Clinical evaluation of an image-guided cochlear implant programming strategy

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Abstract

The cochlear implant (CI) has been labeled the most successful neural prosthesis. Despite this success, a significant number of CI recipients experience poor speech understanding, and, even among the best performers, restoration to normal auditory fidelity is rare. While significant research efforts have been devoted to improving stimulation strategies, few developments have led to significant hearing improvement over the past two decades. We have recently introduced image processing techniques that open a new direction for advancement in this field by making it possible, for the first time, to determine the position of implanted CI electrodes relative to the nerves they stimulate using CT images. In this article, we present results of an image-guided, patient-customized approach to stimulation that utilizes the electrode position information our image processing techniques provide. This approach allows us to identify electrodes that cause overlapping stimulation patterns and to deactivate them from a patient's map. This individualized mapping strategy yields significant improvement in speech understanding in both quiet and noise as well as improved spectral resolution in the 68 adult CI recipients studied to date. Our results indicate that image-guidance can improve hearing outcomes for many existing CI recipients without requiring additional surgery or the use of "experimental" stimulation strategies, hardware or software.

Introduction

Cochlear implants (CIs) are surgically implanted neural prosthetic devices used to treat severe-to-profound hearing loss [NIDCD, 2011]. To date, the CI has arguably been the most successful neural prosthesis. CIs use implanted electrodes to stimulate spiral ganglion (SG) nerves to induce hearing sensation (see Figure 1a-b). Implants available today yield remarkable results for the vast majority of recipients with average postoperative word and sentence recognition approximating 60% and 70% correct, respectively, for unilaterally implanted recipients and 70% and 80% correct for bilateral recipients [Buss et al., 2008; Dorman et al., 2009; Gifford et al. 2007; Gifford et al. 2013a; Litovsky et al. 2006]. Despite this success, a significant number of users receive marginal benefit, and restoration to normal fidelity is rare even among the best performers. This is due, in part, to several well-known issues with electrical stimulation that prevent CIs from accurately simulating natural acoustic hearing. Electrode interaction is an example of one such issue that, despite significant improvements made by advances in hardware and signal processing, remains challenging [Fu and Nogaki, 2005; Boex et al., 2003]. In natural hearing, a nerve pathway is activated when the characteristic frequency associated with that pathway is present in the incoming sound. Neural pathways are tonotopically ordered by decreasing characteristic frequency along the length of the cochlear duct, and this finely tuned spatial organization is well known (see Figure 1c) [Stakhovskaya et al., 2007]. CI electrode arrays are designed such that each electrode should stimulate nerve pathways corresponding to a pre-defined spectral bandwidth [Wilson and Dorman, 2008]. However, in surgery the array is blindly threaded into the cochlea with its insertion path guided only by the walls of the spiral-shaped intra-cochlear cavities. Since the final positions of the electrodes are generally unknown, the only option when programming has been to assume the electrodes are situated in the correct scala with a relatively uniform electrode-to-neuron interface across the array. Given this assumption, most implant recipients are programmed using a default frequency allocation table delivered across as many viable intracochlear electrodes as possible. Research has demonstrated, however, that many implanted arrays are not in the correct scala throughout the entire insertion depth [Finley et al.,

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2007; Holden et al., 2013; Skinner et al., 2007; Wanna et al. 2011] thus challenging the assumptions underlying a one-size-fits-all strategy.

Programming efficacy is sensitive to sub-optimal electrode positioning [Rubenstein, 2004; Wilson and Dorman, 2008], which can lead to excessive spread of intracochlear electric current, which is more commonly referred to as "channel interaction" [Fu and Nogaki, 2005; Boex et al., 2003]. Thus, it follows that more effective implant programming could result from consideration of individualized electrode position. This is particularly true if we are able to reduce or eliminate the deleterious effects of channel interaction, which most notably includes poor spectral resolution. Spectral resolution is associated with peripheral filtering achieved via both the bank of overlapping auditory filters positioned along the basilar membrane as well as the tonotopic organization of spiral ganglion (SG) neurons located within the modiolus. Impaired spectral resolution is known to result in significantly poorer speech recognition—particularly in background noise, as it results in a "smearing" of the speech and noise spectra (e.g., [Baer and Moore, 1994; Moore and Glasberg, 1993; Moore et al., 1995]). Cochlear implant recipients are known to have poor spectral resolution [Henry and Turner, 2003; Henry and Turner, 2005; Litvak et al., 2007; Saoji et al., 2007; Saoji et al. 2009; Spahr et al., 2011]. This is generally attributed to a number of factors, including that there is a discrete number of intracochlear electrodes, which limits the number of independent neural populations that can be stimulated (e.g., [Friesen et al., 2001]); the population of surviving SG cells is unknown; and electric current spreads widely in the cochlea (i.e. channel interaction).

We have recently introduced image processing techniques that make it possible, for the first time, to estimate the position of implanted CI electrodes relative to the SG nerves they stimulate in CT images [Noble et al., 2012]. Work from many other groups over the past decade has paved the way for in vivo electrode position analysis motivated by the importance of electrode location relative to the SG. For example, work has aimed to predict electrode insertion depth based on external cochlear dimensions such as basal turn diameter [Stakhovskaya et al. 2007] or the distance from the lateral wall to the round window [Escude et al. 2006]. Other groups have presented approaches for post-operatively imaging the location of implanted electrode arrays. For example, Verbist et al. 2005 and Aschendorff et al. 2005 optimized imaging protocols to permit better visualization of the electrodes in vivo. However, these techniques are still limited in that intra-cochlear structures are not well contrasted in CT, and the metallic electrodes further degrade tissue contrast due to image artifacts. Skinner et al. 2007 proposed an approach where intra-cochlear anatomy can be estimated in vivo in pre- and post-implantation CTs by rigidly aligning them with a high resolution histological atlas of a specimen. The limitations of this approach are that it requires time and expertise to manually register the datasets, and it does not account for non-rigid variations in intra-cochlear anatomy. Our approach is automatic, accurately estimates patient-specific intra-cochlear anatomy shape, and permits determining the location of the electrodes relative to intra-cochlear structures. This approach allows us to identify electrodes that provide overlapping stimulation patterns and to deactivate them from a patient's map [Noble et al., 2013]. In this article, we present results of experiments testing how this image-guided, patient-customized approach to stimulation, which we have termed Image-Guided Cochlear Implant Programming (IGCIP), affects outcomes. Given that our current IGCIP approach is to deactivate electrodes thought to be causing overlapping electrical excitation, the overarching hypothesis driving this research is that our IGCIP strategy would reduce channel interaction thereby improving spectral resolution. Should this strategy be effective at improving spectral resolution, we hypothesize that this would lead to corollary improvements in speech recognition, particularly in the presence of noise, and would further improve subjective perceptions of speech understanding abilities and overall sound quality.

The IGCIP methodology was originally proposed in [Noble et al., 2013]. That publication also presents results from preliminary tests with 11 subjects. The principal contribution of this article is the reporting of results of an expanded clinical study with 68 new subjects. With this more substantial dataset, it is now possible to draw stronger conclusions from statistical analysis about how IGCIP strategies affect hearing performance.

Materials and methods

In the following sections, we first present our approach for visualizing and analyzing the spatial relationship between the electrodes and the SG to facilitate the design of IGCIP strategies. Next, we introduce our proposed IGCIP strategy that uses this subject-specific spatial information to reduce interaction of electrode stimulation patterns. Finally, we present details about our experimental design.

Electrode position analysis

We use the image processing techniques we have recently presented to detect the position of the implanted electrodes relative to the SG nerves [Noble et al., 2013]. These techniques are designed to detect electrode position using pre- and post-implantation CT scans. An example result of this process for one of our study subjects in shown in Figure 1d. As can be seen in the figure, our software permits identifying the location of each contact in the array relative to the tonotopically mapped modiolus. Also shown in the figure are synthetically generated colored “stimulation fields” that show how the neural stimulation patterns from each electrode might be shaped and how they overlap among neighboring electrodes. Analysis of the spatial relationship between the electrodes and the spiral ganglion is necessary to extract programming-relevant information. Thus, to support the design of new image-guided programming strategies, we rely on a technique we developed in [Noble et al., 2013] for visualizing programming-relevant spatial information that we call electrode distance-vs.-frequency (DVF) curves. DVF curves are shown in Figure 1e for one of our study subjects whose electrode positions are shown in Figure 1d. In the plot, a DVF curve for each of the 16 electrodes is shown and colored similarly to the corresponding simulated current pattern shown in Figure 1d. The x-axis corresponds to SG Characteristic Frequency (CF) in log-scale and the y-axis corresponds to distance to the SG in millimeters. Each DVF curve shows the distance from the respective electrode to the closest regions of the SG, organized by CF. Each curve takes on a roughly parabolic shape with a minimum corresponding to the electrode’s closest

SG nerves (shown with a circle in the plot) and with tails that increase in distance for adjacent nerve groups that are further from the electrode. Given the nature of the spread of electrical current through tissue, the shape of each electrode's excitation pattern on the SG is inversely related to its distance from the SG neural populations. This is supported by electrical modeling simulations [Whiten et al. 07] as well as the pitch discrimination tests commonly performed clinically with CIs that show that electrodes that are closer to more apical SG regions generally create lower perceived pitches [Donaldson et al. 05]. Thus, using the DVF curves, not only is it easy to infer the region of the SG that a specific electrode will best stimulate, e.g., the nerve pathways with CFs around 10 kHz are closest to electrode 13; but also it is easy to detect when two electrodes stimulate the same region. For instance, a substantial portion of the DVF curve for electrode 7 falls above the DVF curve for electrode 8, which suggests that if both electrodes are active they will both stimulate many of the same neural populations, and hence create channel interaction.

