer. The results are very encouraging for an acute study presented in this thesis. This has motivated us to undertake long-term clinical evaluation of the platform with take-home trials. Portability and wearability of the PDA platform makes it possible for the users to wear the platform on a daily basis until they fully adapt to the new processor. The possibility of conducting chronic studies with the PDA processor allows researchers to carry out long-term evaluation of novel coding algorithms and conduct experiments that would otherwise not n possible This in turn will open new possibilities in cochlear implant research and development.

My specific contributions to this thesis can be summarized as follows:

- *Realtime Software Development:* Develop and streamline software routines on existing framework. Test and debug the code and improve the algorithm implementation in C such that it is suitable for human testing. This involved meticulous attention to cochlear signal processing while taking care of fixed point implementation issues, and devising solutions for unexpected bugs and performance-deteriorating variables.
- *Offline Software Development:* Modify and redevelop the existing infrastructure for offline streaming to support frame by frame streaming of stimulation to the PDA server using Windows Sockets API. Devising a “clean” solution based on a single socclient dll as a backbone for the complete software infrastructure to stream the processed stimuli to the implant. Design software suites and range of applications for researchers to experiment with broad range of experimental tools. Develop applications for performing psychophysics experiments and a comprehensive mapping utility (like a commercial fitting software) to create subject maps in laboratory environment using psychophysics principles.
- *Bimodal Features:* Upgrade the existing electric-only platform to include electric plus acoustic stimulation for bimodal studies.
- *Evaluation with human subjects:* Clinical evaluation of the platform in realtime mode, offline mode and bimodal mode with human subjects.

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VITA

Hussnain Ali was born in Sialkot, Pakistan on January 11th, 1985. He did his undergraduate from National University of Science & Technology (NUST) and was awarded Bachelors in Electrical Engineering degree in April 2008. He worked in Center of Advanced Research in Engineering, one of the reputed research organizations in Islamabad, for one and a years where he undertook an innovative project on cardiac telemedicine system. He joined Electrical Engineering department at University of Texas of Dallas in January 2010 to undertake graduate studies. He has since been working in Cochlear Implant Laboratory on the PDA Platform project. His research interests are biomedical signal processing, implantable and wearable medical devices/systems and emerging healthcare technologies.