Effects of stimulus level and rate on psychophysical thresholds for interleaved pulse trains in cochlear implants

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This study examined channel interactions using interleaved pulse trains to assess masking and potential facilitative effects in cochlear-implant recipients using clinically relevant stimuli. Psychophysical thresholds were measured for two adjacent mid-array electrodes; one served as the masker and the other as the probe. Two rates representative of those found in present-day strategies were tested: 1700 and 3400 pulses per second per channel. Four masker levels ranging from sub-threshold to loud-but-comfortable were tested. It was hypothesized that low-level maskers would produce facilitative effects, shifting to masking effects at high levels, and that faster rates would yield smaller masking effects due to greater stochastic neural firing patterns. Twenty-nine ears with Cochlear or Advanced Bionics devices were tested. High-level maskers produced more masking than low-level maskers, as expected. Facilitation was not observed for sub-threshold or threshold-level maskers in most cases. High masker levels yielded reduced probe thresholds for two Advanced Bionics subjects. This was partly eliminated with a longer temporal offset between each masker-probe pulse pair, as was used with Cochlear subjects. These findings support the use of temporal gaps between stimulation of subsequent electrodes to reduce channel interactions.

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I. INTRODUCTION

Channel interaction in cochlear implants (CIs) results in temporal and spectral degradations in how the auditory periphery codes sound with the device. There are two types of channel interaction that can occur at the auditory periphery with electrical stimulation. The first is electrical field interaction, which occurs with simultaneous stimulation of two or more electrodes. Depending on the relative phase of each current pulse (or analog signal), electrical fields either sum or cancel prior to neural excitation (e.g., Shannon, 1983; Abbas and Brown, 1988; Boëx *et al.*, 2003a; Stickney *et al.*, 2006; Hughes and Stille, 2009). The second type is interaction at the level of the neural membrane, which can occur with either simultaneous or sequential stimulation. Depending on the relative stimulus levels, timing, and spatial separation between electrodes, either masking or facilitation can occur (e.g., Shannon, 1983; Cartee *et al.*, 2000; de Balthasar *et al.*, 2003; Hughes and Stille, 2008; Kwon and van den Honert, 2009). Interleaved pulses are used in CI speech processing because sequential stimulation results in less channel interaction than simultaneous stimulation (Middlebrooks, 2004; Boëx *et al.*, 2003a; de Balthasar *et al.*, 2003). Interestingly, non-simultaneous channel interaction has been studied more extensively using forward masking (e.g., Lim *et al.*, 1989; Chatterjee and Shannon, 1998;

Chatterjee, 1999; Boëx *et al.*, 2003b; Cohen *et al.*, 2003; Chatterjee *et al.*, 2006; Kwon and van den Honert, 2006b; Nelson and Donaldson, 2002; Hughes and Stille, 2008) than with the interleaved stimuli that are used clinically (Tong and Clark, 1986; McKay *et al.*, 2001; de Balthasar *et al.*, 2003; Zeng *et al.*, 2005; Kwon and van den Honert, 2009). The effect of channel interaction with the latter paradigm was the focus of the present study.

To understand the peripheral mechanisms involved with channel interaction resulting from sequential stimulation, it is important to review the physiological response patterns when pulsatile stimuli are delivered to auditory neurons. *Masking* refers to the state when neurons depolarize in response to a stimulus, enter a refractory period, and then are either unable to respond or have reduced excitability to a succeeding stimulus (e.g., Brown *et al.*, 1990; Cartee *et al.*, 2000). Masking is more likely to occur with closely-spaced electrodes due to current spread, although physiological and psychophysical studies have shown that substantial spatial overlap of excitation patterns can occur for even widely spaced electrodes with monopolar stimulation (e.g., Cohen *et al.*, 2003; Hughes and Stille, 2008; Zhu *et al.*, 2012). *Facilitation* describes the phenomenon in which a sub-threshold pulse (often termed a “conditioner” or “perturbation signal”) partially depolarizes neural tissue so that less current is needed for a successive pulse (probe) to generate an action potential (Cartee *et al.*, 2000). In this case, the initial low-level pulse is not sufficient to generate an action potential in isolation. The detection threshold for the probe is therefore lower in the presence of

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the sub-threshold pulse compared with that of the probe alone. Facilitation requires a shorter time period between successive pulses ($\sim 300\text{--}400\ \mu\text{s}$ or less) than for masking effects (Cartee *et al.*, 2000; Morsnowski *et al.*, 2006). Because current strength decays with distance, current amplitudes can reach threshold or sub-threshold levels at the spatial edges of the current field, potentially resulting in facilitation effects between more widely spaced electrodes (Tong and Clark, 1986). The functional consequence of this could include unwanted intermediate pitch percepts.