Image-guided programming strategy

Since its introduction, continuous interleaved sampling (CIS) [Wilson et al., 1991] has been widely adopted, and all CI manufacturers today use CIS-based strategies [Rubenstein, 2004]. CIS uses non-simultaneous, interleaved pulses to decrease cross-electrode electric field channel interactions; however, this is implemented without precise knowledge of the relative location of the neural pathways and the electrodes. By integrating spatial information provided by our image-processing techniques, we can extend this concept to decrease electrode interactions at the neural level, i.e., reduce the cross-electrode neural stimulation site overlap. In our experiments, the reprogramming strategy is straightforward. We deactivate electrodes that are likely to cause stimulation overlap. One approach to reduce competing stimulation without image-guidance would be to drastically reduce the number of active electrodes, e.g., with only 1 active electrode there would be no competition. However, the tradeoff with reducing the number of active electrodes is that this reduces the already very limited number of spectral channels, further compressing the frequency spectrum. Thus, blindly deactivating enough electrodes to ensure reduced competition risks deactivating potentially useful electrodes, i.e., those with stimulation regions that have little competition, and this would also result in sub-optimal signal quality. With our approach, we assume that each electrode best stimulates SG sites that are closest to it (we call this the peak activation region), and we hypothesize that a better electrode configuration is one that consists of as many active electrodes as possible under the constraint that they all receive relatively little competition in their peak activation region. This approach allows for stimulation overlap at SG sites between electrodes, which is inevitable with a non-trivial number of electrodes, while ensuring that there is a subset of nerves that each electrode stimulates somewhat independently. Thus, we chose to keep active a maximal subset of electrodes that have DVF curves that do not substantially overlap around their minima. For the subject whose DVF curves are shown in Figure 1e, we chose to keep active electrodes 1, 3, 5, 8, 10, 11, 12, 13, and 14, resulting in a DVF plot shown in Figure 1f. As can be seen in the plot, each DVF curve with this reduced set of electrodes has a concave segment around its minima that is closer to the SG than any other electrode, which indicates that more independent stimulation patterns are achieved than with the traditional all-electrodes-on strategy. Conveniently, this approach does not conflict with existing signal processing strategies, and thus reprogramming does not require major processing changes.

In our experiments, after identified electrodes are deactivated, the sound spectrum is remapped to the remaining active electrodes and the stimulation speed adjusted to account for the deactivated electrodes using the CI manufacturer’s clinical software. The following section details our experimental design.

Experimental Design

Participants: Table 1 summarizes details about the study participants. Image-guided programming was completed for 72 ears in 68 post-lingually deafened CI users (22 bilateral, 46 unilateral). Prior to this study, each of these subjects had undergone several iterations of traditional programming adjustments and was considered by an expert audiologist to have achieved a stable map and hence the best hearing performance possible using the traditional behavioral programming approach. Length of CI use among subjects ranged from 0.5 to 14.7 years with an average of 2.9 years. The right-hand columns of the table contain results that will be discussed below. Informed consent was obtained from each participant in accordance with the study protocols approved by the Vanderbilt Institutional Review Board.

Experiment summary: For each participant, a battery of hearing and speech recognition tests was administered in up to three listening conditions: listening in the bilateral, best-aided condition; listening with the implanted ear being remapped alone; and, if the contralateral ear is also implanted, listening with the other implanted ear alone to serve as a control. Following baseline testing, the participant's CI was reprogrammed according to our image-guided programming strategy. So that performance with a new program could be measured in a semi-chronic condition, each subject returned for post-adjustment re-testing 3-6 weeks following the reprogramming. During this 3- to 6-week period, each subject was asked to live as they normally would; however, they were only provided with the new map so as to require compliance with full-time use of the experimental map. The difference between post- and pre-reprogramming hearing test results was used to quantify the benefit of our image-guided strategy. For bilateral recipients, we completed the reprogramming experiment on the poorer performing ear. Subjects 2, 4, and 37 had pre-implantation CT scans on both ears and were able to participate in the study for both CIs. For these subjects, we subsequently performed the experiment on the contralateral CI.

The assessment of speech recognition was accomplished with the adult minimum speech test battery, MSTB for adult cochlear implant recipients in the U.S. [MTSB, 2011]. Estimates of spectral resolution were obtained using a spectral modulation detection (SMD) task, which is a non-speech based hearing performance metric that provides a psychoacoustic estimate of spectral resolution, i.e., the ability of the auditory system to decompose a complex spectral stimulus into its individual frequency components [Saoji et al., 2009; Henry and Turner, 2003; Drennan et al., 2010, 2014; Gifford et al., 2014].

Finally, to measure performance qualitatively, participants completed the Abbreviated Profile of Hearing Aid Benefit (APHAB, [Cox and Alexander, 1995]) as well as the Speech, Spatial and Qualities of Hearing Scale (SSQ, [Gatehouse and Noble, 2004]). All speech and non-speech stimuli were presented at a calibrated presentation level of 60 dBA using a single loudspeaker presented at 0^{o} azimuth at a distance of 1 meter.

For participants with Advanced Bionics (AB) implants, prior to deactivating the selected electrodes in the clinical software, we removed Fidelity120 processing (i.e. current steering) to allow selective deactivation of electrodes without deleting a viable electrode in the "pair." For individuals making use of ClearVoice prior to study enrollment, removal of Fidelity120 required deactivation of ClearVoice, as well. ClearVoice is described by AB as a signal or speech enhancement strategy. Though details of ClearVoice are guarded by proprietary restraint, it has been described as an algorithm designed to estimate the signal-to-noise (SNR) level in each channel, and subsequently, for those channels in which noise and/or poor SNR is identified, channel gain is reduced. We kept the stimulation strategy consistent with respect to paired or sequential stimulation. Further, pulse width was manipulated within +/- 7 microseconds to keep channel stimulation rate consistent with the participant's clinical map. In some cases, this required switching from automatic pulse width (APW) in the SoundWave software to manually determined pulse width. We were careful to ensure that the resultant channel stimulation rate was within 400 pps of the participant's beginning rate with Fidelity120. Stimulation rates for all AB recipients remained above 1500 pps even after removing Fidelity120. For participants with MED-EL implants, we kept the participant's strategy consistent (High Definition CIS (HDCIS) or Fine Structure Processing (FSP)) and fixed the stimulation rate manually to that which was used in the patient's own map. All participants with Cochlear Corporation implants used Advanced Combination Encoder (ACE) programming strategy. For these participants, after deactivating the selected electrodes, if the participant had 12 or fewer active electrodes in their MAP, we set the maxima equivalent to the number of active electrodes thereby converting to a CIS program. For participants with more than 12 active electrodes in their experimental MAP, we kept the maxima consistent with what was used in the patient's own MAP which ranged from 8 to 12 maxima. Channel stimulation rate and pulse width were unaltered. For all implant recipients, regardless of implant manufacturer, manipulation of M/C or T levels of individual electrodes was not performed. For patients reporting significantly 'softer' or less frequently 'louder' programs following selective electrode deactivation, we would globally increase or decrease M/C levels to the participant's desired overall volume.

Study data were collected and managed using the REDCap (Research Electronic Data Capture) secure data managements tools hosted at Vanderbilt [Harris et al., 2009].

