Consonant recognition as a function of the number of stimulation channels in the Hybrid short-electrode cochlear implant^{a)}

Lina A. J. Reiss,^{b)} Christopher W. Turner, Sue A. Karsten, Sheryl R. Erenberg,
and Jessica Taylor
Department of Communication Sciences and Disorders, University of Iowa, Iowa City, Iowa 52242

Bruce J. Gantz
Department of Otolaryngology–Head Neck Surgery, University of Iowa, Iowa City, Iowa 52242

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Consonant recognition was measured as a function of the number of stimulation channels for Hybrid short-electrode cochlear implant (CI) users, long-electrode CI users, and normal-hearing (NH) listeners in quiet and background noise. Short-electrode CI subjects were tested with 1–6 channels allocated to a frequency range of 1063–7938 Hz. Long-electrode CI subjects were tested with 1–6, 8, or 22 channels allocated to 188–7938 Hz, or 1–6 or 15 channels from the basal 15 electrodes allocated to 1063–7938 Hz. NH listeners were tested with simulations of each CI group/condition. Despite differences in intracochlear electrode spacing for equivalent channel conditions, all CI subject groups performed similarly at each channel condition and improved up to at least four channels in quiet and noise. All CI subject groups underperformed relative to NH subjects. These preliminary findings suggest that the limited channel benefit seen for CI users may not be due solely to increases in channel interactions as a function of electrode density. Other factors such as pre-operative patient history, location of stimulation in the base versus apex, or a limit on the number of electric channels that can be processed cognitively, may also interact with the effects of electrode contact spacing along the cochlea.

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I. INTRODUCTION

The Hybrid or “short-electrode” cochlear implant (CI) is a short, thin version of the traditional CI designed to be implanted atraumatically and preserve residual low-frequency hearing (Gantz and Turner, 2003, 2004). While the longer, traditional 24 mm electrode array is implanted at least one turn around the cochlea, the shorter, 10 mm short electrode CI is implanted into just the basal region of the cochlea. The short electrode CI has only 6 electrodes compared to 22 electrodes in the equivalent full-length or “long-electrode” CI array from Cochlear, and has the same inter-electrode spacing (0.75 mm between adjacent electrode contacts). The short-electrode CI is designed to provide high frequency information electrically together with low-frequency information from residual acoustic hearing, thus allowing combined “acoustic + electric” hearing in the same ear.

The short-electrode CI design is targeted toward a different patient population, those with a mild-to-moderate

hearing loss at low frequencies sloping to a severe-to-profound hearing loss at higher frequencies. Patients in this population have too much usable residual hearing to qualify for a full-length CI, but still struggle with speech recognition with hearing aids. Recent clinical trials with the short-electrode CI show that the combination of the electrical speech processing together with the residual hearing has enabled this group to improve their consonant-nucleus-consonant (CNC) word understanding from mean pre-operative scores of 35% with two hearing aids to post-operative scores of 74% with the CI and both hearing aids after at least one year of implant use (Reiss *et al.*, 2008).

The consonant recognition benefit obtained from the short-electrode CI is interesting in the context of previous studies in long-electrode CI users, which suggest that speech recognition benefit is limited to 7–8 stimulation channels. Stimulation channels refer to the number of frequency bands of information provided, and correspond to the “active” electrodes in CI electrode arrays (in cases when some electrodes are deactivated), or the noiseband channels in acoustic CI simulations presented to normal-hearing (NH) listeners. When the number of stimulation channels is reduced, with the stimulation channels set up to stimulate the same approximate cochlear distance and the analysis filters expanded accordingly to span the same frequency range as the full electrode array, as few as 3–4 stimulation channels are needed to understand a single talker in quiet (Shannon *et al.*, 1995; Fishman *et al.*, 1997). For more difficult speech with multiple talkers, and for speech in noise, more channels are needed; long-electrode CI users continue to improve their

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^{b)}Author to whom correspondence should be addressed. Electronic mail: reiss@ohsu.edu

speech recognition up to 7–8 channels (Friesen *et al.*, 2001). In contrast, NH subjects listening to a simulation of long-electrode processed speech improve their speech recognition up to 22 or more channels and so achieve higher speech recognition scores than CI subjects (Friesen *et al.*, 2001; Xu and Pfingst, 2008).

One proposed explanation for the limited ability of CI users to improve speech recognition with more than 7–8 stimulation channels out of up to 22 available intracochlear electrodes is that current spread and the resulting channel interactions limit the speech recognition benefit to every third electrode. If this is the case, then short-electrode CI subjects may similarly only be able to use every third electrode (of 6 total); thus 2–3 stimulation channels out of 6 intracochlear electrodes would be expected at best. This is relevant for design considerations as the electrodes in later generations of the Hybrid device may be spaced closer together with the capability to provide more stimulation channels over a smaller available distance.

Many short-electrode CI users perform remarkably well on speech recognition with just the implant alone. Are short-electrode CI users able to use more electrodes per mm than long-electrode CI listeners? Are short-electrode CI users able to extract as much information as NH listeners in equivalent CI simulations? This is certainly possible for a number of reasons. First, there are differences in audiometric and speech recognition eligibility criteria for the two devices. These lead to differences in factors shown to be correlated with speech recognition performance, including bilateral duration of profound deafness and pre-operative sentence recognition scores (Rubinstein *et al.*, 1999; Leung *et al.*, 2005; Boisvert *et al.*, 2011). The better residual hearing status of short-electrode CI patients means that they have zero duration of total profound deafness, which could translate into better nerve survival and better local stimulation of nerve fibers by electrodes even in the basal region. Short-electrode CI users also have better pre-operative sentence scores and a late-onset or progressive hearing loss. This could translate into a greater capability to make use of the available speech information in the signal than long-electrode CI users, who may not have had sufficient early exposure to the full range of speech information to make use of it even if it is available (Leung *et al.*, 2005). Second, the short-electrode CI differs from the long-electrode CI because it is restricted to the basal 10 mm of the cochlea. It is possible that stimulation efficacy is better in one part of the cochlea than the other due to anatomical differences, such as differences in nerve fiber paths between the basilar-membrane and spiral ganglion in the apex versus the base (Stakhovskaya *et al.*, 2007). However, the psychophysical data related to this explanation is conflicting. Some previous studies have found better electrode discrimination limens in the apex than in the base (Nelson *et al.*, 1995; Zwolan *et al.*, 1997; Henry *et al.*, 2000). More recently, one study found better basal than apical electrode discrimination for some subjects with short durations of deafness (Busby and Clark, 2000), and another study found electrode pitch discrimination to be significantly correlated with physiological spread of excitation, which was narrower in the base than apex (Snel-Bongers *et al.*, 2012). Modulation detection may be another factor correlated with channel usage; one study found better modulation detection

thresholds in the base than the apex for high pulse rates (4000 pps), and vice versa for low pulse rates (250 pps), suggesting that stimulation efficacy may depend on an interaction of stimulation site and stimulation rate (Pfingst *et al.*, 2007).