Channel interactions with sequential stimulation will depend on the electrode-nerve interface (e.g., electrode proximity to neurons, neural survival, tissue impedance), distance between stimulated electrodes, and stimulation parameters such as level, timing, and rate (Throckmorton and Collins, 1999; Middlebrooks, 2004; Kwon and van den Honert, 2009). The present study focuses specifically on how level and rate affect temporal-spatial channel interaction for interleaved stimuli. Only a few studies have examined level effects for interleaved pulse trains on adjacent or nearby electrodes (de Balthasar *et al.*, 2003; Zeng *et al.*, 2005; Kwon and van den Honert, 2009). These studies each involved four subjects with early-generation devices that utilized either bipolar (Zeng *et al.*, 2005) or monopolar (de Balthasar *et al.*, 2003; Kwon and van den Honert, 2009) stimulation modes with varied timing between the offset and onset of the interleaved pulses. In general, larger threshold shifts (i.e., greater amounts of masking) were observed for higher masker levels. For sub-threshold stimulation on an adjacent electrode, Zeng *et al.* (2005) found facilitation effects for their Cochlear recipients, whereas de Balthasar *et al.* (2003) found minor masking effects for their Ineraid recipients. Given the small number of subjects in each study, it is unclear whether these conflicting findings were due to differences in stimulation mode (bipolar versus monopolar, respectively), device type (Cochlear versus Ineraid, respectively), temporal offset between the interleaved pulses (offset by one-half the stimulus period versus 0- μs offset to onset, respectively), or simply inter-subject differences. It was therefore of interest to examine level effects for interleaved stimuli in a larger group of subjects using temporal offsets that are similar to those used in clinical stimulation paradigms.

Kwon and van den Honert (2009) was the only study to examine growth of masking across several different rates [125, 500, 2000, and 6410 pulses per second per channel (pps/ch)] using interleaved pulse trains for various masker-probe electrode separations. For the “on-frequency” condition (masker and probe both delivered to electrode 10), growth of masking was modeled separately for each rate across the group data. As the stimulus rate increased from 125 to 2000 pps/ch, the masker level had to be at a higher percentage of the dynamic range before masking was observed. They found that the slope of the growth-of-masking function increased from 125 to 500 pps/ch, and then decreased from 500 to 6410 pps/ch, with the shallowest function occurring for the fastest rate (6410 pps/ch). In general, the masker was less effective as the stimulation rate increased. Although all of the individual data are plotted in

their Fig. 5, it is difficult to discern whether the same trends observed for the on-frequency condition hold for the off-frequency condition across the group. The latter condition is a topic of the present study and is of interest because it is more clinically relevant.

A limitation of several of the studies discussed thus far is how the interleaved stimuli were delivered. Specifically, the masker was longer in duration than the probe, and the probe was typically presented in the steady-state portion of the masker (Tong and Clark, 1986; de Balthasar *et al.*, 2003; Kwon and van den Honert, 2009). This is particularly important because signal detection thresholds can be altered depending on whether the probe is presented in the temporal center or near the onset/offset edge of the masker (Elliot, 1965; Zwicker, 1965; Zeng *et al.*, 2005); this can be further complicated by the level of the masker. Zeng *et al.* (2005) showed that for low-level maskers (sub-threshold and threshold), facilitation was more likely to occur when the probe was presented at the onset of the masker versus in the steady-state portion. However, when maskers were at a comfortable level, larger threshold shifts (more masking) were observed for probes presented at the onset than the steady-state portion of the masker. To our knowledge, no study has investigated the perceptual effect of channel interaction for equal-duration interleaved masker and probe pulses presented to adjacent electrodes, which might represent a more clinically relevant scenario. For example, the second and third formants in high vowels (/i/, /I/) from a female talker have similar durations. The respective frequencies for these formants will likely fall into adjacent filter bands, potentially producing channel interactions. Or, in a broader sense, similar interactions may occur when listening to target speech in a background of competing speech.