Hearing aid verification: The hearing aid (HA) settings for the contralateral ear of unilaterally implanted participants with residual acoustic hearing in the non-implanted ear were verified prior to each test session. Hearing aids were verified for all subjects using probe microphone measurements to NAL-NL2 [Dillon, 2006; Keidser et al., 2011] target audibility for 60-dB-SPL speech. In cases where settings were undershooting NAL-NL2 target audibility, the participants' own HA was reprogrammed. In cases for which the participants' own HA was not adjustable due to either lack of reserve gain or incompatibility with NOAH programming software, a clinic stock HA was programmed and used for testing purposes. This occurred for one subject (20) for which a Phonak Naida S V UP behind-the-ear hearing aid was programmed, with SoundRecover deactivated, and affixed the participant's own fitted earmold. The aided speech intelligibility index (SII) values for 60-dB-SPL speech and the unaided low-frequency pure tone average (125, 250, and 500 Hz), in dB HL, are provided in Table 1 for the nineteen bimodal participants in the current study.

Speech Recognition: Speech recognition was assessed as recommended by the revised minimum speech test battery (MSTB, 2011) for adult CI recipients. The MSTB outlines the administration of Consonant Nucleus Consonant (CNC, [Peterson and Lehiste, 1962]) monosyllabic words and AzBio sentences [Spahr et al., 2012] in quiet and noise. In addition to CNC words and AzBio sentences, we also assessed speech recognition in pseudoadaptive noise with the Bamford-Kowal-Bench Speech-in-Noise (BKB-SIN, [Bench et al., 1979; Etymotic Research, 2005; Killion et al., 2004]) test. As compared to CNC and AzBio which are scored in terms of percent correct, the BKB-SIN metric provides a score corresponding the signal-to-noise ratio (SNR) at which the listener would achieve approximately 50% correct performance. This is reported as the SNR-50.

Participants scoring 50% or higher for AzBio sentences in quiet were also tested at +10 dB signal-to-noise ratio (SNR) using a continuous, multi-talker background noise. Similarly, participants scoring 50% or higher for AzBio sentences at +10 dB SNR, were also tested at +5 dB SNR using the same continuous multi-talker babble. All speech tests were administered to each implanted ear independently as well as in the bilateral-aided condition whether that included bilateral implants (CI + CI) or bimodal hearing (CI + HA). For participants with bimodal hearing, the CI only condition was assessed with the contralateral ear occluded via foam earplug.

Assessment of Spectral resolution: Spectral modulation detection: Spectral resolution was assessed via spectral modulation detection (SMD). The quick SMD task [Gifford et al., 2014] used in the current study included a 3-interval, forced-choice procedure to contrast flat-spectrum noises with spectrally modulated noises. Spectral modulation was achieved by applying logarithmically spaced, sinusoidal modulation to the broadband carrier stimulus. The carrier stimulus had a bandwidth of 125-5600 Hz [Gifford et al., 2014]. SMD was assessed in the current study using a procedure based on a modified method of constant stimuli [Fechner, 1966; Gescheider, 1997]. There were six trials presented for each of the five modulation depths (10, 11, 13, 14, and 16 dB) and frequency (0.5 and 1.0 cyc/oct) for a total of 60 trials. Each trial was scored as correct or incorrect and spectral resolution is described as the overall percent correct score for the task (chance = 1/3).

Perceived hearing handicap and quality of life: The APHAB provides a global estimate of the percentage of problems associated with listening in a variety of listening conditions and assesses aided benefit in four subscales including ease of communication, background noise, reverberation and aversiveness. As such, lower scores on APHAB indicate fewer problems and better perceived benefit. The SSQ employs a visual analog scale which gauges hearing ability across listening domains including speech understanding in various listening conditions, spatial hearing associated with distance, movement and direction, and the overall quality of speech including clarity and naturalness of sound. Higher scores on this metric are correlated with better speech understanding, spatial hearing, and sound quality.

Screening for cognitive impairment. To enable detection of cognitive function related effects, we administered the mini-mental state examination (MMSE, [Folstein et al., 1975]) screening tool at the time of study enrollment. The MMSE is a validated screening tool of cognitive function and includes tests of orientation, attention, memory, language and visual-spatial skills. It is generally accepted that MMSE scores under 25 indicate impaired cognition.

Results

In Figure 2, the boxplots show the distributions across subjects of raw benefit computed as pre-adjustment scores subtracted from post-adjustment scores. For each measure, benefit is shown for the adjusted ear alone (blue, left boxplot), bilateral (green, middle boxplot), and control ear alone (magenta, right boxplot) listening conditions. In each boxplot, the red line indicates the median, the box indicates the 25th and 75th percentiles, red pluses indicate outliers, and the whiskers indicate the range of non-outlier data. Below each boxplot, the number of scores in the dataset (N) is shown and scores that are statistically significant as measured by the two-tailed Wilcoxon signed rank test [Wilcoxon, 1945] at $p<0.05$ are indicated with a green “W.” The dataset size, N, differs from plot to plot since test materials were not returned by some participants and not all measures were tested for every subject. The BKB-SIN measure is scored in terms of dB, and units for benefit in dB for this measure are shown on the right. The remaining measures (CNC & AzBio) are scored in terms of percent correct, and these units are shown on the left. As seen in the figures, results of the pre- and post-adjustment tests performed on the unadjusted control ear alone show, on average, little change. In contrast, the group average test results in the adjusted and bilateral conditions for several measures improved substantially and were statistically significant for several measures. Measures scored in percent correct are sensitive to range saturation effects when scores are closer to 0% or 100%. When viewing raw benefit alone, range saturation effects can confound significant decline or improvement, especially in the bilateral listening condition where many participants already have relatively high scores with use of the contra-lateral ear. Thus, in Figure 3, we show the same data for these measures in terms of percent benefit rather than raw benefit to account for ceiling and floor effects. Percent benefit is computed by normalizing the benefit or decline in each score with respect to the maximum possible benefit or decline. Thus, in this group, scores of 100%, 0%, and -100% represent a change from the pre-adjustment score to 100% correct, the same score, and 0% correct, respectively. We detect significant percent benefit for all quantitative measures of speech recognition in quiet and noise when tested in the bilateral condition. In contrast, we detect significant raw benefit as shown in Figure 2 in the bilateral condition only for measures of speech in noise. This highlights the potential confound of benefit compression due to ceiling effects that occurs for the tests in quiet that are relatively easier and where raw pre-remapping scores are generally higher. Improvement in raw benefit for three of five measures when testing with the re-mapped ear alone are statistically significant, and improvement in percent benefit for three of four measures when testing in the same listening condition are also statistically significant.

Mean SMD scores are plotted in Figure 4. SMD scores are shown, in percent correct, for each of the 5 spectral modulation depths (in dB) as well as the mean across all depths for 0.5 and 1.0 cycle/octave for the remapped ear alone. A two-way, repeated measures analysis of variance was completed with modulation depth and time point (pre vs post adjustment) as the independent variables and the SMD score as the dependent variable. For 0.5 cyc/oct, statistical analysis revealed a significant effect of modulation depth [$F_{(4, 65)} = 49.9$, $p < 0.001$], a significant effect of time point [$F_{(4, 65)} = 6.53$, $p = 0.013$], and an interaction [$F_{(4, 65)} = 3.9$, $p = 0.004$]. Post hoc testing using an all pairwise multiple comparison procedure (Holm-Sidak method) revealed a significant effect of time point for the two shallowest modulation depths of 10 dB ($t = 2.5$, $p = 0.013$) and 11 dB ($t = 3.9$, $p < 0.001$). For 1.0 cyc/oct. statistical analysis revealed a significant effect of modulation depth [$F_{(4, 65)} = 44.2$, $p < 0.001$], no effect of time point [$F_{(4, 65)} = 0.48$, $p = 0.49$], yet a significant interaction [$F_{(4, 65)} = 2.6$, $p = 0.037$]. Post hoc testing using an all pairwise multiple comparison procedure (Holm-Sidak method) revealed a significant effect of time point for the shallowest modulation depth of 10 dB ($t = 2.6$, $p = 0.01$). Significant improvement in SMD is itself a substantial finding, as few developments in strategy in the past 20 years have been shown to significantly improve spectral resolution for CI recipients [Drennan et al., 2010]. More noteworthy is that the improvement in SMD was observed for the most shallow modulation depths (10 and 11 dB) which are the most challenging conditions with respect to spectral envelope perception.

Benefit in qualitative metrics (APHAB and SSQ) is shown in Figure 5. Improvements in five of seven individual components to these tests were statistically significant as were overall scores, indicating significant subjective preference to the experimental map on average.

