Published in final edited form as:

Ear Hear. 2011 ; 32(3): 323–330. doi:10.1097/AUD.0b013e3182008c56.

Electrically evoked compound action potential measures for virtual channels versus physical electrodes

Michelle L. Hughes, Ph.D.¹ and Adam M. Goulson, Au.D.²

¹Boys Town National Research Hospital, Omaha, Nebraska

²University of Nebraska-Lincoln, Lincoln, Nebraska

Abstract

Objectives—The number of distinct pitch percepts for cochlear implant (CI) listeners is somewhat limited by the number of physical electrodes in the array. Newer-generation CIs have the capability to potentially increase this number by stimulating areas of the cochlea between the physical electrodes. Currently this is achieved by electrically coupling adjacent electrodes or by simultaneously activating two electrodes with independent current sources (i.e., “current steering”). Presumably, either type of dual-electrode stimulation will generate neural excitation patterns that are intermediate to those generated by either physical electrode alone (henceforth termed virtual channel). However, it is not clear whether virtual-channel stimulation yields neural recruitment patterns with similar shapes and rates of growth as compared with each physical electrode alone. The purpose of this study was to compare basic electrically evoked compound action potential (ECAP) measures for physical electrodes and virtual channels to determine whether properties of the respective excitation patterns were similar.

Design—Data were collected for 12 adult CI recipients (N = 6 Nucleus Freedom CI24RE, N = 2 Advanced Bionics HiResolution 90K, N = 4 Advanced Bionics CII). ECAP responses were measured for a set of three adjacent physical electrodes and two corresponding intermediate virtual channels (e.g., physical electrodes 4, 5, and 6 and virtual channels 4+5 and 5+6) at three positions along the electrode array (basal, middle, and apical). Virtual channels for Nucleus subjects were produced via electrical coupling of adjacent electrode pairs (“dual-electrode mode”). For AB subjects, virtual channels were produced via simultaneous, in-phase stimulation of adjacent electrode pairs with 50 % of the total current delivered to each electrode in the pair. Specific ECAP measures were: (1) threshold and slope of the input/output (I/O) functions, (2) amplitude for a masker-probe interval of 1500 μ sec (measure of refractory recovery), and (3) the relative location of spread-of-excitation (SOE) functions among virtual channels and adjacent physical electrodes. Measures for virtual channels were compared with those for the flanking physical electrodes using a multivariate analysis of variance.

Results—There were no statistically significant differences between physical electrodes and virtual channels for ECAP thresholds, slope of the I/O function, or refractory recovery. On average, SOE functions for the virtual channels were spatially located approximately halfway between SOE functions for the adjacent physical electrodes.

Conclusions—Results from this study suggest that virtual channels produce neural recruitment patterns with properties similar to those elicited by the adjacent physical electrodes.

Keywords

cochlear implant; electrically evoked compound action potential; virtual channels; dual electrode; current steering; input/output; refractory recovery; spread of excitation

Introduction

Multi-channel cochlear implants (CIs) can provide significant speech perception benefits for individuals with severe to profound hearing loss. However, one factor that may limit recipients' performance is reduced spectral resolution caused by the limited number of physical electrodes in the array (e.g., Donaldson et al., 2005; Firszt et al., 2007; Koch et al., 2007; Bonham & Litvak, 2008; Frijns et al., 2009; Landsberger & Srinivasan, 2009). Current CI devices have 12 to 22 electrodes in the intracochlear array, depending on the manufacturer. One way to achieve additional pitch percepts with a limited number of electrodes is to stimulate the neural populations between adjacent electrodes using either sequential or simultaneous stimulation. The resulting percept is typically termed a “virtual channel.”¹

Sequential stimulation involves the consecutive (or interleaved, in the case of pulse trains) presentation of stimuli on two adjacent or non-adjacent electrodes within a very short period of time (e.g., less than 500 μ sec; McDermott & McKay, 1994; McKay et al., 1996; Kwon & van den Honert, 2006; Frijns et al., 2009; Saoji et al., 2009). With this method, the first pulse should recruit a population of neurons in the vicinity of the stimulated electrode. The second pulse should ideally recruit an additional population of neurons that are either physically separate from the first population or partially polarized by the first pulse. With a short inter-stimulus interval, the two electrodes can effectively elicit one combined neural response pattern that is perceived as a single pitch (Tong and Clark, 1986; McDermott and McKay, 1994; McKay et al., 1996; Frijns et al., 2009). The pitch percept can change as a function of the relative amplitude of the current pulses delivered to consecutive electrodes. Kwon and van den Honert (2006) reported an average of 6.3, 3.4, and 2.6 discriminable steps between adjacent physical electrodes for apical, middle, and basal electrode pairs, respectively. Saoji & Litvak (Reference Note 1) showed that the pitch percept for stimuli applied sequentially to non-adjacent electrodes (e.g. E6 and E8) can be matched to that for the intermediate physical electrode (e.g., E7). However, electrophysiological studies indicate that different spatial excitation patterns are produced with sequential stimulation of non-adjacent electrodes (e.g. E6 and E8) compared to that produced by the intermediate physical electrode (e.g., E7; Saoji et al., 2009).