The purpose of the present study was to measure masking effects for two low-level maskers (sub-threshold and threshold) and two high-level maskers (comfortable and loud-but-comfortable) for two stimulus rates (1700 and 3400 pps/ch) using interleaved pulse trains of equal duration in a psychophysical three-interval forced choice (3IFC) task to assess the basic perceptual effects of channel interaction. The masker and probe were delivered via monopolar mode to adjacent electrodes in the middle of the array using a short temporal offset between masker-probe pulse pairs, as would occur with clinical stimulation. The four masker levels were chosen because they represent the lower and upper stimulus-level boundaries used to program clinical processors (“maps”). The lower program levels for Cochlear devices are called T-levels, which are set to behavioral threshold. For Advanced Bionics and MED-EL devices, T-levels are often not measured.¹ Rather, T-levels are set to a certain percentage of the upper program level (e.g., 10%), and are typically not audible when stimulated in isolation (Holden *et al.*, 2011). It is therefore important to determine whether this sub-threshold stimulation can produce perceptual facilitative effects, especially for monopolar stimulation of adjacent electrodes. Upper program levels are set to a loud-but-comfortable level (Cochlear or MED-EL) or a most-comfortable level (Advanced Bionics), so it was also important to determine masking effects at these two upper-comfort levels. It

was hypothesized that probe thresholds would be reduced (facilitative effects) for sub-threshold maskers, and possibly for threshold-level maskers due to charge integration on the neural membrane secondary to spatial overlap of excitation between adjacent electrodes. For supra-threshold maskers, it was hypothesized that probe thresholds would increase as the masker level increased.

Regarding stimulation rate, the two rates were chosen because (1) they represent a range that is commonly used in present-day clinical processing strategies² and (2) neural firing patterns generally differ between these two rates. The first rate, 1700 pps/ch, is in a range that has been shown to yield a synchronized alternating-amplitude pattern of neural responses (Hughes *et al.*, 2012; Hughes *et al.*, 2014). The second rate, 3400 pps/ch, was twice that amount and is in a range that has been shown to induce a relatively stochastic state among auditory nerve fibers (Hughes *et al.*, 2012; Hughes *et al.*, 2014). The goal was to determine whether rates that produce different firing patterns at the periphery will produce different patterns of perceptual masking or interaction. The physiological data suggest that

as the stimulation rate increases, fewer fibers are available to respond to each pulse in the masker, which could make the masker less effective than for a slower rate. In addition, neurons that overlap spatially between the masker and probe electrodes could be in a relative refractory state from prior stimulation by the masker and then discharge in response to the probe, making the probe easier to detect. It was therefore hypothesized that 3400 pps/ch would produce smaller probe threshold shifts (less masking) compared with 1700 pps/ch, consistent with the shallower growth of masking observed by Kwon and van den Honert (2009) for faster rates. In a traditional forward-masking paradigm, the probe rate would typically be fixed while the masker rate is varied in order to directly compare the effects of masker rate. However, for most strategies used clinically, the per-channel rates are the same across all electrodes.³ The present study therefore sought to examine perceptual effects with clinically relevant stimuli. Results from this study will provide insight into whether psychophysical threshold shifts reflect masking or potentially facilitative effects using clinical stimuli.

TABLE I. Demographic information for participating subjects. Speech perception scores were obtained with recorded stimuli at 60 dB SPL in the CI-only condition using the AzBio sentence test in quiet [four subjects were tested with the hearing in noise test (HINT), as indicated below]. SNHL, sensorineural hearing loss; +p, electrode array with positioner; ACE, Advanced Combination Encoder; CIS, continuous interleaved sampling; F120, Fidelity 120; pps, pulses per second; M, male; F, female. Age at CI, duration of CI use, and duration of deafness are reported as years, months.