Results for the adjusted ear of each subject are also shown as line plots in Figure 5 and are color-coded by hearing test. The left and right endpoints of each line plot indicate the pre- and post-remapping scores for the corresponding measure. Lines with positive slope indicate scores that improved from pre- to post-adjustment. While some participants experienced decreases in hearing performance scores, the majority (64%) of the line slopes are positive. Using critical difference tables generated for the CNC words [Thornton and Raffin, 1978] and AzBio sentences [Spahr et al., 2012] using a binomial distribution statistic for individual speech perception metrics, we can detect significant differences between pre- and post-adjustment scores on an individual basis for these measures. Further, for the BKB-SIN measure, Etymotic Research has published the critical difference value for adult CI recipients tested with 1 list pair being 4.4 dB based on the 95% confidence interval. Below the plots, scores that significantly improve and decline are indicated with green stars and blue circles. Measures that were not tested are indicated by a black "x." There were *statistically significant improvements at the individual level* in one or more measure for 39 of 72 ears.

The right columns of Table 1 indicate, for each experiment, whether the subject kept the experimental map and the subject's MMSE and Aided SII scores. The far right column shows the number of electrodes that were deactivated clinically, prior to this study, and the number of electrodes that our experimental techniques recommended for deactivation. As seen in the table, participants requested changes or did not elect to keep their experimental maps for only 16 out of 72 ears (22%). The remaining participants/ears (78%) elected to keep the experimental MAPs with many strongly opposed to returning to their old maps. Many subjects who did not exhibit an individually significant benefit for speech or SMD testing reported improved sound quality and this is reflected here in the 78% rate of experimental map retention. Comments from individuals who participated in this study immediately after reprogramming and at the end of the study are included online as a supplemental tables 1 & 2.

V. Discussion

The results of our image-guided CI programming tests show that our approach leads to significantly improved speech recognition, spectral resolution, and subjective hearing quality, and thus could improve hearing quality-of-life, for many CI recipients. Statistically significant improvement was noted for the majority of the quantitative metrics tested. For the best-aided, bilateral condition, statistically significant raw improvement was noted for metrics that include noise, and statistically significant percent improvement was noted for all metrics. We believe this difference is due to the fact that improvements in metrics that are measured in quiet are masked in the raw scores due to range saturation effects for many of the subjects because they already had relatively high scores with the use of their better performing contralateral ear. The quality of overall change is best reflected in the APHAB and SSQ scores. Improvement in performance in background noise is especially significant considering that speech recognition in noise is one of the most common problems even among the best performing CI users [Fu and Nogaki, 2005]. It is also of note that these results generally agree with the preliminary results reported in [Noble et al., 2013], suggesting that results with our technique hold and are determined to be statistically significant when applied to a larger, relatively heterogeneous population.

It is clear that the information provided by our image-guidance techniques is critical for the improvements in hearing performance seen in our results since electrode deactivation performed using other criteria has been studied by many groups without significantly affecting performance. For example, some groups have experimented with deactivating different numbers of electrodes in regular patterns and found little effect on average speech recognition performance as long as more than 4-8 electrodes are active [Fri01a, Gar02]; and other groups deactivated electrodes based on psychoacoustics criteria, resulting in increases in certain speech recognition measures and decreases in others [Zwo97, Gar12]. Our results clearly demonstrate the impact this IGCIP strategy can have on even long time CI users.

Since the reprogramming strategy we use only requires deactivating electrodes, it is simple to integrate with existing sound processing strategies, such as CIS, using the existing clinical software provided by CI manufacturers. Typically when changes to a program are made, quantitative and qualitative hearing scores tend to favor the original program [Tyler et al., 1986]. Thus, it is remarkable that the majority of the subjects in our experiments noted substantial improvement in sound quality immediately after re-programming, and these improvements are reflected in our quantitative results. It is likely that long-term experience with the new program will result in further improvements in hearing performance. According to the NIDCD, over 200,000 people have received CIs as of 2010 [NIDCD, 2011]. We hypothesize that our electrode deactivation strategy could improve hearing in many of these CI users, thus improving their communication abilities and hearing-related quality of life, without requiring additional surgical procedures. Our results show that our personalized IGCIP approach to programming can improve spectral resolution and speech recognition in quiet and noise. However, the electrode deactivation strategy we present exploits only a small fraction of the programming relevant information captured by our image processing and analysis techniques. Thus, we believe that the strategy tested in this study is just the first of many new and significant IGCIP stimulation strategies that will be developed now that analysis of the spatial relationship between electrodes and stimulation targets is possible.

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References

Aschendorff A, Kubalek R, Turowski B, Zanella F, Hochmuth A, Schumacher M, Klenzner T, Laszig R. Quality control after cochlear implant surgery by means of rotational tomography. Otol Neurotol. 2005;26:34–37.

Baer T, Moore BC: Effects of spectral smearing on the intelligibility of sentences in the presence of interfering speech. J Acoust Soc Am 1994;95:2277-2280.

Bench J., Kowal A., Bamford J., “The BKB (Bamford-Kowal-Bench) sentences lists for partially-hearing children,” Br. J. Audiol. 13: 108-12, 1979.

Boex C, de Balthasar C., Kos M.I., Pelizzone M., “Electrical field interactions in different cochlear implant systems,” J Acoust Soc Am 114:2049–2057, 2003.

Buss E, Pillsbury HC, Buchman CA, Pillsbury CH, Clark MS, Haynes DS, Labadie RF, Amberg S, Roland PS, Kruger P, Novak MA, Wirth JA, Black JM, Peters R, Lake J, Wackym PA, Firszt JB, Wilson BS, Lawson DT, Schatzer R, S. DHP, Barco AL: Multicenter u.S. Bilateral med-el cochlear implantation study: Speech perception over the first year of use. Ear Hear 2008;29:20-32.

“Cochlear Implants,” National Institute on Deafness and Other Communication Disorders, No. 11-4798, 2011.

Cootes, T.F., Taylor, C.J., Cooper, D.H., Graham, J, “Active Shape Models—Their Training and Application,” Comp. Vis. Image Unders. 61(1), 39-59, 1995.

Cox R.M., Alexander G.C., “The abbreviated profile of hearing aid benefit,” Ear Hear. 16(2): 176-86, 1995.

Dice, L.R., “Measures of the amount of ecologic association between species,” Ecology 26, 297–302, 1945.

Dillon H: What’s new from nal in hearing aid prescriptions? The Hearing Journal 2006;59:10-16.

Donaldson, G.S., Kreft, H.A., Litvak, L., 2005. Place-pitch discrimination of single-versus dual-electrode stimuli by cochlear implant users (L). J. Acoust. Soc. Am. 118, 623-626.

Dorman MF, Yost W, Wilson BS, Gifford RH: Speech perception and sound localization by adults with bilateral cochlear implants. Seminars in Hearing 2009;32:73-89.

Drennan W.R., Won J.H., Nie K., Jameyson E., Rubinstein J.T., “Sensitivity of psychophysical measures to signal processor modifications in cochlear implant users,” Hear Res 262: 1-8, 2010.

Etymotic Research: Bamford-kowal-bench speech-in-noise test (version 1.03) [audio cd]. 2005.

Fechner GT: Elemente der psychophysik. Breitkopf & härtel, leipzig (reprinted in 1964 by bonset, amsterdam); english translation by he adler (1966): . New York, Holt, Rinehart & Winston, 1860, 1966.

Finley CC, Holden TA, Holden LK, Whiting BR, Chole RA, Neely GJ, Hullar TE, Skinner MW: Role of electrode placement as a contributor to variability in cochlear implant outcomes. Otol Neurotol 2007;29:920-928.

Folstein MF, Folstein SE, McHugh PR: Mini-mental state. A practical method for grading the cognitive state of patients for the clinician. J Psychiatr Res 1975;12:189-198.

Friesen L.M., Shannon R.V., Baskent D., Wang X., “Speech recognition in noise as a function of the number of spectral channels: Comparison of acoustic hearing and cochlear implants,” J. Acoust Soc Am. 110(2): 1150-63, 2001.

Fu Q.J., Nogaki G., “Noise susceptibility of cochlear implant users: The role of spectral resolution and smearing,” J Assoc Res Otolaryngol. 6(1):19–27, 2005.

Gatehouse S, Noble W: The speech, spatial and qualities of hearing scale (ssq). Int J Audiol 2004;43:85-99.

Garadat S.N., Zwolan T.A., Pfingst B.E., “Across-site patterns of modulation detection: Relation to speech recognition,” J. Acoust. Soc. Am. 131(5): 4030-41, 2012.

Garnham C., O’Driscoll M., Ramsden R., Saeed S., “Speech understanding in noise with a Med-El COMBI 40+ cochlear implant using reduced channel sets,” Ear Hear. 23(6): 540-52, 2002.