In the present experiment, consonant recognition was measured as a function of number of stimulation channels for short-electrode CI listeners, long-electrode CI listeners, and NH listeners hearing through an acoustic simulation of a CI. Short-electrode CI listeners were tested with active electrodes allocated to a frequency range of 1063–7938 Hz, the range most often used in everyday programs for the short-electrode array. Long-electrode CI listeners were tested either with active electrodes allocated to a frequency range of 188–7938 Hz, the range most often used in everyday programs for the long-electrode CI, or with basal electrodes allocated to 1063–7938 Hz to match the frequency range provided to short-electrode CI listeners, with the basal electrodes chosen to match those normally allocated with those frequencies. NH listeners were tested with noiseband simulations for frequency ranges of 188–7938 Hz or 1063–7938 Hz. All groups were tested in quiet and in steady background noise at a 0 dB signal-to-noise ratio.

II. METHODS

A. Subjects

Nine adults implanted with the Hybrid short-electrode CI and nine adults implanted with the traditional long-electrode CI from Cochlear Americas (Centennial, CO) participated in this study. All were postlingually deafened, were native speakers of American English, and had at least one year of experience listening with their CI. The general demographic information for the 18 subjects is shown in Table I.

It should be noted that the short-electrode CI subjects were selected from those implanted at the Iowa site of the clinical trial. Those who lost residual hearing completely in the implanted ear were excluded from the study. Though no bias was used in study recruitment of short-electrode CI users, the recruited subjects had a higher average acoustic + electric speech perception score (71%) than the multicenter average (47%) due to the sampling from a Hybrid implantation site with higher than average CNC scores—as well as the exclusion of patients with complete loss of residual hearing from the present study. For reference, the CI-only scores of these subjects are also included in parentheses in Table I. Similarly, the long-electrode CI subjects have a higher score than the national average, due to an emphasis on recruitment for the present study of long-electrode users with better CI-only scores to find the upper limits of performance in this group. Specifically, while the three N24 CI users had a group mean CNC score (45%) that is consistent with the N24 population mean (47%), the 6 Freedom CI users had higher than average group mean scores (81%) than the Freedom population mean (57%) (Balkany *et al.*, 2007).

The short-electrode CI subjects had the S8 electrode array with the equivalent of either the N24 internal receiver ($n = 7$) or the Freedom internal receiver ($n = 2$). In everyday use, seven out of nine short-electrode CI subjects had all 6 electrodes activated, and the remaining two had 1 electrode deactivated. The long-electrode CI subjects had either the N24 internal device ($n = 3$) or the Freedom internal device ($n = 6$).

TABLE I. Demographic information on short- and long-electrode CI subjects (HF S/P HL = high-frequency severe/profound hearing loss, CNC = consonant-nucleus-consonant word score). Subject numbers beginning with SE and CI denote short-electrode and long-electrode subjects, respectively. For post-operative CNC scores for short-electrode CI subjects, monaural CI+HA scores are shown with CI-only scores in parentheses. For the pre- and post-operative speech recognition scores in the table, all subjects were tested with CNC words, except for one subject who was tested pre-operatively with HINT sentences, indicated by the “***” symbol. Post-operative scores were recorded with the CI alone for the long electrode array CI subjects and with the CI and residual acoustic hearing in the ipsilateral ear only (contralateral ear plugged) for the short electrode array CI subjects. These scores were recorded at 12 months after implantation unless otherwise noted as 24 months or 9 months with a “*” or “***” symbol, respectively.

Subject	Age (yr)	Gender	Etiology	Duration of HF S/P HL (yr)	Pre-operative CNC (%)	Post-operative CNC (%)	Duration of CI use (yr)
SE1	63	F	Unknown	unknown	16	66 (56)	6
SE3	35	F	Unknown	15	25	80 (75)	5
SE5	67	M	Noise Exposure	1	15	35 (6**)	3
SE6	50	M	Unknown	2	26	54 (40)	2
SE10	69	F	Unknown	20	19	79 (33**)	4
SE11	48	M	Unknown	6	30	79 (59)	1
SE17	53	M	Unknown	1	37	91 (67)	1
SE21	52	F	Autoimmune	15	24	85 (35)	2
SE27	57	F	Unknown	4	19	72 (62)	2

Subject	Age (yr)	Gender	Etiology	Duration of deafness (yr)	Pre-operative CNC (%)	Post-operative CNC (%)	Duration of CI use (yr)
CI1	81	M	Infection (chronic ear disease)	10	0	62	12
CI2	53	F	Congenital progressive	10	no data	26	10
CI51	72	M	Noise exposure	7	0	66	4
SE4	62	F	Autoimmune	1	no data	94*	3
CI61	72	M	Unknown	1	0***	90	2
CI26	69	F	Meniere’s disease	1	CNT	48	10
CI70	54	F	Unknown	5	2	80	1
CI74	60	F	Unknown	1	4	75	2
CI81	49	F	Hereditary progressive	1	CNT	90	1

Five out of nine long-electrode CI subjects had all 22 electrodes activated; two had 1 electrode deactivated, and two had 2 electrodes deactivated. Four long-electrode CI subjects were bilateral CI users, either with a short-electrode ($n=1$) or long-electrode array ($n=3$) in the contralateral ear. Note that there is some cross-subject variability in the type of processor used in the tested ear, the use of a hearing aid or a CI in the contralateral ear, and other variables such as duration of hearing loss which were not controlled for and which may confound the results.

All subjects were accustomed to using the ACE (Advanced Combination Encoder) processing strategy with either a SPRINT or Freedom speech processor. The details of the CI tested, including device type, ear implanted, number of active electrodes, and clinical map frequency range for each subject is shown in Table II. Table II also shows other devices used in their everyday listening, including a second CI or hearing aid(s).

Twenty NH listeners, ranging from 22 to 30 years of age, were also recruited as controls. All had hearing within 20 dB of the normal standards at octave audiometric frequencies ~0.25–8.0 kHz and were native speakers of American English.

B. Experimental speech processor conditions

For CI subjects, the experimental speech processor conditions consisted of various numbers of stimulation channels or “active” electrodes in the electrode array, with the input sound frequency range divided among these active electrodes and the remaining electrodes deactivated as in

Friesen *et al.* (2001). Specifically, as the number of stimulation channels or active electrodes was decreased, the original frequency range allocated to the full number of electrodes was re-allocated to those remaining electrodes. The active electrodes were selected to be equally spaced throughout the electrode array, with the new frequency allocation for each active electrode spanning approximately the same range of frequencies allocated to the range of electrodes surrounding the active electrode. For short-electrode CI users, the frequency allocations were assigned by the clinical software, based on the lower frequency cutoff chosen by the experimenter. For long-electrode CI users, the frequency allocations were controlled by the experimenter using the SPEAR software. (for detailed active electrode and frequency allocation information for each channel condition, see supplementary material¹). For all CI subjects, the processing strategy was ACE with the number of maxima = number of channels [essentially CIS (Continuous Interleaved Sampling)] and the stimulation mode was monopolar (MP1 + 2). Pulse phase duration and interphase gap were set to 25 and 8 μ s, respectively. Pulse rates were selected to be similar to those in the clinical map, ranging between 720–1200 pulses/s. No additional processing features such as autosensitivity and directional microphone processing were used.