With simultaneous stimulation, two slightly different methods can be used. With the first method, the amount of current normally delivered to a single electrode is split between two simultaneously stimulated electrodes. The ratio of current can then be varied between the two electrodes to produce systematic changes in the shape of the electric field and subsequent neural recruitment patterns. This method has been referred to as “current steering” (Firszt et al., 2007; Koch et al., 2007; Bonham & Litvak, 2008; Choi & Hsu, 2009; Frijns et al., 2009; Saoji et al., 2009) and can be used with devices having individual current

¹Typically, the term “channel” is used in regard to perceptual measures, whereas “electrode” refers to the physical point of current injection. For example, a 12-electrode array may only produce 8 distinct pitch percepts, or channels. Although the present study describes results for physiological measures and not perceptual measures, the term “virtual channel” is used here in a more generic sense to describe an electrode configuration that is used to recruit neural populations between two adjacent physical electrodes. Reference Note Saoji, A. A. & Litvak, L. M. (2007). Equal loudness contours and discrete pitch percepts for simultaneous and sequential dual electrode stimulation as a function of spatial electrode separation. Conference on Implantable Auditory Prostheses, Lake Tahoe, CA.

sources for each electrode, such as the Advanced Bionics (AB) and Med-El I¹⁰⁰ devices. With current steering, Firszt et al. (2007) reported an average of 5.3, 6.0, and 3.8 discriminable steps between adjacent physical electrodes for apical, middle, and basal electrode pairs, respectively. These results were similar to those reported by Kwon and van den Honert (2006) for sequential stimulation. With the second method, two adjacent electrodes are shorted or electrically coupled together to create a single “electrode” that ideally recruits a slightly different pattern or population of neurons than either of the single electrodes alone. This method has been referred to as “dual-electrode” stimulation, which can be used with Cochlear Corporation's Nucleus 24RE Freedom (Busby & Plant, 2005; Busby et al., 2008) or N5 devices. In most cases, an intermediate pitch is distinguishable, and it is generally tonotopically ordered relative to the two single contributing electrodes (Busby & Plant, 2005; Busby et al., 2008). One limitation with dual-electrode stimulation is the inability to manipulate the precise location of the electrical field due to the lack of independent current sources for each electrode. Further, the shape and location of the resultant electric field will depend on the relative impedance of each coupled electrode.

Few studies have compared electrophysiological measures for physical electrodes versus virtual channels. Busby et al. (2008) examined electrically evoked compound action potential (ECAP) spread-of-excitation (SOE) functions for single versus dual electrodes in nine adults with the Nucleus 24RE Freedom device. In general, they found no significant difference in the widths of the SOE functions for single versus dual electrodes. In most cases, the peaks and edges of the SOE functions for dual electrodes were spatially located between those for the flanking single electrodes. In some instances, however, the peak amplitude of the SOE function for the dual electrode was larger than for the flanking single electrodes. In a study using current steering in AB devices, Saoji et al. (2009) compared ECAP SOE functions for physical electrodes and virtual channels. For the virtual channels, the total amount of current was equally split between a pair of non-adjacent physical electrodes (i.e., E6 and E8). The resulting SOE function was compared with that for the intermediate physical electrode (i.e., E7). There was no significant difference in either the area under the curve or center of gravity for physical electrodes versus virtual channels. In general, results from these two studies suggest that virtual channels (created with either electrical coupling or current steering) yield patterns of excitation that are similar to those generated by physical electrodes.

There do not appear to be any published reports comparing other basic ECAP measures such as threshold, slope of the input/output (I/O) function, or refractory recovery for physical electrodes versus virtual channels. Based on the existing results for SOE functions (Busby et al., 2008; Saoji et al., 2009), we might expect other basic ECAP measures to be similar for physical electrodes and virtual channels. This hypothesis is based on the assumption that virtual channels will produce neural recruitment patterns that behave in the same way as those produced by physical electrodes. However, even though SOE functions are similar for both physical electrodes and virtual channels at relatively high levels, there remains the possibility that the neural recruitment patterns exhibit different properties as a function of stimulus level or timing. Therefore, an alternative hypothesis is that virtual channels may yield significant differences in level-dependent (e.g., threshold or slope of the I/O function) or temporal-dependent (e.g., refractory recovery) measures, and these differences may depend on the specific method used to create the virtual channel (i.e., electrical coupling or current steering).