Gescheider G: Chapter 3: The classical psychophysical methods: Psychophysics: The fundamentals (3rd edition). New York, Lawrence Erlbaum Associates, 1997.

Gifford RH, Shallop JK, Peterson AM., “Speech Recognition Materials and Ceiling Effects: Considerations for Cochlear Implant Programs,” Audiol Neurotol, 13:193-205, 2008.

Gifford RH, Dorman MF, Sheffield SW, Teece K, Olund AP: Availability of binaural cues for bilateral implant recipients and bimodal listeners with and without preserved hearing in the implanted ear. Audiology & Neurotology in review, 2013.

Gifford RH, Hedley-Williams, A, Spahr, AJ. Clinical assessment of spectral modulation detection for cochlear implant recipients: a non-language based measure of performance outcomes. Audiology & Neurotology in press, 2013

Green, B. F., “The orthogonal approximation of an oblique structure in factor analysis,” Psychometrika 17, 429-440, 1952.

Paul A. Harris, Robert Taylor, Robert Thielke, Jonathon Payne, Nathaniel Gonzalez, Jose G. Conde, Research electronic data capture (REDCap) - A metadata-driven methodology and workflow process for providing translational research informatics support, J Biomed Inform. 2009 Apr;42(2):377-81.

Henry B.A., Turner C.W., “The resolution of complex spectral patterns by cochlear implant and normal-hearing listeners,” J Acoust Soc Am 113(5):2861-73, 2003.

Henry BA, Turner CW, Behrens A: Spectral peak resolution and speech recognition in quiet: Normal hearing, hearing impaired, and cochlear implant listeners. J Acoust Soc Am 2005;118:1111-1121.

Holden LK, Finley CC, Firszt JB, Holden TA, Brenner C, Potts LG, Gotter BD, Vanderhoof SS, Mispagel K, Heydebrand G, Skinner MW: Factors affecting open-set word recognition in adults with cochlear implants. Ear Hear 2013;34:342-360.

Keidser G, Dillon H, Flax M, Ching TY, Brewer S: The nal-nl2 prescription procedure. . Audiology Research 2011;1:88-90.

Killion MC, Niquette PA, Gudmundsen GI, Revit LJ, Banerjee S: Development of a quick speech-in-noise test for measuring signal-to-noise ratio loss in normal-hearing and hearing-impaired listeners. . J Acoust Soc Am 2004;116.

Litovsky RY, Parkinson A, Arcaroli J, Sammeth C: Simultaneous bilateral cochlear implantation in adults: A multicenter clinical study. Ear Hear 2006;27:714-730.

Litvak LM, Spahr AJ, Saoji A, Fridman GY: Relationship between perception of spectral ripple and speech recognition in cochlear implant and vocoder listeners. J Acoust Soc Am 2007;122:982-991.

Moore BCJ, Glasberg BR: Simulation of the effects of loudness recruitment and threshold elevation on the intelligibility of speech in quiet and in a background of speech. J Acoust Soc Am 1993;94:2050-2062.

Moore BC, Glasberg BR, Vickers DA: Simulation of the effects of loudness recruitment on the intelligibility of speech in noise. Br J Audiol 1995;29:131-143.

(MSTB) MSTB: The new minimum speech test battery. http://auditorypotential.com/MSTB.html, 2011.

Noble J.H., Labadie R.F., Majdani O., Dawant B.M., “Automatic segmentation of intra-cochlear anatomy in conventional CT”, IEEE Trans. on Biomedical. Eng., 58(9): 2625-32, 2011.

Noble, J.H., Schuman, A., Wright, C.G., Labadie, R.F., Dawant, B.M., "Automatic Identification of Cochlear Implant Electrode Arrays for Post-Operative Assessment," Proc. of the SPIE conf. on Med. Imag., 7962: 796217, 2011.

Noble, J.H., Gifford, R.H., Labadie, R.F., Dawant, B.M., “Statistical Shape Model Segmentation and Frequency Mapping of Cochlear Implant Stimulation Targets in CT,” N. Ayache et al. (Eds.): MICCAI 2012, Part II, LNCS 7511, pp. 421-428. 2012.

Noble JH, Labadie RF, Gifford RH, Dawant BM, “Image-guidance enables new methods for customizing cochlear implant stimulation strategies,” IEEE Trans. on Neural Systems and Rehabilitiation Engineering, vol. 21(5):820-9, 2013.

Peterson G.E., Lehiste I., “Revised CNC lists for audiotory tests,” J. Speech and Hear. Dis. 27: 62-70, 1962.

Rogers M., Graham J., “Robust Active Shape Model Search,” ECCV 2002, LNCS, vol. 2353, pp. 517-53, 2002.

Rubenstein J.T., “How cochlear implants encode speech,” Curr Opin Otolaryngol Head Neck Surg. 12(5): 444-8, 2004.

Saoji AA, Eddins DA: Spectral modulation masking patterns reveal tuning to spectral envelope frequency. J Acoust Soc Am 2007;122:1004-1013.

Saoji A.A., Litvak L.M., Spahr A.J., Eddins D.A., "Spectral modulation detection and vowel and consonant identifications in cochlear implant listeners," J. Acoust. Soc. Am. 126 (3): 955-8, 2009.

Skinner MW, Holden TA, Whiting BR, Voie AH, Brundsen B, Neely GJ, Saxon EA, Hullar TE, Finley CC: In vivo estimates of the position of advanced bionics electrode arrays in the human cochlea. Annals of Otology, Rhinology and Laryngology Supplement 2007;197:2-24.

Spahr AJ, Saoji A, Litvak LM, Dorman MF: Spectral cues for understanding speech in quiet and in noise. Cochlear Implants International 2011;12:S66-S69.

Spahr A.J., Dorman M.F., Litvak L.M., Van Wie S., Gifford R.H., Loizou P.C., Loiselle L.M., Oakes T., Cook S., "Development and validation of the AzBio sentence lists," Ear Hear. 33(1): 112-7, 2012.

Stakhovskaya O., Spridhar D., Bonham B.H., Leake P.A., "Frequency Map for the Human Cochlear Spiral Ganglion: Implications for Cochlear Implants," J. Assoc. Res. Otol. 8: 220–33, 2007.

Thornton, A. R., Raffin, M. J., "Speech-discrimination scores modeled as a binomial variable," J Speech Hear Res 21, 507–518, 1978.

Tyler R.S., Preece J.P., Lansing C.R., Otto S.R., Gantz B.J., "Previous experience as a confounding factor in comparing cochlear-implant processing schemes," J. Speech Hear. Res. 29: 282-7, 1986.

Verbist, B.M., Frijns, J.H.M., Geleijns, J., van Buchem, M.A., "Multisection CT as a Valuable Tool in the Postoperative Assessment of Cochlear Implant Patients," AJNR Am J Neuroradiol 26:424–429, February 2005.

Wanna GB, Noble JH, McRacken TR, Dawant BM, Dietrich MS, Watkins LD, Schuman TA, Labadie RF: Assessment of electrode placement and audiologic outcomes in bilateral cochlear implantation. Otol Neurotol 2011;32:428-432.

Whiten D, "Electro-anatomical models of the cochlear implant," Ph.D. Thesis, Massachusetts Institute of Technology Library, 2007.

Wilcoxon, F., "Individual comparisons by ranking methods," Biometrics Bulletin 1 (6): 80–83, 1945.

Wilson B.S., Finley C.C., Lawson, D.T., Wolford, R.D., Eddington, D.K., Rabinowitz, W.M., "Better speech recognition with cochlear implants," Nature **352**, 236-238, 1991.

Wilson B.S., Dorman M.F., "Cochlear implants: Current designs and future possibilities," J. Rehab. Res. Dev. 45(5): 695-730, 2008.

Zijdenbos A.P., Dawant B.M., Margolin R., "Morphometric Analysis of White Matter Lesions in MR Images: Method and Validation," IEEE Trans. on Med. Imag. 13(4), 716-724, 1994.

Zwolan T.A., Collins L.M., Wakefiled G.H., "Electrode discrimination and speech recognition in postlingually deafened adult cochlear implant subjects," J. Acoust. Soc. Am. 102(6): 3673-85, 1997.

Figure Legend

Figure 1. Spatial analysis of an implanted subject. The scala tympani (red) and scala vestibuli (blue), the two principal cavities of the cochlea, are shown in (A-C). In (B), also shown is a rendering of the auditory nerve fibers leading to the SG in green. In (C-D), the AR (the surface representing the interface between the nerves of the SG and the intra-cochlear cavities) is colorcoded with the tonotopic place frequencies of

the SG in Hz. Also shown in (D) are the implanted electrodes of the CI, numbered 1-16. An illustration of current spread from each electrode is rendered transparently with different colors between neighboring electrodes. Electrode distance-vs.-frequency curves are plotted in (E) and (F).