The channel conditions for the short-electrode CI included 6, 5, 4, 3, 2, and 1 stimulation channels, allocated with a frequency range of 1063–7938 Hz. For two short-electrode CI subjects with only five electrodes activated in their personal CI speech processors, only 5-, 4-, 3-, 2-,

TABLE II. Individual device and programming specifics for short- and long-electrode CI subjects. Subject numbers beginning with SE and CI denote short-electrode and long-electrode subjects, respectively. For CI users tested on the short-electrode condition, the subject number, internal device, CI ear, number of activated electrodes and frequency range of the everyday clinical program(s), and hearing aid use are reported. If the number of activated electrodes is less than the full available number, the deactivated electrodes are indicated in parentheses. None of the CI users in this group had a second CI. For CI users tested on the long-electrode and/or long-electrode basal conditions, the same parameters are specified except for the frequency range of the clinical program in the fifth column, which was 188–7938 Hz for all subjects in this group. Instead, the type of second CI is reported if applicable.

Subject	Internal device	CI ear tested	Number of electrodes	Frequency range (Hz)	Hearing aid(s)
SE1	N24	Right	6	688–7938	Contralateral
SE3	N24	Right	6	438–7938 / 628–5603	None
SE5	N24	Right	5 (1)	1063–7938 / 1000–8000	Bilateral
SE6	N24	Right	5 (5)	563–7938	Bilateral
SE10	N24	Right	6	688–7938	Bilateral
SE11	N24	Right	6	688–7938	None
SE17	N24	Right	6	1063–7938	Bilateral
SE21	Freedom	Right	6	688–7938	Bilateral
SE27	Freedom	Right	6	563–7938	Bilateral
Subject	Internal device	CI ear tested	Number of electrodes	Second CI	Hearing aid(s)
CI1	N24	Right	22	n/a	None
CI2	N24	Right	22	n/a	None
CI51	Freedom	Right	22	n/a	Contralateral
SE4	RP8	Right	20 (1,2)	Short-electrode/N24	n/a
CI61	Freedom	Left	22	n/a	None
CI26	N24	Left	20	Long-electrode/N24	n/a
CI70	Freedom	Left	21 (1)	n/a	None
CI74	Freedom	Right	21 (22 ^a)	Long-electrode/freedom	n/a
CI81	Freedom	Right	22	Long-electrode/freedom	n/a

^aThis subject had electrode 22 deactivated in the clinical map, but we were able to activate this electrode in the subject's experimental map without discomfort.

and 1-channel conditions could be tested. All short-electrode CI subjects ($n = 9$) were tested on at least the four most numerous channel conditions in quiet, and a subset of these subjects ($n = 6$; not including subjects SE5, SE10, and SE17) were also tested on at least the four most numerous channel conditions in noise. Because of the time-consuming nature of the test, we were only able to recruit a subset of the subjects tested in quiet for the noise condition. Short-electrode CI users were programmed and tested with the same type of processor as they used in everyday listening (SPRINT or Freedom). Note that the input sound frequency range of 1063–7938 Hz was not necessarily the range that the subjects were used to listening to with their own processor; the subjects varied in their preferred frequency ranges, as shown in Table II.

The long-electrode CI subjects were presented with two different sets of channel conditions, “full” and “basal.” The “full” set of conditions included 22, 8, 6, 5, 4, 3, and 2 stimulation channels, allocated with a frequency range of 188–7938 Hz. The “basal” set of conditions included 15, 6, 5, 4, 3, and 2 stimulation channels, allocated with a frequency range of 1063–7938 Hz and the active electrodes selected from just the basal 15 electrodes. The frequency allocations for the basal conditions were chosen to be identical to the short-electrode CI frequency allocations in the equivalent channel conditions. In cases of subjects with less than 22 electrodes activated in their personal CI speech processors, the 22-channel condition was replaced by the actual number of everyday-use activated channels. Subjects were tested with one or more of the following sets of channel conditions:

the “full” set in quiet with all the channel conditions ($n = 7$; not including CI74), the “basal” set in quiet with all the channel conditions ($n = 6$; not including SE4 and CI81), the “full” set in noise with at least the seven highest channel conditions ($n = 7$; not including CI81), and the “basal” set in noise with at least the seven highest channel conditions ($n = 6$; not including SE4 and CI81). Again, because of the time-consuming nature of the test, which took approximately 7–8 h per condition, we were not able to have all nine subjects complete all four conditions. Instead, we aimed for at least seven out of the nine subjects for the “full” set in quiet and in noise, and at least six out of the nine subjects for the “basal” set in quiet and in noise, with the same subjects in the noise condition selected from those tested in the corresponding quiet condition. Long-electrode CI users were programmed and tested with the SPEAR processor with an input sound frequency range of 188–7938 Hz. The SPEAR processor was used with the long-electrode CI subjects instead of the subjects' own processors because the SPEAR software was more flexible than the clinical software (Custom Sound 1.3, Cochlear Americas, Centennial, CO) available at the start of the study in reallocating frequencies for the long-electrode CI.

For NH simulations of CI listening, noiseband-vocoded speech was generated for each short-electrode CI and long-electrode CI condition. As for CI listeners, conditions included the various channel conditions in both quiet and with white noise at 0 dB signal-to-noise ratio (S/N). Twelve of the NH subjects were tested on simulations of short-electrode CI listening in quiet ($n = 12$) and in noise ($n = 10$; not including two of the original subjects tested in quiet),

and the remaining eight were tested on simulations of long-electrode CI listening in both quiet ($N = 8$) and noise ($N = 7$; not including one of the original subjects tested on this condition in quiet).

For each channel condition, the simulations consisted of the stimuli band-pass filtered into the desired number of stimulation channels or noiseband channels, using the same filter edge frequencies as in the corresponding CI channel condition. The band-pass filters were minimum order Butterworth infinite impulse response filters with roll-off slopes of 96 dB/octave. The envelope was then extracted from the output of each band using a Hilbert-transform and low-pass filtering with a Kaiser finite-impulse response filter at 100 Hz. This envelope was then used to modulate a white noise signal, and the resulting output was then band-pass filtered using the same set of filters used to extract the envelopes for each band.

Channel conditions were tested in quiet and in speech-shaped white noise with a S/N ratio of 0 dB. For CI listeners in the noise conditions, noise (noise low-pass filtered at 800 Hz at 10 dB/octave) was added to the speech signals at a 0 dB S/N. For the NH listeners, this same speech-shaped noise was added to the speech signals prior to the simulation processing.

C. Stimuli and procedures

For each experimental speech processor condition, threshold (T) and comfort (C) levels were measured for each electrode using standard clinical procedures, and the live stimulation mode was then used to verify through experimenter-patient interaction that speech was loud enough and comfortable to listen to. For short-electrode CI listeners, clinical software (Custom Sound) was used to program the SPRINT or Freedom processor; the processor type was selected based on the patient's internal device. For long-electrode CI listeners, research software (SeedSpeak, CRC Hear, Sydney, Australia) was used to program a SPEAR processor. Once stimulation parameters were determined, they were saved to the lab experimental speech processor to be used during the testing.