For current steering, Frijns et al. (2009) suggested that electrodes with greater spatial selectivity (particularly where the electrode-nerve distance is small) need higher stimulus levels for the proper fusion of current fields from two simultaneously stimulated electrodes. For an I/O function, each contributing current field would overlap to a lesser extent as the

probe level is reduced, eventually resulting in two separate excitation areas for each electrode at the lowest levels. With the standard forward-masking subtraction method (e.g., Abbas et al., 1999, 2004) in which the masker is fixed at a high level, this would result in a higher ECAP threshold and steeper I/O slope for the virtual channel. Smaller amplitudes might be expected in the recovery function as well; although this would depend on the overall stimulus level used. We would not expect higher thresholds and steeper slopes in cases where virtual channels are achieved via electrical coupling of adjacent electrodes because that method does not rely on the summation of two separate electrical fields, and is therefore not subject to the same level-dependent limitations described by Frijns et al. (2009).

For electrical coupling, Busby and Plant (2005) showed significantly lower impedances for dual electrodes than for each contributing physical electrode alone. This is because electrical coupling essentially results in a larger overall surface area for the combined dual electrode. As stated above, Busby et al. (2008) also found larger peak amplitudes of some SOE functions for the dual electrode as compared with the flanking single electrodes. It is possible that electrical coupling might yield a slightly larger number of recruited neurons than either adjacent physical electrode alone. This might result in lower ECAP thresholds, steeper I/O slopes, and larger amplitudes in the refractory recovery function.

The objective of the present study was to determine whether ECAP threshold, I/O slope, refractory recovery, and the relative location of SOE functions for virtual channels are consistent with those measures for the corresponding adjacent physical electrodes. Measures made with both current steering (AB devices) and electrical coupling (Nucleus devices) were examined.

Materials and Methods

Subjects

Data were collected for 12 adult CI recipients [N = 6 Nucleus Freedom CI24RE (Cochlear Ltd., Lane Cove, Australia), N = 2 HiResolution 90K (AB, Sylmar, CA), N = 4 CII (AB, Sylmar, CA)]. Demographic information for participants is listed in Table 1. The Nucleus devices had the perimodiolar Contour electrode array, which consists of 22 electrodes spaced 0.4 mm (apical end) to 0.81 mm (basal end) apart (center to center) over a length of 15 mm. The AB devices had the straight HiFocus array, which consists of 16 electrodes spaced 1.1 mm apart (center to center) over 19 mm. Only subject C1 had an electrode positioner. Electrode impedance measures indicated short circuits for C19 (electrodes 9 and 14), and open circuits for F2 (electrode 8) and C15 (electrode 16); these electrodes were not included in the study. This study was approved by the Boys Town National Research Hospital (protocol 03-07-XP) Institutional Review Board.

ECAP Stimuli and Procedure

All ECAP measures were made using the standard forward-masking subtraction technique described previously (e.g., Abbas et al., 1999, 2004; Brown et al., 1998; Hughes and Abbas, 2006a & b). For Nucleus recipients, Custom Sound EP v.2.0 experimental research software (Cochlear Ltd., Lane Cove, Australia) was used to control stimuli and record ECAPs. Stimuli were delivered through a laboratory Freedom speech processor interfaced with a programming pod. Virtual channels were produced via electrical coupling of adjacent electrode pairs (e.g., coupling physical electrodes 4 and 5 produced virtual channel 4+5). The following default parameters were used: 25 μ s/phase pulse width; 7 μ sec interphase gap; 80 Hz probe rate; 50 dB gain; 400 μ sec masker-probe interval (MPI); stimulating reference electrode MP1 (extracochlear monopolar ball electrode); and recording reference electrode

MP2 (extracochlear monopolar case electrode). The recording location for each physical electrode was two electrode positions apical to the probe. For virtual channels, the recording location was two electrode positions apical to the more basal electrode in the physical pair (software default). Responses were typically an average of 50 sweeps with a 122- μ sec recording delay.

For AB recipients, the Bionic Ear Data Collection System (BEDCS, v1.18.295) experimental research software (Advanced Bionics, Sylmar, CA) was used. Stimuli were delivered through a laboratory Platinum Series speech processor interfaced with a Clinical Programming Interface (CPI-II). Virtual channels were produced via simultaneous stimulation of adjacent electrode pairs with 50 % of the total current delivered to each electrode in the pair (50-50 split). The following parameters were used: 32 μ s/phase pulse width; 20 Hz probe rate; 1000 gain (linear multiplier); 500 μ sec MPI; monopolar stimulating reference electrode (implant case); and monopolar recording reference electrode (implant case electrode for CII subjects; extracochlear ring electrode for 90K subjects). The relative recording electrode positions were the same as for Nucleus subjects. Responses were recorded using 80 averages.