Figure 2. Box plots of the raw benefit between post- and pre-adjustment condition testing of each hearing performance measure. Shown are the median (red line) and the range of the 75th to 25th percentile, i.e., Q_3 to Q_1 (box). Whiskers extend to data points that lie within the range $Q_3 + 1.5(Q_3 - Q_1)$ to $Q_1 - 1.5(Q_3 - Q_1)$. Outlier points that lie beyond the whiskers are shown as red pluses. Below each plot, a 'W' indicates that the measure is statistically significant. Also shown is the dataset size (N). For each measure, benefit is shown for the adjusted ear alone (blue, left boxplot), bilateral (green, middle boxplot), and control ear alone (magenta, right boxplot) listening conditions.

Figure 3. Box plots of the percent benefit between post- and pre-adjustment condition testing of each hearing performance measure. Plot information is defined identically to Figure 2.

Figure 4. Bar plots of mean SMD scores measured on the remapped ear alone. Scores are shown in percent correct for each of the 5 spectral modulation depths (in dB) as well as the mean across all depths for 0.5 and 1.0 cycle/octave. A two-way, repeated measures analysis of variance was completed with modulation depth and time point (pre vs post adjustment) as the independent variables and the SMD score as the dependent variable. Statistically significant results are indicated by asterisks.

Figure 5. Box plots of the qualitative benefit between post- and pre-adjustment condition measured by APHAB and SSQ. Plot information is defined identically to Figure 2.

Figure 6. Individual hearing performance results for the adjusted ear of each subject shown as line plots. The left and right ends of each line plot show pre- and post-adjustment results for the indicated subject

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and hearing performance measure, respectively. Below the plots, scores that significantly improve and decline are indicated with green stars and blue circles. Measures that were not tested are indicated by a black “x.”

Table 1. This table contains information about our experiments. For each participant, shown are whether (Y) or not (N) they have bilateral CIs, age in years, mini mental state examination (MMSE) score, and aided speech intelligibility index (SII) score. For each experiment, shown are the adjusted ear, left (L) or right (R); length of prior use of the adjusted CI in years; the manufacturer of the adjusted CI, Advanced Bionics (AB), Cochlear (CO), or Med El (ME); whether the subject elected to keep the experimental map (green ‘Y’) or not (blue ‘N’); the number of electrodes that were deactivated in the clinical map; and the number of electrodes that were deactivated in the experimental map.

Supplemental Table 1. This table contains comments from several study participants that were made immediately after re-mapping.

Supplemental Table 2. This table contains comments from several study participants that were made at the end of the study.

Figure 1

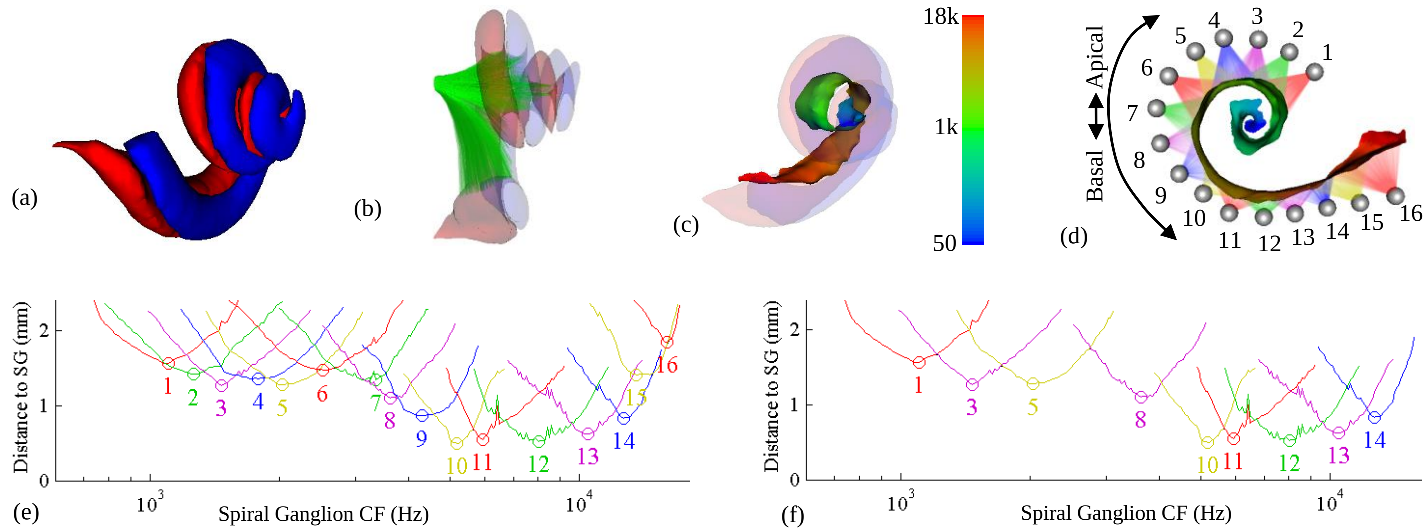
(a)
(b)
(c)
18k
1k
50
(d)
Apical
Basal
1
2
3
4
5
6
7
8
9
10
11
12
13
14
15
16
(e)
Distance to SG (mm)
Spiral Ganglion CF (Hz)
(f)
Distance to SG (mm)
Spiral Ganglion CF (Hz)

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Figure 2

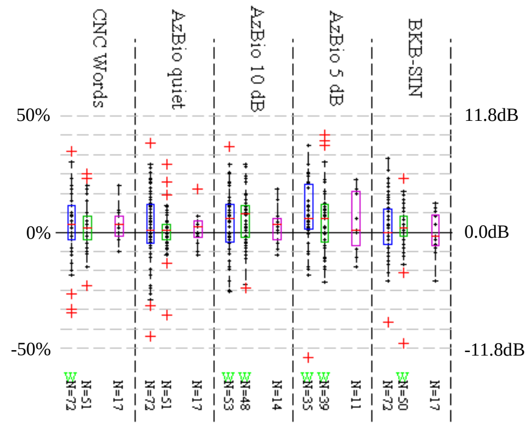

Figure 3

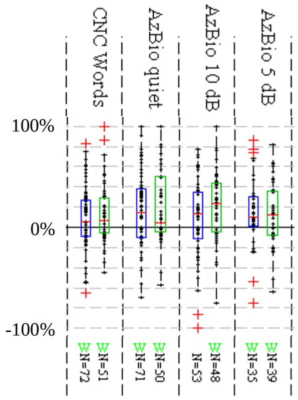
CNC Words
AzBio quiet
AzBio 10 dB
AzBio 5 dB
100%
0%
-100%
N=72
N=51
N=71
N=50
N=53
N=48
N=35
N=39

Figure 4

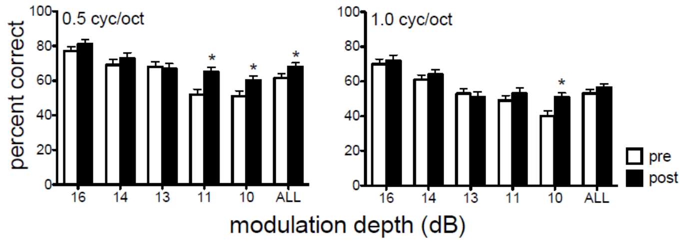
0.5 cyc/oct
1.0 cyc/oct
percent correct
0
20
40
60
80
100
16
14
13
11
10
ALL
*
pre
post
modulation depth (dB)

Figure 5

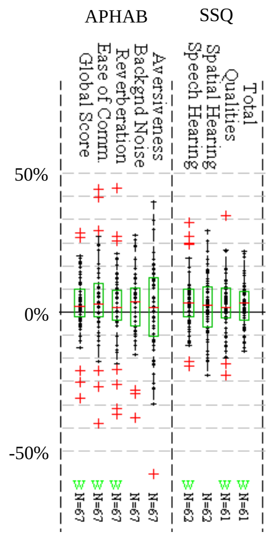
APHAB
SSQ
Global Score
Ease of Comm.
Reverberation
Backgnd Noise
Aversiveness
Speech Hearing
Spatial Hearing
Qualities
Total
50%
0%
-50%
W W W
N=67 N=67 N=67 N=67 N=67
W W W
N=62 N=62 N=61 N=61