The full-channel condition in quiet corresponding most closely to the listener's personal processor programming was always presented first to familiarize the subject with the task and if applicable, a different speech processor frequency range. After at least two runs with the full-channel condition, each condition within a set of channel conditions was presented to each listener in random order. Subjects did not have additional listening experience in each condition beyond that provided during testing. Each set of channel conditions was presented in different sessions conducted over separate days/visits. At subsequent sessions after the initial programming session, the live stimulation mode was always used prior to testing to verify that the T and C levels were still appropriate for stimulation, and levels were adjusted if needed.

All testing was conducted in a sound-treated booth. For CI listeners, stimuli were presented via a direct-electric connection to the implant processor; no acoustic information was provided to either short-electrode or long-electrode CI subjects. For NH listeners, stimuli were presented monaurally via Sennheiser HD-25 headphones (Old Lyme, CT) to the subject's preferred

ear. The direct-electric stimulation level for CI listeners was selected to give a current output equivalent to the current output delivered for an acoustic speech presentation level of 65 dB SPL (sound pressure level) processed through the microphone input, and this output was verified to fit within the dynamic range of the processor similar to an acoustic input at 65 dB SPL. The headphone presentation level for NH listeners was presented at a comfortable level whereby the high-frequency portion of the speech (1000–8000 Hz) was at the same level as wideband speech presented between 65 and 75 dB SPL. All stimuli were stored on a Macintosh G4 (Apple, Inc. Cupertino, CA) computer and output through a DigiDesign (Burlington, MA) 16-bit digital-to-analog converter.

Subjects were tested on recognition of 16 consonants $/\beta \delta \gamma \pi \tau \kappa \mu \nu \phi \sigma \varpi \zeta \tau\eta \sigma\eta \tau\eta\zeta \delta Z/$ displayed as $/b d g p t k m n f s v z th sh thz dz/$. These consonants were presented in an a/C/a format and spoken by four different talkers (Turner *et al.*, 1995), for a total of 64 stimuli in a run. In each run, the 64 stimuli were presented in random order using code written in MATLAB (The Mathworks, Natick, MA), and subjects were instructed to indicate which consonant they thought they heard by touching the corresponding letter on a touchscreen display. Feedback was provided. At least two runs or eight presentations of each consonant were presented for each subject and condition.

Improvements in percent correct score of 5% or more between two consecutive runs were considered to reflect learning effects. In such cases, runs were conducted until improvements in scores were no longer observed, and results from all but the last three runs were discarded.

Statistical analyses were performed using statistical software (SAS 9.2, Cary, NC). Significance was evaluated using two-sided statistical tests at a 5% significance level, without adjustments for Type I error. For assessment of effects of channel conditions and quiet versus noise conditions across conditions and groups, a likelihood-based mixed effects model was used instead of analysis of variance (ANOVA) to control for conditions with missing data. Specifically, the SAS MIXED procedure was employed to evaluate differences in speech scores across conditions, with any missing data was assumed to be missing at random.

The consonant confusion matrices were calculated for each subject and channel condition. Place, manner, and voicing confusions and information transmission were analyzed for each subject and channel condition as in Miller and Nicely (1955). Information transmission indicates the correlation between the variance in the input (CI signal) and the variance in the output (subject response), and is a measure of the amount of information in the observer response that is dependent on the information present in the stimuli as defined by the feature analysis, and not due to random noise in the response.

III. RESULTS

Figures 1–3 show the consonant recognition results for the short-electrode, long-electrode-full, and long-electrode-basal conditions, respectively. Within each figure, panels (a) and (b) at top show recognition performance for CI users in quiet and 0-dB SNR noise, respectively. Panels (c) and (d) at

FIG. 1. Short-electrode CI and equivalent NH simulation consonant recognition scores in quiet and in noise. Thin lines are the individual subject scores as a function of the number of stimulation channels; individual CI subject numbers are indicated by different symbols and listed in the figure legend. Heavy black or gray lines show the mean and standard deviation of group data, respectively. (a) Short-electrode condition performance for CI users in quiet ($N=9$). (b) Short-electrode condition performance for CI users in 0 dB SNR noise ($N=6$; not including SE5, SE10, SE17). (c) NH performance in short-electrode simulation condition in quiet (1063–7938 Hz, divided into 1–6 channels; $N=12$). (d) NH performance in short-electrode simulation condition in noise [$N=10$; not including 2 subjects from (c)].

bottom show results for CI simulations presented to NH listeners in quiet and in 0-dB SNR noise. For Fig. 3, only the results for CI users are shown in panels (a) and (b); the equivalent CI simulations are the same as for Fig. 1.

In each panel, recognition scores are plotted as a function of the number of stimulation channels. The thin lines are data from individual subjects and the heavy solid lines represent the group average, with vertical bars indicating plus and minus one standard deviation. Individual CI subjects are shown as different symbols in panels (a) and (b) (for group mean and standard deviation values, see supplementary material¹).

Figure 4 shows the averaged results plotted for the different implant conditions compared with the simulations.

The top panels [Figs. 4(a) and 4(b)] show a comparison of the short-electrode and long-electrode-basal conditions and equivalent NH simulations for conditions with frequency ranges of 1063–7938 Hz in quiet and noise. The bottom panels [Figs. 4(c) and 4(d)] compare the full long-electrode CI and equivalent NH simulations for frequency ranges from 188–7938 Hz in quiet and noise.

A. Performance variation as a function of the number of stimulation channels and in quiet versus noise

A maximum-likelihood based mixed effects model was used to evaluate the effects of number of stimulation channels and quiet versus noise conditions. A significant effect of

FIG. 2. Long-electrode CI and equivalent NH simulation consonant recognition scores in quiet and in noise. Plotted as in Fig. 1. (a) Long-electrode condition performance for CI users in quiet ($N=7$; not including CI74). (b) Long-electrode condition performance for CI users in 0 dB SNR noise ($N=7$; not including CI81). (c) NH performance in long-electrode simulation condition in quiet (188–7938 Hz, divided into 1–22 channels; $N=8$). (d) NH performance in long-electrode simulation condition in noise [$N=7$; not including one subject from (c)].

FIG. 3. Long-electrode CI with 15 basal channels only consonant recognition scores in quiet and in noise. Plotted as in Fig. 1, except that NH simulation results are the same as for the bottom panels in Fig. 1 and are not shown again here. (a) Long-electrode-basal condition performance for CI users in quiet ($N=6$; not including SE4 and CI81). (b) Long-electrode-basal condition performance for CI users in 0 dB SNR noise ($N=6$; not including SE4 and CI81).

both number of channels and quiet versus noise condition was seen for all subject groups ($P < 0.0001$). No significant interaction between number of channels and quiet versus noise conditions was seen in any of the subject groups. Further *post hoc* pairwise analyses were conducted to estimate the number of channels used by each subject group in quiet and in noise; these results are summarized in Table III and described below.

For short-electrode CI users, significant differences ($P < 0.05$) were seen for 2 versus 3, 4, 5, and 6 channels, and for 3 versus 4 channels in quiet ($N=9$; 1 and 2 data points missing from the 2- and 6-channel conditions, respectively). In noise, significant differences were seen for 3 versus 4, 5, and 6 channels, and for 5 versus 6 channels ($N=6$; 1 and 3 data points missing from the 6- and 2-channel conditions, respectively). These results suggest that short-electrode subjects can use up to 4 channels in quiet, and possibly up to 6 channels in noise (or more, since no more than 6 channels could be tested). However, the latter result should be considered preliminary because of the smaller number of subjects for the 6-channel condition in noise. The argument for the use of 6 channels in noise is also weakened by the lack of significant differences between 4 and 5 channels and 4 and 6

channels in noise. Certainly the results show that short-electrode CI users use at least 4 channels in noise.