ECAP data were measured for a set of five electrodes/channels at three positions along the array (basal, middle, and apical). For each cochlear position, the electrode set consisted of three adjacent physical electrodes and two corresponding intermediate virtual channels. The rightmost column of Table 1 lists the electrode sets used for each participant. For example, electrode set 4-6 included physical electrodes 4, 5, and 6, and virtual channels created through dual stimulation of electrodes 4+5 and 5+6. For Cochlear and AB devices, electrodes are labeled 1-22 and 16-1 in a basal to apical direction, respectively.

For each subject and electrode, behavioral loudness estimates for Cochlear devices were measured using log-based² 5-current-level (CL) ascending increments. For AB devices, behavioral loudness estimates were measured using a similar log-based ascending approach with a step size of approximately 0.44³. Subjects used a visual rating scale (from Advanced Bionics) that ranged from 0 (no sound) to 10 (too loud) to indicate when the sound was first heard (1), first gauged as loud but tolerable (8), and upper loudness comfort limit (9). A single ascending run was used to obtain loudness estimates on each electrode. Because most of the measures in the present study are highly dependent on stimulus level, current levels for a loudness rating of “8” were compared across electrodes to determine whether intermediate electrodes required more current for equal loudness compared with physical electrodes. A one-way repeated-measures analysis of variance (ANOVA; SigmaStat 3.0, SPSS Inc., Chicago, IL), separated by device type, revealed no significant differences in current levels across electrodes for either Nucleus (1-way ANOVA on ranks due to non-normal distribution of data; $X^2 = 14.94$, $p = 0.38$, $df = 14$) or AB devices ($F = 1.15$, $p = 0.34$, $df = 14$).

I/O functions were measured by fixing the masker at a current level rated “8”⁴. For Cochlear devices, the probe was decreased in steps of 5 CL until ECAP responses were no longer visually identifiable. The stimulus was then raised by 2-3 CL to obtain a more precise ECAP threshold. For AB devices, the probe was decreased using a step size of 20-25 μ A, until the ECAP was no longer visually identifiable. Threshold was defined as the lowest probe level

²The formula to convert to current (I) for the Nucleus CI24RE device is $I = 17.5 \cdot 100^{\wedge}(\text{CL}/255)$.

³The formula is $b = a10^{\wedge}(\text{step}/20)$ where a was the starting current level, b was the next computed current level, and $step$ was the step size of approximately 0.44 (from Hughes & Stille, 2009).

⁴The exception was subject F4, who required lower current levels due to voltage compliance limits.

where a response was visually identifiable. The slopes of the I/O functions for each subject were calculated by applying a simple linear regression to each function.

Refractory recovery functions were obtained with the masker and probe at equal levels on the same electrode, using the same current levels as for the fixed masker in the I/O function. The MPIs were systematically varied from 100-10,000 μsec . With the standard forward-masking technique, the maximum amplitude of recovery functions generally occurs at short MPIs (i.e. 300-500 μsec) with decreasing amplitudes for longer MPIs. Responses are typically no longer measurable at MPIs in the range of 3000-6000 μsec (Abbas et al., 1999; Dillier et al., 2002; Morsnowski et al., 2006; Shpak et al., 2004). Thus, refractory recovery functions were quantified as the ECAP amplitude at an MPI of 1500 μsec , which allowed for differences across subjects to be best examined.

SOE functions were measured using methods described previously (Abbas, et al., 2004; Cohen, et al., 2003; Hughes & Abbas, 2006 a, b). Briefly, the probe electrode and level (current level corresponding to a loudness rating of “8”) were fixed while the masker electrode and level were systematically changed along the electrode array. Current levels for each masker electrode corresponded to a loudness rating of “8”. The masker was presented to all physical electrodes (except the recording electrode)⁵. For virtual-channel SOE functions, the masker was also applied to the same virtual channel as the probe. Raw (not normalized) ECAP amplitudes were used, and the same masker/probe level was used for the virtual channel and adjacent physical electrodes to avoid confounding level effects (described further in the Results). Within each cochlear position, the lowest current level that yielded a loudness rating of “8” across the set of five probe electrodes (three physical and two virtual) was used.

To quantify the location of the SOE function for the virtual channel relative to the functions for the adjacent physical electrodes, the following equation was applied for each masker electrode in the function:

$$(A_{VC} - A_{APE}) / (A_{BPE} - A_{APE}),$$

where A_{VC} is the amplitude for the virtual channel, A_{APE} is the amplitude for the apical flanking physical electrode, and A_{BPE} is the amplitude for the basal flanking physical electrode. The resulting value is a proportion of the amplitude difference between the two flanking physical electrodes. Ideally, amplitudes for the virtual-channel SOE function should be approximately 0.5, or 50% of the difference between amplitudes for the adjacent physical electrodes. Values of 1 or 0 would represent exact overlap with the functions for the basal or apical flanking physical electrodes, respectively.