Subject ID	Device	Gender	Ear	Age at CI	Duration CI use	Duration deafness	Etiology & Onset	Daily-use strategy, per-channel rate (pps)	Speech perception (% correct)
R3	24R(CS)	F	R ^a	56, 2	10, 11	6, 0	Unknown, progressive	ACE, 900	60%
R6	24R(CS)	F	R	52, 5	8, 9	6, 0	Autoimmune disease, unknown	ACE, 1200	84%
R7	24R(CS)	M	R	62, 2	7, 11	5, 0	Unknown, progressive	ACE, 900	89%
R10	24R(CS)	M	R	61, 10	9, 9	2, 0	Unknown, progressive	ACE, 1200	72%
F1	24RE(CA)	F	L ^a	60, 7	6, 5	11, 0	Unknown, progressive	ACE, 900	66%
F10 ^g	24RE(CA)	F	R ^b	8, 3	6, 10	8, 3	Waardenburg, congenital	ACE, 900	59%
F11 ^g	24RE(CA)	F	L ^b	1, 10	13, 3	1, 10	Waardenburg, congenital	ACE, 900	67%
F13	24RE(CA)	F	L	35, 2	4, 5	8, 0	Unknown, progressive	ACE, 1800	84%
F17	24RE(CA)	M	R ^c	42, 11	6, 8	11, 0	Congenital, progressive	ACE, 900	97%
F18	24RE(CA)	M	L ^c	45, 6	4, 1	3, 0	Congenital, progressive	ACE, 500	66%
FS19	CI422	M	R ^d	68, 11	0, 3	7, 0	Unknown, progressive	ACE, 900	76%
N1 ^g	CI512	M	L	67, 5	3, 10	9, 0	Familial/Noise, progressive	ACE, 500	95%
N5	CI512	F	R	50, 8	1, 11	1, 0	Unknown, sudden SNHL	ACE, 900	84%
N6 ^g	CI512	M	R	82, 0	3, 4	4, 0	Unknown, progressive	ACE, 500	90%
N11	CI512	M	L ^d	67, 5	1, 3	6, 0	Unknown, progressive	ACE, 900	85%
C8	Clarion CII	M	L	55, 7	9, 8	3, 0	Unknown, progressive	HiRes-P w/F120, 3712	83%
C18	90K HF 1J	F	R	36, 8	6, 5	34, 0	Unknown, sudden SNHL	HiRes-S, 1513	86% (HINT)
C24	CII HF 1+p	F	R	67, 4	11, 7	15, 0	Unknown, progressive	HiRes-S w/F120, 3712	60%
C26	90K HF 1J	F	R	67, 6	5, 1	14, 0	Unknown, progressive	HiRes-S w/F120, 2750	72%
C29	90K HF 1J	F	R ^e	30, 11	5, 6	20, 11	Meningitis, sudden	HiRes-S w/F120, 2970	96%
C33	90K HF 1J	M	R	36, 0	4, 6	34, 0	Genetic, congenital	HiRes-S w/F120, 3712	28%
C36	90K HF 1J	M	R	38, 4	3, 11	38, 0	Unknown, congenital	HiRes-S w/F120, 3093	23% (HINT)
C39	90K HF 1J	F	L	63, 0	4, 2	0, 6	Unknown, sudden SNHL	HiRes-S w/F120, 3712	81%
C40	CII HF 1+p	M	L ^f	59, 5	11, 11	21, 0	Unknown/Familial, progressive	HiRes-P w/F120, 2750	70%
C41	90K HF 1J	M	L	51, 6	0, 6	36, 0	Genetic, progressive	HiRes-P w/F120, 3712	56% (HINT)
C42	90K HF 1J	F	R	70, 11	0, 4	2, 0	Unknown/Familial, progressive	HiRes-S, 3256	81%
C43	Clarion CII	F	R	1, 6	10, 1	1, 6	Genetic, congenital	CIS, 884	71% (HINT)
C44	90K HF 1J	M	R ^f	69, 9	1, 9	8, 0	Unknown/Familial, progressive	HiRes-S w/F120, 3712	69%
C45	90K HF 1J	F	L ^c	35, 1	1, 9	25, 0	Meningitis, sudden	HiRes-S w/F120, 3712	48% (HINT)

^{a-f}Indicate right and left ears for bilateral subjects.

^gSubject was explanted and re-implanted.