For long-electrode CI users in the “full” map condition, significant differences were seen for 2 channels versus 3, 4, 5, 6, and 8 channels, 3 channels versus 4, 5, 6, and 8 channels, and 2–8 channels versus 22 channels in quiet ($N=7$). In noise, significant differences were seen between 2 channels versus 3, 4, 5, 6, and 8 channels, 3 channels versus 5, 6, and 8 channels, and 2–8 channels versus 22 channels ($N=7$). It is not known whether the significant difference seen for all of the channel conditions versus the 22-channel condition was due to subjects obtaining more information from 22 channels, or due to the greater amount of experience with the 22-channel condition which was most like the patient’s own map. If the difference between the 22-channel condition and the other conditions is due to greater information usage, the results indicate that long-electrode CI users in this study can use more than 8 channels in quiet and noise. Alternatively, since no differences were seen between 4, 5, 6, and 8 channels, the difference in the 22 channel condition and the other channel conditions may have been due to more experience with 22 channels. If that is the case, then the results suggest that long-electrode CI users can use up to 4

FIG. 4. Comparison of the mean group consonant recognition results for the various device conditions. Solid lines with circles indicate simulated results in NH listeners. Dashed lines indicate CI subject results, with different symbols indicating different device conditions. (a) Performance on short-electrode CI (triangles), long-electrode-basal CI (squares), and equivalent NH simulation (circles; 1063–7938 Hz, divided into 1–6 channels) conditions in quiet. (b) Performance on short-electrode CI, long-electrode-basal CI, and equivalent NH simulation conditions in 0 dB SNR noise. (c) Performance on long-electrode CI (diamonds) and equivalent NH simulation (circles; 188–7938 Hz, divided into 1–22 channels) conditions in quiet. (d) Performance on long-electrode CI and equivalent NH simulation conditions in noise.

TABLE III. Within-group pairwise *post hoc* channel comparisons of consonant recognition scores for each group and condition. Significant results ($P < 0.05$) are shown in bold text. A “—” indicates no comparison could be made for that pair.

Group	Short-electrode		Long-electrode		Long-electrode-basal		Short-electrode simulation		Long-electrode simulation	
Channel pairs	Quiet	Noise	Quiet	Noise	Quiet	Noise	Quiet	Noise	Quiet	Noise
2 vs 3	0.0111	0.1719	0.0040	0.0083	0.0056	0.0005	<1e-4	—	0.1505	0.5785
2 vs 4	0.0041	0.1087	0.0021	0.0027	0.0010	0.0006	<1e-4	—	0.0018	0.0019
2 vs 5	0.0085	0.1206	0.0026	0.0012	0.0024	0.0004	<1e-4	—	0.0003	<1e-4
2 vs 6	0.0231	0.0920	0.0017	0.0021	0.0021	0.0018	<1e-4	—	<1e-4	0.0006
3 vs 4	0.0234	0.0303	0.0169	0.0645	<1e-4	<1e-4	0.1246	0.0030	0.0460	0.0122
3 vs 5	0.0648	0.0254	0.0144	0.0195	<1e-4	<1e-4	0.0014	0.0010	0.0072	0.0015
3 vs 6	0.1178	0.0028	0.0074	0.0258	0.0292	0.0422	0.0030	0.0003	0.0026	<1e-4
4 vs 5	0.1572	0.6968	0.1750	0.2923	0.6686	0.1744	0.0006	0.4631	0.0260	0.0854
4 vs 6	0.4804	0.0672	0.1418	0.0949	0.7537	0.8097	0.0041	0.0186	0.0093	0.0271
5 vs 6	0.6916	0.0098	0.6916	0.2534	0.7260	0.5680	0.2368	0.0013	0.0293	0.1747
2 vs 8	—	—	0.0045	0.0018	—	—	—	—	<1e-4	<1e-4
3 vs 8	—	—	0.0235	0.0473	—	—	—	—	0.0005	0.0001
4 vs 8	—	—	0.3569	0.1587	—	—	—	—	0.0042	0.0056
5 vs 8	—	—	0.8599	0.7516	—	—	—	—	0.0115	0.0156
6 vs 8	—	—	0.3770	0.6539	—	—	—	—	0.6910	0.7297
2 vs 15	—	—	—	—	0.0015	0.0027	—	—	—	—
3 vs 15	—	—	—	—	0.0191	0.0193	—	—	—	—
4 vs 15	—	—	—	—	0.2011	0.2034	—	—	—	—
5 vs 15	—	—	—	—	0.2263	0.3037	—	—	—	—
6 vs 15	—	—	—	—	0.1035	0.1355	—	—	—	—
2 vs 22	—	—	0.0029	0.0013	—	—	—	—	<1e-4	<1e-4
3 vs 22	—	—	0.0094	0.0072	—	—	—	—	<1e-4	<1e-4
4 vs 22	—	—	0.0299	0.0017	—	—	—	—	<1e-4	<1e-4
5 vs 22	—	—	0.0223	0.0270	—	—	—	—	<1e-4	0.0002
6 vs 22	—	—	0.0477	0.0122	—	—	—	—	0.0235	<1e-4
8 vs 22	—	—	0.0365	0.0177	—	—	—	—	0.0156	0.0002

channels in quiet, and may use either 4 or 5 channels in noise (since the difference between 3 and 5 channels was significant, but the differences between 3 and 4 channels, and between 4 and 5 channels, were not significant).

For long-electrode CI users in the “basal” condition, significant differences were seen for 2 versus 3, 4, 5, 6, and 15 channels, and 3 versus 4, 5, 6, and 15 channels in both quiet and noise ($N=6$). This suggests that long-electrode CI users can use 4 channels in both quiet and noise in the basal condition.

For NH subjects listening to short-electrode CI simulations, significant differences were seen for 2 channels versus 3, 4, 5, and 6 channels, 3 channels versus 5 and 6 channels, and 4 channels versus 5 and 6 channels ($N=12$). In noise, significant differences were seen for 3 channels versus 4, 5, and 6 channels, 4 channels versus 6 channels, and 5 channels versus 6 channels ($N=10$; 2-channel condition not presented). These results suggest that NH listeners can use the equivalent of 5 short-electrode or basal long-electrode channels in quiet and at least 6 channels in noise (since more channels were not tested).

For NH subjects listening to long-electrode CI simulations, significant differences were seen for 2 channels versus 4, 5, 6, and 8 channels, 3 channels versus 4, 5, 6, and 8 channels, 4 channels versus 5 and 6 channels, 5 channels versus 6 channels, and 2–8 channels versus 22 channels ($N=8$). In noise, significant differences were seen for 2 channels versus 4, 5, 6, and 8 channels, 3 channels versus 4, 5, 6, and 8 channels, 4 channels

versus 6 channels, and 2–8 channels versus 22 channels ($N=7$). Because experience is not a confound for NH listeners, these results indicate that NH listeners can use the equivalent of 22 long-electrode channels in both quiet and noise.