Data Analysis

All ECAP amplitudes were measured as the difference in voltage between the N1 and P2 peaks. For Cochlear subjects, the peaks were initially selected by the automatic algorithm in the Custom Sound EP software. Peak markers were manually adjusted as necessary, based on visual inspection. For AB subjects, ECAP data were read into a custom Matlab (Mathworks Inc., Natick, MA) program, and peaks were marked manually. A mixed-model MANOVA (multivariate ANOVA; SPSS v. 17.0, SPSS Inc., Chicago, IL) with the factors of electrode (repeated measure) and device (fixed factor) were used to test for differences

⁵For a subset of functions, maskers were also applied to all virtual channels to demonstrate the effects of stimulus level and normalization (discussed further in Fig. 5).

across threshold, slope, and recovery. Post-hoc analyses were completed using Tukey's Honestly Significant Difference (HSD) test to determine whether significant differences existed between measures for virtual channels and the adjacent physical electrodes.

Results

Input/Output Functions

Figure 1 shows I/O functions for one virtual channel (open circles) and two adjacent physical electrodes (filled symbols) from subject F5. The shape of the I/O functions for the virtual channel and physical electrodes were similar. In this example, amplitudes for the basal physical electrode (E19) were lower than for the apical physical electrode (E20). As expected, the amplitudes for the intermediate virtual channel (E19 +20) were generally between the amplitudes for the adjacent physical electrodes.

Figure 2 shows the mean normalized ECAP thresholds (+1 standard deviation, SD) for the basal, middle, and apical electrode sets across all subjects. First, each subject's ECAP thresholds were normalized to the threshold for the most basal electrode within each set. This was done so that the relative amount of threshold shift across the five electrodes/channels in a set could be compared across subjects. Normalized thresholds were then averaged across subjects for each electrode. In Fig. 2, the physical electrodes are labeled in a basal-to-apical direction as B1-B3 (basal), M1-M3 (middle), and A1-A3 (apical). As an example, B1-B3 corresponds with electrodes 4-6 for Nucleus and 13-11 for AB (see rightmost column in Table 1). Black and gray bars represent the normalized ECAP thresholds for the physical electrodes and virtual channels, respectively. The number of subjects with measurable thresholds for each electrode set is listed above each grouping.

A MANOVA (Table 2) revealed a significant main effect of electrode ($F = 2.144$, $p = 0.014$, $df = 14$) and a significant interaction between device and electrode ($F = 2.071$, $p = 0.018$, $df = 14$). There were no significant effects of device on threshold ($F = 2.679$, $p = 0.104$, $df = 1$). The minimum mean difference in threshold for pair-wise comparisons of electrode using Tukey's HSD was 0.132. (This means that the difference in mean threshold between electrode pairs had to be 0.132 or greater for the comparison to be significant at an alpha of 0.05.) Post-hoc analyses on threshold data for virtual channels versus the adjacent physical electrodes revealed no significant differences for any of the electrode pairs (mean differences between adjacent pairs ranged from 0.004 to 0.081). Regarding the significant interaction between electrode and device, the minimum mean difference in threshold for pair-wise comparisons of electrode within device was 0.354 (Tukey's HSD). Post-hoc analyses on threshold (each device separately) for virtual channels versus the adjacent physical electrodes revealed no significant differences for any of the electrode pairs (mean differences between adjacent pairs ranged from 0 to 0.146).

Figure 3 shows the mean normalized I/O slopes (+1 SD) for the basal, middle, and apical electrode sets across all subjects. Data are plotted and were normalized using the same procedures as for Fig. 2. A MANOVA (Table 2) showed no significant main effect of electrode ($F = 0.461$, $p = 0.949$, $df = 14$) or device ($F = 2.971$, $p = 0.087$, $df = 1$) for all three cochlear regions. There was no significant interaction between electrode and device ($F = 1.528$, $p = 0.111$, $df = 14$). As with threshold, there was no significant difference in the slope of the I/O function between any of the virtual channels and the adjacent physical electrodes.

Refractory Recovery

Figure 4 shows the mean normalized ECAP amplitudes for an MPI of 1500 μsec (+1 SD) for the basal, middle, and apical electrode sets across all subjects. Data were normalized as for Figs. 2 and 3. A MANOVA (Table 2) revealed no significant main effect of electrode ($F =$

0.513, $p = 0.922$, $df = 14$), or device ($F = 1.637$, $p = 0.203$, $df = 1$) for all three cochlear regions. There was no significant interaction between electrode and device ($F = 0.728$, $p = 0.743$, $df = 14$). As with threshold and slope, there was no significant difference in refractory recovery between any of the virtual channels and the adjacent physical electrodes.