Figure 6

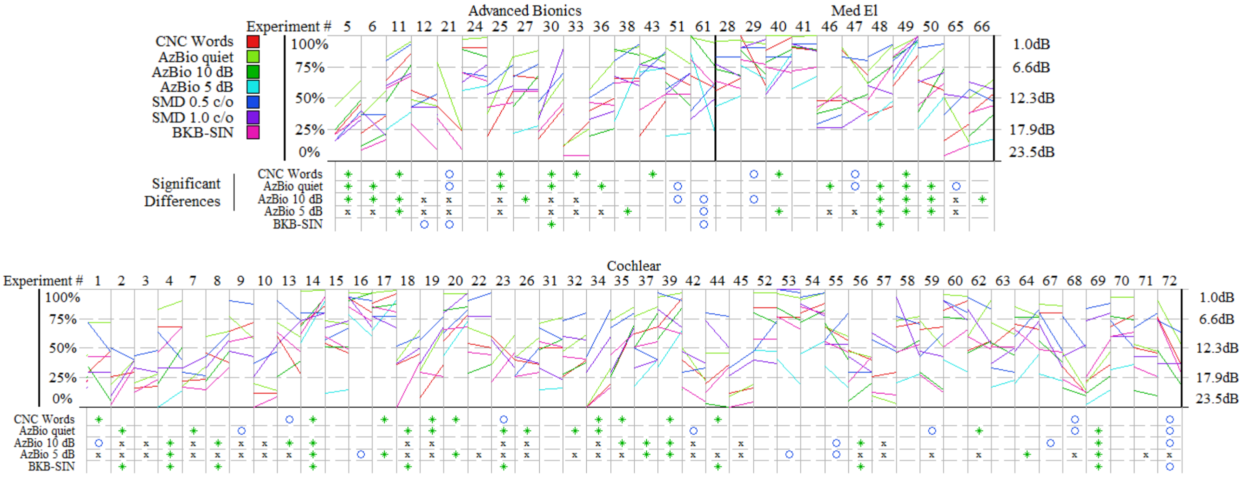
Advanced Bionics
Med El
Experiment # 5 6 11 12 21 24 25 27 30 33 36 38 43 51 61 28 29 40 41 46 47 48 49 50 65 66
CNC Words
AzBio quiet
AzBio 10 dB
AzBio 5 dB
SMD 0.5 c/o
SMD 1.0 c/o
BKB-SIN
100%
75%
50%
25%
0%
1.0dB
6.6dB
12.3dB
17.9dB
23.5dB
Significant Differences
Cochlear
Experiment # 1 2 3 4 7 8 9 10 13 14 15 16 17 18 19 20 22 23 26 31 32 34 35 37 39 42 44 45 52 53 54 55 56 57 58 59 60 62 63 64 67 68 69 70 71 72

Table 1

Experimen	Subject #	Bilateral Cis	Adjusted Ear	Age (yrs)	CI Use (yrs)	CI Brand	Kept Ma	MMSE	Aided SII	# clinical contacts	# experimental contact
1	1	No	R	46	1.27	CO	Y			20	13
2	2	Yes	R	76	1.15	CO	Y	24		22	16
3	2	Yes	L	76	1.53	CO	Y			22	16
4	3	No	R	69	2.14	CO	Y	30	23	20	19
7	5	No	L	83	0.66	CO	Y	27	15	19	10
8	6	No	L	65	2.16	CO	Y	28	32	17	11
9	7	Yes	L	80	1.22	CO	N	29		20	18
10	8	No	L	70	3.03	CO	Y	28		22	16
13	11	No	R	86	1.27	CO	N	30		22	15
14	12	Yes	L	55	3.43	CO	Y	30		22	15
15	13	No	R	53	2.45	CO	Y	30	25	22	11
16	14	Yes	R	62	7.58	CO	Y	30		22	17
17	15	Yes	L	62	6.17	CO	Y	30		20	15
18	16	No	R	75	0.63	CO	Y	30		22	11
19	17	No	L	53	0.69	CO	Y	28		13	7
20	18	Yes	R	50	1.37	CO	Y	30		22	14
22	20	No	L	69	0.67	CO	Y	29	27	21	14
23	21	No	R	82	0.98	CO	N	30		22	15
26	24	No	R	78	0.9	CO	N			22	12
31	28	No	R	68	1.32	CO	Y	29	42	22	15
32	29	No	R	91	1.27	CO	Y	29	9	22	12
34	31	No	L	84	2.84	CO	Y	23		20	15
35	32	Yes	L	58	1.24	CO	Y	29		17	6
37	34	No	R	74	0.86	CO	Y	28	40	22	13
39	36	No	R	69	1.35	CO	Y	27	30	22	17
42	38	No	R	62	8.59	CO	N	29		20	15
44	40	No	L	83	1.53	CO	Y	30		22	17
45	41	No	R	79	2.13	CO	Y	26		22	12
52	48	No	R	62	3	CO	Y	29	39	22	12
53	49	Yes	L	47	5.22	CO	N	30		20	12
54	50	No	L	55	1.4	CO	Y	29		19	14
55	51	Yes	R	74	2.01	CO	Y	27		22	17
56	52	No	R	84	1.1	CO	Y		56	22	14
57	53	Yes	R	54	2.37	CO	N	25		22	12
58	54	Yes	R	63	4.51	CO	Y	28		22	14
59	55	Yes	L	87	5.45	CO	N	29		22	12
60	56	No	R	66	1.13	CO	Y	29		22	12

Experiment	Subject	Bilateral CIs	Adjusted Ear	Age (yrs)	CI Use (yrs)	CI Brand	Kept Map	MMSE	Aided SII	# clinical contacts	# experimental contacts
62	58	Yes	L	53	0.61	CO	Y	29		20	18
63	59	No	L	79	0.53	CO	N	30	24	21	15
64	60	Yes	R	78	11.91	CO	Y	26		20	11
67	63	No	L	60	1.32	CO	N	30	16	22	10
68	64	No	R	55	0.81	CO	N	29	29	22	10
69	65	No	R	77	1.83	CO	Y	28		20	9
70	66	Yes	R	39	5.58	CO	Y	30		20	11
71	67	No	L	73	1.27	CO	Y	30	37	11	14
72	68	Yes	L	70	3.1	CO	N	26		21	12
28	26	No	L	43	0.75	ME	Y	30		12	9
29	27	No	R	66	0.67	ME	N	30		12	10
40	37	Yes	L	59	2.53	ME	Y	30		12	11
41	37	Yes	R	59	4.04	ME	Y			11	10
46	42	No	R	64	0.86	ME	Y	28		11	8
47	43	No	R	76	11.2	ME	Y			11	9
48	44	No	R	81	0.92	ME	Y	28		11	9
49	45	No	L	49	0.56	ME	Y	30		12	9
50	46	No	R	64	1.4	ME	Y	29	11	11	8
65	61	Yes	R	77	4.89	ME	Y	28		8	6
66	62	Yes	L	64	6.43	ME	Y	30		12	10
5	4	Yes	R	44	5.51	AB	Y	30		16	11
6	4	Yes	L	44	5.51	AB	Y			16	14
11	9	No	R	42	5.35	AB	Y	30		16	9
12	10	No	R	23	14.68	AB	Y	30		16	14
21	19	No	R	65	0.69	AB	N	28		16	10
24	22	No	R	68	5.8	AB	Y	30	20	16	12
25	23	No	L	49	0.76	AB	Y	28		15	7
27	25	No	R	77	12.68	AB	Y	30		16	12
30	27	Yes	R	66	0.51	AB	Y			16	9
33	30	No	R	52	0.65	AB	Y	30		16	11
36	33	No	R	84	1.58	AB	Y	29	56	15	14
38	35	Yes	R	46	5.93	AB	Y	30		16	10
43	39	No	R	60	0.84	AB	Y	30		16	10
51	47	No	R	59	4.27	AB	N		22	16	8
61	57	No	L	54	0.82	AB	N	30		16	11

Online supplemental table 1:

Exp. #	Comments from subjects immediately after re-programming
2	My wife's voice sounds a lot clearer.
3	That sounds really good.
5	I can already tell I'm going to like this.
6	That's a lot clearer now.
7	This is the best my implant has ever sounded.
11	It sounds less cluttered.
12	Sounds better
14	After 2 days with the experimental map: I have spent some 'quality time' over the weekend listening to my iPod and so far a big thumbs up. I want to give it a couple of more weeks for my brain to acclimate to the different simulation but my current observations are very positive in all sound situations. If most people are able to attain similar results in their first mapping this would be a paradigm shift in their overall hearing therapy. I bet those who give up early on accepting mediocre performance would be best served by this procedure. As you know, with both CI's I put (my audiologist) through hell (51 total maps) before I felt we dialed in the best performance; this procedure improved upon that!. I am certain many of these iterations were simply due to me scripting my brain to this new input and then deciding it could sound better. I can't imagine how much improved my acclimation to my first CI would have been if the physiology interface had been honed in from day one!
15	Sounds a little clearer, but I will have to give it a try outside where it is noisier.
18	Now that sounds good
19	It's definitely different but I will try it!
20	Sounds a lot less mechanical.
22	Clearer. I hear a lot more.
24	It is not as tinny. I think that's a good thing, but it's going to take some time to make a decision.
25	This is a little clearer; it sounds more natural; my voice is more natural; I can already tell this is better.
26	There is no more 'wamp wamp' sound. I like it. I think I can hear better.
27	This room has very poor acoustics and I am hearing you just fine. I wasn't able to do that before the new map.
28	I can already tell I'm hearing better.
29	I'll try it. Can't really tell a difference right now.
30	Sounds better but it's going to take some time to adjust.
31	This sounds good.
32	Sounds clearer. I think I hear better.
34	My voice sounds natural.
35	I can tell immediately that this is better.
36	sounds a little louder
39	speech is clearer and crisper
40	I don't know if this is a placebo effect or not but it does really sound clearer.
41	Sounds crisper.
42	I will have to get used to it.
43	Its different. I will have to get used to it.
44	It still has an echo but I'll try it.'
45	This is much clearer. I don't sound like I am in a barrel.
46	I could hear you when you turned your head.
48	It makes a boing sound but I will try it.
49	Sounds like Charlie Brown's teacher - womp, womp, womp
50	It sounds pretty good. I am looking forward to the next month of listening.
52	I was actually able to understand you when not facing you. Your voice tone changed, but there is more clarity. The quality...I'm very encouraged by the quality. Understanding you was not as difficult.
53	It sounds distorted.
54	Not as assaulting as before. Sounds are crisper and clearer. Your voice is smoother. I can hear you much better. I could sit here all day listening to this new sound. This was well worth the 8 hour drive to get here.
55	I think we've made some improvement. I'm hearing really good right now.
56	It's louder and a little more distinctive.
57	I think it sounds good. Your voice sounds rugged.
58	It has an emphasis on the high frequencies. My old program sounds more natural.
59	That's fine. I can hear better than when I cam in here. I hear better now.
60	It sounds good. I sound like a female, but you sound better. I don't have the reverberation in my voice like I used to and I sound more distinct.
62	I like it. That's a lot better.
63	Sounds good.
64	It's sharper, which is good, I think. It has a little bit of an echo.
65	It's a lot clearer.
66	This sounds a lot clearer.
67	It's a little more mechanical and it doesn't sound as good as it was before. But, I can understand you better in background noise than I would have with my old program.
68	This is better. There is no more roar.

Online supplemental table 2:

Exp. #	Comments from subjects at end of study
6	Everything is so much clearer. It's like I don't have a 'better' ear anymore.
8	I am hearing people so much better now. But I still don't like music on the radio.
9	It was a long month. I don't hear as well with this program.
10	I don't know what you did but after 3 years this is the first time I can hear words. It is so much better. It is a little sharp but there is no swoosh.
11	Before this map, after a long day I would have to take off my processor at night because I couldn't handle the overstimulation. Since I can read lips and use captioning, I was able to get by at nights. With this new map, I find myself wearing my processor from the time I wake up until I go to bed. Everything sounds so much smoother. It's not annoying. Listening in noise is effortless.
12	Sounds are clearer with deactivated electrodes but the longer I listened to the new program the quieter and higher pitch it got.
14	I DO NOT want that old program back. If I could have heard like this out of the gate, there's no telling how much better I would be hearing even today.
15	I hear sounds that I've never heard.
16	I knew as soon as I went home that I couldn't hear with this map. I couldn't understand on the phone. /S/ sounds funny and trails when at the end of a sentence. My own voice sounds very funny. But, music is much better. I prefer music with the research map.
19	It's better than my previous maps. it took me a week or two to get used to it. I am still missing out on some things but it is the first time I have really been able to hear speech.
20	I can hear so much better now. One day my battery died on my Left (good) ear and I didn't notice right away. I only noticed after talking to my daughter and thinking that my left ear was quieter so I tried to turn up the volume. I was hearing so well with just the Right implant and I could have never done that before.
21	Too overwhelming, miss clearvoice, can't tolerate in noise
22	I hear more, but I hear more high pitched sounds too. Sometimes it's too tinny.
23	Sometimes this map is more tolerable in noise but overall I think I hear better with my old map
25	Patient reported music is significantly better and feels as though she understands words better using this map
26	This is too high pitched. I can't tolerate it.
27	I am hearing better in noise. The improvement shown on your tests doesn't reflect the improvement that I am experiencing. Don't know why, but it doesn't. The improvement goes beyond the improvement in noise at 10db as I am hearing more distinctly in general. I don't know if you can enter this into your records of the study, but it is more reliable than the booth.
28	I know that I hear better with this map. It's been very obvious especially over the past month.
29	Definitely reduced clarity with this map. Had to ask for a lot of repetition from my wife.
33	I miss ClearVoice.' 'There are definitely things I hear better but in noise or while music is playing, I do better with ClearVoice.
34	Its much better.' 'I can hear some music.' 'I know I can hear more.
35	I can hear so much more than I could with the old program. It was an immediate change.
36	I don't like the sound of this map as much as my old map. But I do hear better in noise with this one.
37	I can hear better now but I still sound like I'm in a barrel.
38	Speech is a lot clearer though some of the higher frequency sounds are much more distinct and annoying. But I have gotten much more used to this over the past month.
39	There is less echo in my own voice with the experimental map.
40	The testing seemed to be easier this time.
41	The testing seemed easier this time.
42	Too high pitched. It gives me a headache. I can't understand my wife as well.
43	I like it. It is better. I definitely want to keep.' LE hearing fluctuates and is significantly worse than when tested for pre-op testing. This affected his bimodal scores. Patient to call and come in for repeat post testing when left ear hearing is better.
44	I like it. I want to keep it.
45	I can hear better in noise and with the TV. Music is much better with this program. It is hard to hear my wife but everything else is better with this program. I want to keep it.
46	There is no discernable difference in the sound quality or my performance.
47	She reported that she noticed no difference in noise and had a harder time hearing her husband and other voices. She noted that she would like to keep the experimental map as an option, but heard better with her previous map.
49	I love it. I can wear it while I sing and play music. Music is not as distorted. Much better quality than before.
50	I intend to use my experimental map as my primary map
51	It's a noisier program. I can't really tell if I do better or worse. I miss Clearvoice.
52	I'm less tired at the end of the day listening with this program. It's a quieter world. It's not a huge difference-maybe 10
53	I don't like it. Everyday listening is okay but I can't hear on the phone.
54	My test scores don't reflect the significant improvement I have had. This makes all the difference in the world. I no longer want to rip it off my head. Voices sound normal. I can wear it all day. I use to regret having the implant but now I don't.
55	I feel like I am hearing better. Other people have noticed the improvement, too, and have commented on it.
57	I think it's worse for understanding. It more balanced (volume) between ears.
58	There is more clarity with this program. There are no 'holes' in the frequency response. I can now use the phone on this ear. It needs more volume in the low frequencies, but I can understand more.
59	I can't hear on the phone.'; 'I can't monitor my own voice.'; 'I sound like I am in a barrel.
60	Sound quality is better and music is much better. I feel like I am hearing worse in noise, though. I'd really like to work with this map because I'd hate to go back to my old map. After listening to his old map: 'The new map is so much better. Wow! I had forgotten how bad it was before. The new map is so much clearer.
61	I hear worse with this program.
62	It's different but I got used to it. Music is harder to understand. It's less noisy, less background noise.
63	I think I hear more but understand less. I have a harder time in background noise than I did before.
65	It seems better is some situations, but I still have difficulty listening with both ears in other situations.
66	Communication is easier. I am hearing sounds that I haven't heard in awhile. Overall the change has been very positive.
67	Not better than previous map. Music is slightly better, but not enough improvement that I would listen to music (goes from 1 to 2 on a scale of 1 to 10).
68	Can't hear as well & can't understand
70	I like this better. I can hear better on that side now (I don't have to look at my instructor as much when she is on that side). I do have a little intermittent tinnitus that comes and goes with the new program. It was there when I first got my implant, but it's back. It doesn't really bother me and I usually ignore it when it's there.
71	There is less buzzing with the new map, but I don't know if I hear any better with it. I would like to keep both maps and keep trying them both.
72	It sounds like a munchkin and I do not understand well with it.