B. Comparison of CIs and acoustic simulations

Another question was whether short-electrode CI users can use the information provided as effectively as NH listeners in equivalent CI simulations. Speech recognition scores were compared between these groups for each channel condition in quiet and in noise (for detailed statistical results, see supplementary material¹).

A pairwise comparison of channels 2, 3, 4, 5, and 6 between short-electrode CI users and NH simulations of short-electrode CI showed significant differences for the 3, 5, and 6 channel conditions in quiet and for the 3, 4, 5, and 6 channel conditions in noise. A pairwise comparison between long-electrode CI users and NH simulation results in the full condition showed significant differences for all channels studied in quiet (2, 3, 4, 5, 6, 8, 22), but no significant differences were seen in noise except for 8 and 22 channels. A pairwise comparison between long-electrode CI users and equivalent NH simulation results in the basal condition showed significant differences for all channels studied in quiet (2, 3, 4, 5, and 6) but no significant differences were seen in noise except for a barely significant difference at 3

channels ($P=0.0485$). These trends are reflected in the group mean comparisons in Fig. 4.

C. Comparison of short-electrode and long-electrode CI conditions

The final question was how short-electrode CI users compare to long-electrode CI users in extracting information for equivalent channel conditions. A one-way ANOVA model was used to compare the speech recognition scores between these groups for each channel condition in quiet and in noise (for detailed statistical results, see supplementary material¹).

When short-electrode CI performance was compared to long-electrode CI performance in the full condition, a significant difference was seen only for the 5-channel condition in noise, in which long-electrode CI users performed better than short-electrode CI users ($P=0.0067$). The more relevant comparison may be for the short-electrode compared to the long-electrode-basal-electrode condition, as this condition provided the same frequency range of information (1063–7938 Hz). The full condition, while providing information most similar to what long-electrode CI users were accustomed to (188–7938 Hz), provided a wider frequency range (and thus more speech information) than the short-electrode condition (1063–7938 Hz) and thus may not be directly comparable to the short-electrode condition. In this comparison between short-electrode and long-electrode-basal conditions, no significant differences were seen in quiet or in noise.

Information transmission analysis is an alternative way to compare the three CI channel conditions while controlling for the amount of information based on place, manner, and voicing feature classifications defined as in Miller and Nicely (1955). Consistent with the speech perception data, Fig. 5 shows that information transmission was similar for all three device conditions in CI users. A one-way ANOVA across the 2-, 3-, 4-, 5-, and 6-channel conditions showed few significant differences in information transmission between any of the device conditions in quiet [Fig. 5(a)] or in noise [Fig. 5(b)], even though the inter-electrode spacings for stimulation electrodes in the equivalent conditions are wider in the long-electrode and long-electrode-basal conditions than the short-electrode condition (Table III). There was just one barely significant difference in information transmission between long-electrode CI users in the full and basal conditions at 2 channels in quiet ($P=0.0431$), and one

significant difference between short-electrode CI users and long-electrode CI users in the full condition at 5 channels in noise ($P=0.0132$).

D. Place, manner, and voicing analyses

The transmission of individual features may indicate the differences in the type of information transmitted for NH listeners versus CI users. Place, manner, and voicing confusion scores were calculated as a function of the number of stimulation channels, and detailed results in quiet are shown in Fig. 6. Similar trends were observed in noise, so are not shown. The three columns of data represent long-electrode, short-electrode, and long-electrode-basal conditions. The three rows represent analyses divided into voicing, manner, and place cues. The individual results for CI users are plotted as thin lines and compared with the mean and standard deviation of the results for NH listeners with the equivalent CI simulations plotted as heavy solid lines.

A few qualitative trends are apparent in Fig. 6. First, note that NH listeners can achieve higher feature recognition scores in the long-electrode condition because of the greater extent of low-frequency information in that condition (188–7938 Hz) compared to the other two conditions (1063–7938 Hz). This is particularly evident for voicing information, as would be predicted. Second, within each feature and device condition, there are a few CI subjects that perform as well as NH listeners on the equivalent simulation, except in the long-electrode and long-electrode-basal conditions on manner. Short-electrode CI users show greater variability on the perception of manner cues, but half of these subjects approach the same level of performance as NH listeners with the full six stimulation channels. Third, a few CI users in each device condition had higher feature recognition performance when intermediate numbers of channels (3, 4, or 5) were activated rather than when all channels were activated, especially for voicing (top row of Fig. 6). Some short-electrode subjects even outperformed NH listeners on voicing recognition scores in the intermediate channel conditions (top row, middle column), a trend also observed in Friesen *et al.* (2001). This higher performance could be attributed to the greater experience CI subjects have compared to NH subjects in extracting these features with minimal channel information, such as from temporal cues, or alternatively, higher maximum performance for CI users compared to the observed NH simulation performance. The drop in score with more stimulation

FIG. 5. Comparison of information transmission for the three device conditions in quiet (a) and in noise (b). Different line styles and symbols indicate different device conditions.

FIG. 6. Place manner voicing results for the three device conditions in quiet. Thin dashed lines indicate the feature recognition performance of individual CI users as a function of the number of channels. Heavy solid lines with circles indicate the mean and standard deviation of NH subjects in equivalent CI simulations. Left column: long-electrode condition (188–7938 Hz). Middle column: short-electrode condition (1063–7938 Hz). Right column: long-electrode-basal condition with only 15 basal electrodes activated (1063–7938 Hz). Voicing, manner, and place are represented in the top, middle, and bottom rows, respectively.

channels may indicate a negative effect of channel interaction at higher channel conditions. While NH simulation performance can theoretically be considered the maximum possible performance, in practice NH performance may be limited by the short duration of training time for the NH listeners compared to the CI users.

IV. DISCUSSION

The number of usable stimulation channels in each device can be defined as the minimum number of channels N that results in optimal speech perception. Specifically, N is found as follows. First, among all channel conditions, determine the number of channels N_i that resulted in the highest average speech perception score, where i is the channel condition index. Then, find all the other numbers of channels ($N_{i1}, N_{i2}, \dots, N_{ij}$) with scores statistically indistinguishable from N_i . N will be the minimum of these channels or $\min(N_{i1}, N_{i2}, \dots, N_{ij})$.

Based on this definition, the preliminary findings suggest that all three device conditions—short-electrode, long-electrode (full), and long-electrode-basal—provided at least four usable stimulation channels. The distances between usable stimulation channels, however, differ markedly across device conditions. The distances between four stimulation channels in the Hybrid short-electrode CI array are much smaller than the distances between four stimulation channels in the long-electrode CI array in either the full or basal electrode condition.¹ The length of the part of the short-electrode CI array containing the electrode contacts is 4 mm, the length of the long electrode CI array in the basal-electrode-only condition is 8.5 mm with a N24 straight array or 11 mm with a Contour Advance array, and the length of the long electrode CI array in the full condition is 17 mm. If CI subjects are using only four channels in all of the conditions, this translates to an average of usable 1 stimulation channel/mm

in the short-electrode condition, 0.47 or 0.36 channels/mm in the long-electrode-basal condition, and 0.24 channels/mm in the long-electrode-full condition.