Spread of Excitation

For much of the ECAP SOE data in the literature, normalization was used to control for amplitude differences that typically result from differences in stimulus level. Figure 5 illustrates the effects of probe level and amplitude normalization on SOE functions. Data are for one virtual (open circles) and two adjacent physical (filled symbols) probe electrodes for two subjects (F4, left column; F1, right column). To better illustrate this issue, complete SOE functions in which maskers were applied to all physical electrodes and virtual channels are shown. Figures 5A and 5B show raw ECAP amplitudes as a function of masker electrode for subjects F4 and F1, respectively. For F4 (Fig. 5A), probe levels were the same for all three functions (210 CL), resulting in approximately equal height, width, and shape of the functions. For F1 (Fig. 5B), probe levels were 200 CL for P14 and 205 CL for P14+15 and P15 (loudness rating of “8”), yielding an overall lower-amplitude function for P14.

Figures 5C and 5D show the raw ECAP amplitudes from Figs. 5A and 5B, respectively, normalized to the amplitude obtained with the masker and probe on the same electrode. In this example, the normalization serves to neutralize differences in the height of functions due to level for F1; however, it also changes the relative location of the edges of the SOE patterns. This latter issue is also illustrated with the data for F4, for whom there were no stimulus level differences (thus normalization should have had a minimal effect). In Fig. 5A, the basal-side edge of the function (i.e., masker E1-10) for P11+12 (open circles) is located near the function for the basal physical electrode (P11, filled circles). On the apical side, the edge of the function for P11+12 is located closer to the function for the apical physical electrode (P12, filled triangles). These relative relationships change for the normalized data in Fig. 5C, where the edge of the function for P11+12 is located closer to the function for the apical physical electrode (P12, filled triangles) on the basal side of the function (i.e., masker E1-10). On the apical side, the edge of the function for the intermediate electrode is located closer to the function for the basal physical electrode (P11, filled circles).

Based on the results from Busby et al. (2008) for normalized data and Saoji et al. (2009) for non-normalized data, it was expected that the peaks and edges of the SOE functions for virtual channels would be spatially located between those for the flanking physical electrodes. Theoretically, the SOE functions for the virtual channels should fall roughly half-way between the SOE functions for the adjacent physical electrodes. Figure 6 shows the location of virtual-channel SOE functions relative to the flanking physical electrodes (top: Nucleus, bottom: AB). Symbols represent group mean data (± 1 SD) for basal (filled circles), middle (open circles), and apical (gray triangles) electrode sets. Horizontal solid lines indicate the expected result of 0.5. The number of virtual-channel probe electrodes included in each mean is indicated in the figure legends. Averages across all three electrode regions within each device type were 0.4 (SD = 0.81) for AB and 0.4 (SD = 0.79) for Nucleus. Given the large standard deviations, these means were not significantly different from the expected value of 0.5 ($p > 0.5$, Mann-Whitney Rank Sum test). The results indicate that, on average, virtual-channel SOE functions were located approximately halfway between those for the two flanking physical electrodes.

Discussion

Input/Output Functions and Refractory Recovery

This study augments earlier reports describing the size and relative location of SOE functions obtained with virtual channels (Busby et al. 2008; Saoji et al., 2009), and represents a first report of ECAP threshold, slope, and recovery measures made with virtual-channel stimulation for both current steering (AB devices) and electrical coupling (Nucleus devices). In general, there were no statistically significant differences between virtual channels and the adjacent physical electrodes for ECAP thresholds, slopes of the I/O functions, or refractory recovery. These results suggest that stimulation patterns for virtual channels are generally similar in size compared with adjacent physical electrodes; they exhibit similar growth properties, and are generally spatially located between those for the adjacent physical electrodes. However, limited conclusions can be drawn from the present data set due to the small sample size split between two device types. It would therefore be of interest to validate the present findings with a larger group of subjects with both devices.

As discussed in the Introduction, better spatial selectivity may be disadvantageous for current steering, particularly at low stimulus levels (Frijns et al., 2009). In the present study, all AB subjects had the HiFocus array, which is a straight electrode array that typically occupies a mid-scalar or lateral-wall position (i.e., Skinner et al., 2007). A non-perimodiolar position is associated with reduced spatial selectivity (Hughes & Abbas, 2006b), which would be more desirable for current steering when the goal is proper fusion of electric fields from two electrodes (Frijns et al., 2009). Thus, the lack of perimodiolar electrode positioning in the AB subjects may account for virtual channels and physical electrodes yielding similar results. Only one subject (C1) in the present study had an electrode positioner, which is designed to place the electrode array in close proximity to the modiolus. Data from that subject showed no notable differences in threshold, slope, or recovery between virtual channels and physical electrodes. However, the subject was congenitally deafened and may have had a limited neural population that precluded good spatial selectivity, even with a positioner.

Spread of Excitation

The forward-masking subtraction paradigm (Abbas et al., 2004) was used to determine whether the SOE functions for virtual channels were spatially located between the SOE functions for adjacent physical electrodes. Theoretically, the SOE functions for the virtual channels should fall roughly half-way between the SOE functions for the adjacent physical electrodes. With current steering, half of the total current is delivered by each electrode. Assuming complete summation of the contributing fields, the center of gravity of the fused electric field should be located between the two contributing physical electrodes (Frijns et al., 2009; Saoji et al., 2009). With electrical coupling (as in Cochlear's dual-electrode mode), the current path for the virtual channel is determined by the relative impedance of the two contributing electrodes. If one electrode in the pair has significantly lower impedance than the other, the center of gravity of the resulting field for the dual electrode may possibly be weighted toward the lower-impedance single electrode. In the Contour array, the electrode surface area decreases at electrode 11 and again at electrode 17 (C. van den Honert, personal communication). Larger surface area is associated with lower impedance, so impedance differences may be more of an issue for dual electrodes 10+11 and 16+17. In the present subject group, the difference in impedance between adjacent pairs of test electrodes was no more than 2 kOhms (re: MP1). It is unclear how much of a difference in impedance would yield a significant shift in the pattern; this is a topic worthy of further detailed investigation. For dual-electrode stimulation, Busby et al. (2008) showed, in most cases, the SOE functions for the virtual channel were tonotopically ordered with SOE functions for adjacent

electrodes. Results from the present study showed that, on average, SOE functions for virtual channels were located approximately halfway between those for the two flanking physical electrodes. There was, however, large variability across the measures, which was likely the result of averaging narrow SOE functions for some subjects with broader SOE functions for other subjects.

Normalization of ECAP SOE functions has been used routinely to control for differences in stimulus level (or differences in amplitude independent of stimulus level), particularly when used in comparison with psychophysical data (e.g., Hughes & Abbas, 2006a, 2006b; Hughes & Stille, 2008; Busby et al., 2008). Results from Fig. 5 in the present study illustrate how normalization of SOE functions changes the relative relation between the edges and/or peaks of the functions. This is important to consider if the relative locations of the edges of the functions are to be compared with other measures, as in Busby et al. (2008). In that study, the relative ordering of the peaks and edges of the SOE functions were not significantly correlated with psychophysical pitch ranking results. Two issues regarding SOE normalization may have affected that outcome. First, SOE functions were normalized to the highest ECAP amplitude within each function. Generally, the peak of the function occurs with the masker and probe delivered to the same electrode. However, the masker was only delivered to physical electrodes, so the normalization point for the virtual-channel probe function was likely not ideal. Second, as shown in Fig. 5, normalization of SOE functions changes the relative location of the edges of the SOE functions. In the present study, SOE functions were obtained using the same current levels for the physical and virtual probe electrodes so as to avoid stimulus level effects and the need for normalization. This might be the preferred approach for studies in which the relative location of pattern peaks or edges is a measure of interest.

In conclusion, this study demonstrated similar ECAP thresholds, I/O slopes, and refractory recovery measures for virtual channels versus flanking physical electrodes. On average, SOE functions for the virtual channels were spatially located approximately halfway between SOE functions for the adjacent physical electrodes. Results from this study suggest that virtual-channel stimulation with either current steering or electrical coupling typically produces neural recruitment patterns with properties similar to those elicited by physical electrodes.

Acknowledgments

This research was supported by the NIH/NIDCD, R01 DC009595 and P30 DC04662. The content of this project is solely the responsibility of the authors and does not necessarily represent the official views of the NIH or the NIDCD. The authors thank Lisa Stille and Katelyn Rosemond for assistance with data collection and analysis, Tom Creutz for programming support, Leo Litvak for the BEDCS data collection program, Ryan McCreery for statistical assistance, and three anonymous reviewers for helpful comments on an earlier version of this manuscript.

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Fig. 1. Individual example of input-output (I/O) functions for two physical electrodes (filled symbols) and one virtual channel (open circles) for subject F5. Probe electrode numbers are indicated in the legend.

Fig. 2.