Surprisingly, short-electrode CI users performed as well as long-electrode CI users in the basal electrodes-only condition with the same amount of information (1063–7938 Hz). No significant differences were seen between these two groups in quiet or in noise for any channel condition. In other words, total information transmission per millimeter is much higher for our sample of short-electrode CI users than for our sample of long-electrode CI users (see Sec. IIC for definition of information transmission). The average maximum information transmission for the short-electrode condition in quiet is calculated to be 2.72 bits in the 6-channel condition, yielding a maximum information/distance of 0.68 bits/mm. For the long-electrode-basal condition, 2.71 bits in the 15-channel condition yields a maximum information/distance of 0.32 bits/mm. This is equivalent to at least double the information transmission per distance in the short-electrode condition compared to the long-electrode condition. Even for the full long-electrode condition, the maximum information per millimeter is 3.08 bits in 17 mm or 0.18 bits/mm. In noise, information transmission per millimeter drops slightly but remains highest for the short-electrode condition at 0.62 bits/mm, followed by the long-electrode-basal condition at 0.30 bits/mm and long-electrode condition at 0.16 bits/mm. Thus, the group of short-electrode CI subjects received approximately double the information transmitted per millimeter than the long-electrode CI subjects.

Compared to NH subjects listening to the equivalent CI simulations, Hybrid short-electrode CI subjects performed similarly for the 4-channel condition in quiet, but underperformed relative to NH subjects for all channel conditions in noise. In noise, the difference in performance between short-electrode CI and NH listeners in the equivalent simulation

increased so that short-electrode CI users no longer performed as well as NH listeners for any channel condition. In contrast, long-electrode CI users in both the long-electrode and long-electrode-basal conditions underperformed relative to NH listeners for all channel conditions in quiet, but performed similarly to NH listeners for the lower channel conditions in noise.

How can short-electrode CI subjects perform similarly in the equivalent channel conditions as long-electrode CI subjects, despite the smaller channel spacing, as suggested by these findings? Some possibilities include greater local nerve survival in Hybrid short-electrode CI patients due to differences in candidacy criteria (especially in the low-frequency pre-operative thresholds), differences in the geometry of the arrays—a straight insertion for the short-electrode CI versus curvature around the cochlear turns for the long electrode CI, or the use of soft surgery techniques. Soft surgery techniques lead to better preservation of low-frequency residual hearing and are likely to also reduce trauma and the resulting fibrosis, bone growth, and other morphological changes in the cochlea in the vicinity of the CI electrode, which may in turn allow more effective, localized stimulation of the nearby neural structures. Indeed, a recent retrospective study of residual hearing preservation outcomes with soft surgery techniques in long-electrode CI users showed significantly better speech recognition scores with the CI alone for those with preserved hearing compared to those without preserved hearing (Carlson *et al.*, 2011).

One factor associated with performance, pre-operative word recognition score, was significantly different between the short-electrode CI and long-electrode CI subject groups ($P < 0.001$, Wilcoxon rank-sum test). Although this factor is associated with the implantation of a short-electrode CI instead of long-electrode CI because it reflects the better low-frequency residual hearing status, this suggests another interesting possibility that continued exposure and ability to listen to speech between the loss of hearing and implantation may be a factor in channel usage.

However, these factors do not explain why long-electrode subjects in the basal condition perform similarly in the equivalent channel conditions as in the full condition, despite the closer channel spacing in the basal condition. Another potential factor that may better explain this effect is differences between stimulation in the cochlear base versus cochlear apex. Could better efficacy of electrical stimulation or better neural tuning in the cochlear base than the cochlear apex explain the higher information transmission per millimeter for the short-electrode and long-electrode basal condition? While the majority of electrode discrimination studies suggest that electrodes are more discriminable in the apex than the base, the number of usable stimulation channels may also depend on other factors such as modulation detection or gap detection, and modulation detection has been shown to be better for basal than apical stimulation, at least at very high pulse rates (Pfungst *et al.*, 2007). Thus, this question remains unanswered until an equivalent long-electrode-apical condition can be tested using stimuli with more low-frequency information than consonants, such as vowels, to minimize the need for long-electrode CI users to adapt to severe frequency-place shifts.

One last possibility is that there may be a limit on the number of electric channels that can be processed cognitively, independent of electrode spacing. If so, then this has implications for the design of future electrode arrays, which may be shortened if electrode interactions and usage do not depend on electrode spacing. While this would introduce a frequency-place mismatch, several studies suggest that CI users can adapt at least partially to new frequency allocations (Rosen *et al.*, 1999; Fu *et al.*, 2005; Reiss *et al.*, 2008). In addition, this study shows that short-electrode CI subjects perform nearly as well as NH subjects in equivalent, *unshifted* simulations, even though the short-electrode CI has a severe frequency-place mismatch with acoustic stimulation from 1063–7938 Hz delivered to electrodes stimulating cochlear place frequencies from 4500 Hz and up (Greenwood, 1990). However, it has been observed of many of these short-electrode patients that pitch perception has adapted over months of implant experience by as much as two octaves to approach the acoustically provided frequencies (Reiss *et al.*, 2007, 2008); it is possible that pitch plasticity may somehow compensate for the frequency-place mismatch or even increase the number of usable stimulation channels.

An interesting observation from the breakdown of the scores into place, manner, and voicing is that some CI subjects appeared to perform best on these features with intermediate numbers of channels activated, rather than with all channels activated. This was observed for both short-electrode and long-electrode CI users, and has also been observed in other studies that CI users sometimes perform better with half of their electrodes activated (e.g., best performing CI subjects in Friesen *et al.*, 2001). Perhaps contrary to the above, this suggests that channel interaction may cause interference for some CI users when more than half of the available electrodes are activated.

Generally, the finding that performance asymptotes at a lower number of channels in quiet than in noise for short-electrode CI users is consistent with the previous findings in long-electrode CI users by Friesen *et al.* (2001). The increased difficulty of speech recognition in background noise increased the utility of additional channels in both studies. In addition, similar to Friesen *et al.* (2001), NH listeners continued to improve performance and asymptoted at a higher number of channels than CI subjects, indicating a better ability to make use of the increase in information with more channels.

The lack of significant differences between 4–8 channels in long-electrode CI subjects is consistent with Fishman *et al.* (1997), but differs from Friesen *et al.* (2001) which found significant differences between 4 channels and 7 channels. One likely reason for the difference may be the statistical power of numbers, as Friesen *et al.* (2001) recruited 19 subjects, while 11 subjects were recruited in Fishman *et al.* (1997), and 6–7 subjects were recruited in this study for each long-electrode condition. Interestingly, long-electrode CI subjects in this study also showed significant differences between the highest channel condition (22 channels in the current study) and the lower channel conditions, which were not seen in the previous studies for the highest channel condition tested (20 channels in the previous studies). One possibility is that this discrepancy is due to a lack of comparable

experience with the new frequency allocations of the lower channel conditions in the current study; the subjects in Fishman *et al.* (1997) but not Friesen *et al.* (2001) had at least 2 days of experience with the experimental allocations before testing. Another major difference is that the subjects in Friesen *et al.* (2001) and Fishman *et al.* (1997) used bipolar (BP +1) rather than monopolar stimulation. This discrepancy may also be due to recent changes in implant candidacy criteria and the resulting differences in nerve survival in patients in the current study compared to the previous studies, as well as improvements in CI array design and speech processing.