Mean normalized ECAP thresholds (+1 SD) for the basal, middle, and apical electrode sets. Physical electrodes and virtual channels are indicated by black and gray bars, respectively. The physical electrodes are labeled as B1-3, M1-3, and A1-3. For Nucleus and Advanced Bionics subjects B1-3 corresponds with electrodes 4-6 and 13-11, respectively. The same numbering system is used for the middle and apical physical electrodes (see Table 1). The number of subjects included in each electrode set is indicated by the (N) above each electrode set.

Fig. 3. Mean normalized input-output (I/O) function slopes (+1 SD) for the basal, middle, and apical electrode sets. Physical electrodes and virtual channels are indicated by black and gray bars, respectively. The number of subjects included in each electrode set is indicated by the (N) above each electrode set.

Fig. 4. Mean normalized ECAP amplitudes (+1 SD) for the basal, middle, and apical electrode sets. Physical electrodes and virtual channels are indicated by black and gray bars, respectively. The number of subjects included in each electrode set is indicated by the (N) above each electrode set.

Fig. 5. Spread-of-excitation functions for the physical probe electrodes (filled symbols) and virtual probe (open circles) for subjects F4 and F1. Panels A and B represent raw ECAP amplitudes. Panels C and D represent the ECAP amplitudes normalized to the amplitude obtained with the masker and probe on the same electrode. Probe electrode numbers are indicated in the legend.

Fig. 6.

Group data showing the average proportion at which the spread-of-excitation functions for the virtual channel fell between the amplitudes for the physical electrodes. Top and bottom panels represent data for Nucleus and Advanced Bionics subjects, respectively. Horizontal solid lines ($y = 0.5$) represent the expected value. The number of virtual-channel probes (N) is indicated in each panel.

Demographic information for study participants, including the electrode sets tested for each subject (rightmost column). All electrode numbers are listed basal to apical and are inclusive of one intermediate virtual channel between each pair of adjacent physical electrodes (example: 4-6 includes physical electrodes 4, 5, and 6 and virtual channels 4+5 and 5+6). Electrode sets with no measurable ECAP response are indicated by NR (no response). Electrodes 9 and 9+8 were not included for C19 due to short circuit. SNHL = sensorineural hearing loss.

Table 1

Subject	Device	Age at CI (yrs, mos)	Duration CI use (yrs, mos)	Etiology	Electrode Set (Basal, Middle, Apical)
F1	Nucleus 24RE Freedom	60, 7	3, 0	Unknown	4-6, 9-11, 19-21
F2	Nucleus 24RE Freedom	60, 2	1, 10	Unknown	4-6, 10-12, 18-20
F4	Nucleus 24RE Freedom	17, 6	1, 11	Ototoxicity	4-6, 10-12, 18-20
F5	Nucleus 24RE Freedom	48, 3	0, 9	Unknown	4-6, 10-12, 18-20
F6	Nucleus 24RE Freedom	31, 0	3, 11	Unknown	4-6, 10-12, 18-20
F7	Nucleus 24RE Freedom	39, 1	2, 4	Unknown	4-6, 10-12, 18-20
C1	Advanced Bionics CII w/positioner	18, 4	6, 11	Unknown	13-11, 9-7, 5-3
C7	Advanced Bionics CII	57, 2	5, 2	Genetic-Unspecified	NR, 9-7, NR
C8	Advanced Bionics CII	55, 7	4, 10	Sudden	13-11, 9-7, 5-3
C13	Advanced Bionics CII	77, 1	6, 6	Genetic-Unspecified	13-11, 9-7, 5-3
C15	Advanced Bionics HiRes 90K	39, 3	4, 8	Unknown	NR, NR, 5-3
C19	Advanced Bionics HiRes 90K	17, 3	3, 6	Sudden from established SNHL	13-11, 8-7, 5-3

Multivariate analysis of variance for the effects of device (fixed factor) and electrode (repeated measure) on threshold, slope, and recovery.

Table 2

Tests of Between-Subjects Effects						
Source	Dependent Variable	Type III Sum of Squares	df	Mean Square	F	Partial Eta Squared
Device	Threshold	.022	1	.022	2.679	.104
	Slope	.193	1	.193	2.971	.087
	Recovery	.213	1	.213	1.637	.203
Electrode	Threshold	.243	14	.017	2.144	.014
	Slope	.419	14	.030	.461	.949
	Recovery	.933	14	.067	.513	.922
Device * Electrode	Threshold	.235	14	.017	2.071	.018
	Slope	1.387	14	.099	1.528	.111
	Recovery	1.323	14	.095	.728	.743
Error	Threshold	.997	123	.008		
	Slope	7.979	123	.065		
	Recovery	15.969	123	.130		

^aR Squared = .311 (Adjusted R Squared = .149)
^bR Squared = .201 (Adjusted R Squared = .013)
^cR Squared = .132 (Adjusted R Squared = -.073)