V. CONCLUSIONS

The preliminary findings of this study suggest that consonant recognition as a function of the number of stimulation channels is similar in both quiet and noise for short-electrode CI users and long-electrode CI users in both the full condition and with only the basal electrodes activated. Subjects in all three device conditions reached plateau or near-plateau performance with at least four stimulation channels, and short-electrode CI users may have even benefited from up to six stimulation channels in noise. This result is surprising given the differences in electrode spacing of the four stimulation channels between the device conditions, and suggests that differences in pre-operative patient history, electric stimulation in the cochlear base versus the apex, and the number of channels that can be processed cognitively may also interact with the effects of inter-electrode spacing in determining the channel benefit, at least for the range of electrode numbers and inter-electrode spacing tested in this study.

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¹See supplementary material at <http://dx.doi.org/10.1121/1.4757735> for details of active electrodes and frequency allocations for each channel condition, for group mean and standard deviation values, and for details of pair-wise comparison results across groups.

Balkany, T., Hodges, A., Menapace, C., Hazard, M. S., Driscoll, C., Gantz, B., Kelsall, D., Luxford, W., McMenomey, S., Neely, J. G., Peters, B., Pillsbury, H., Roberson, J., Schramm, D., Telian, S., Waltzman, S., Westerber, B., and Payne, S. (2007). "Nucleus Freedom North American clinical trial," *Otolaryngol.-Head Neck Surg.* **136**, 757–762.

Boisvert, I., McMahon, C. M., Tremblay, G., and Lyxell, B. (2011). "Relative importance of monaural sound deprivation and bilateral significant hearing loss in predicting cochlear implantation outcomes," *Ear Hear.* **32**(6), 758–766.

Busby, P. A., and Clark, G. M. (2000). "Electrode discrimination by early-deafened subjects using the cochlear limited multiple-electrode cochlear implant," *Ear. Hear.* **21**(4), 291–304.

Carlson, M. L., Driscoll, C. L. W., Gifford, R. H., Service, G. J., Tombers, N. M., Hughes-Borst, B. J., Neff, B. A., and Beatty, C. W. (2011). "Implications of minimizing trauma during conventional cochlear implantation," *Otol. Neurotol.* **32**, 962–968.

Fishman, K., Shannon, R. V., and Slattery, W. H. (1997). "Speech recognition as a function of the number of electrodes used in the SPEAK cochlear implant speech processor," *J. Speech Lang. Hear. Res.* **40**, 1201–1215.

Friesen, L. M., Shannon, R. V., Baskent, D., and Wang, X. (2001). "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**, 1150–1163.

Fu, Q. J., Nogaki, G., and Galvin, J. J., III. (2005). "Auditory training with spectrally shifted speech: Implications for cochlear implant patient auditory rehabilitation," *J. Assoc. Res. Otolaryngol.* **6**(2), 180–189.

Gantz, B. J., and Turner, C. W. (2003). "Combining acoustic and electric hearing," *Laryngoscope* **113**, 1726–1730.

Gantz, B. J., and Turner, C. W. (2004). "Combining acoustic and electric speech processing: Iowa/Nucleus Hybrid Implant," *Acta Otolaryngol.* **24**, 344–347.

Gfeller, K., Olszewski, C., Turner, C. W., and Gantz, B. (2006). "Music perception with cochlear implants and residual hearing," *Audiol. Neuro-Otol.* **11**, S1, 12–15.

Greenwood, D. (1990). "A cochlear frequency-position function for several species—29 years later," *J. Acoust. Soc. Am.* **87**(6), 2592–2605.

Henry, B. A., McKay, C. M., McDermott, H. J., and Clark, G. M. (2000). "The relationship between speech perception and electrode discrimination in cochlear implantees," *J. Acoust. Soc. Am.* **108**(3), 1269–1280.

Leung, J., Wang, N. Y., Yeagle, J. D., Chinnici, J., Bowditch, S., Francis, H. W., and Niparko, J. K. (2005). "Predictive models for cochlear implantation in elderly candidates," *Arch. Otolaryngol. Head Neck Surg.* **131**, 1049–1054.

Miller, G. A., and Nicely, P. E. (1955). "An analysis of perceptual confusions among some English consonants," *J. Acoust. Soc. Am.* **27**, 338–352.

Nelson, D. A., Van Tasell, D. J., Schroder, A. C., Soli, S., and Levine, S. (1995). "Electrode ranking of 'place pitch' and speech recognition in electric hearing," *J. Acoust. Soc. Am.* **98**(4), 1987–1999.

Pfingst, B. E., Xu, L., and Thompson, C. S. (2007). "Effects of carrier pulse rate and stimulation site on modulation detection by subjects with cochlear implants," *J. Acoust. Soc. Am.* **121**, 2236–2246.

Reiss, L. A., Gantz, B. J., and Turner, C. W. (2008). "Cochlear implant speech processor frequency allocations may influence pitch perception," *Otol. Neurotol.* **29**(2), 160–167.

Reiss, L. A. J., Turner, C. W., Erenberg, S. R., and Gantz, B. J. (2007). "Changes in pitch with a cochlear implant over time," *J. Assoc. Res. Otolaryngol.* **8**(2), 241–257.

Rosen, S., Faulkner, A., and Wilkinson, L. (1999). "Adaptation by normal listeners to upward spectral shifts of speech: Implications for cochlear implants," *J. Acoust. Soc. Am.* **106**, 3629–3636.

Rubinstein, J. T., Parkinson, W. S., Tyler, R. S., and Gantz, B. J. (1999). "Residual speech recognition and cochlear implant performance: Effects of implantation criteria," *Am. J. Otol.* **20**, 445–452.

Shannon, R. V., Zeng, F.-G., Kamath, V., Wygonski, J., and Ekelid, M. (1995). "Speech recognition with primarily temporal cues," *Science* **270**, 303–304.

Snel-Bongers, J., Briaire, J. J., Vanpouchke, F. J., and Frijns, J. H. (2012). "Spread of excitation and channel interaction in single- and dual-electrode cochlear implant stimulation," *Ear. Hear.* **33**(3), 367–376.

Stakhovskaya, O., Sridhar, D., Bonham, B. H., and Leake, P. A. (2007). "Frequency map for the human cochlear spiral ganglion," *J. Assoc. Res. Otolaryngol.* **8**, 220–233.

Turner, C. W., Souza, P. E., and Forget, L. N. (1995). "Use of temporal envelope cues in speech recognition by normal and hearing-impaired listeners," *J. Acoust. Soc. Am.* **97**, 2568–2576.

Xu, L., and Pfingst, B. E. (2008). "Spectral and temporal cues for speech recognition: Implications for auditory prostheses," *Hear. Res.* **242**(1–2), 132–140.

Zwolan, T. A., Collins, L. M., and Wakefield, G. H. (1997). "Electrode discrimination and speech recognition in postlingually deafened adult cochlear implant subjects," *J. Acoust. Soc. Am.* **102**(6), 3673–3685.