II. METHODS

A. Subjects

Data were collected for 29 ears of 23 subjects. Demographic information for each subject is displayed in Table I. Subject ID is listed in the far left column of the table; each ID contains a letter that identifies the internal device of each subject (see column 2), and the number following the letter was assigned in a successive manner upon enrollment in the laboratory. Six bilaterally implanted subjects participated with both ears, designated in Table I with paired symbols (R3/F1, F10/F11, F17/F18, FS19/N11, C29/C45, C40/C44). Subjects included 15 ears with Cochlear Corporation devices (Cochlear, Ltd., Macquarie, New South Wales, Australia): N=4 Nucleus 24R(CS), N=6 Nucleus 24RE(CA), N=1 CI422, and N=4 CI512; and 14 ears with Advanced Bionics devices (Advanced Bionics Corp., Sylmar, CA): N=4 Clarion CII HiFocus (two subjects had electrode-array positioners; see Table I) and N=10 HiRes 90K 1J. All participants were at least three months post activation of his/her CI before enrolling in the study. Except for subject C42, each participant had a full insertion of the electrode array as confirmed by the operative report. For C42, electrodes 15 and 16 were reported to be outside of the cochlea. This study was approved by the Institutional Review Board at Boys Town National Research Hospital under protocol 03-07-XP.

B. Equipment set-up

Psychophysical thresholds for Cochlear subjects were measured using a custom program that implemented Nucleus Implant Communicator (NIC) subroutines (Cochlear Ltd., Macquarie, New South Wales, Australia). A laboratory programming Pod and Freedom processor were used to deliver the stimuli through direct connect to the implant. Custom Sound EP v.3.2 (Cochlear Ltd., Macquarie, New South Wales, Australia) was used to evaluate electrode impedances prior to testing to ensure values were within normal limits for the test electrodes. For Advanced Bionics subjects, psychophysical thresholds were obtained using the experimental research platform, Bionic Ear Data Collection System (BEDCS) (Advanced Bionics Corp., Valencia, CA). A Platinum Series Processor and a programming interface (CPI II) delivered the stimuli via direct connect. Electrode impedances were measured using SoundWave 2.2 (Advanced Bionics, Valencia, CA). All subjects, except for N6 and C42, had impedance values within normal limits as specified by the manufacturer for all electrodes. Previously identified open circuits were measured for electrode 1 for N6, and electrodes 14, 15, and 16 for C42; none of these electrodes were used as test electrodes in the study.

C. Stimuli

Stimuli consisted of 300-ms cathodic-leading, symmetric, biphasic, periodic pulse trains, as illustrated in Fig. 1(A). Pulse-train rates were 1700 (588- μ s period) and 3400 pps/ch (294- μ s period); the same per-channel rate was used for both the masker and probe. In the present experiment, the leading pulse train is termed the “masker,” while the second

(interleaved) pulse train is the probe. Although the stimulus level of the leading pulse train will dictate whether masking or facilitative effects occur, the term “masker” is used throughout this report for simplicity.

For Cochlear subjects, stimuli were 25 μ s/phase with an 8- μ s interphase gap (IPG), and an 8- μ s masker-probe offset, presented in monopolar mode [referenced to MP1; Fig. 1(B)]. An IPG is used to separate the phases to make the

FIG. 1. Stimulus configurations for Cochlear and Advanced Bionics subjects. (A) Masker and probe stimuli each consisted of 300-ms, periodic, cathodic-leading, biphasic pulse trains that were interleaved with the masker as the leading pulse; masker and probe were presented to adjacent electrodes. (B) Close-up schematic of two masker-probe pulse pairs for Cochlear subjects. Pulses were 25 μ s/phase with an 8- μ s interphase gap and an 8- μ s delay between masker offset and probe onset. Stimulus period is indicated with arrows toward the top of the panel. Time between offset of each probe pulse and onset of the subsequent masker pulse is indicated toward the center of the panel (periods for 1700 pps/ch and 3400 pps/ch are above and below the arrow, respectively). (C) Stimuli for Advanced Bionics subjects, plotted as in panel (B). Pulses were 32 μ s/phase with a 0- μ s interphase gap and 0- μ s masker-probe offset. (D) Stimuli for the subset of seven Advanced Bionics subjects who completed an “Offset” experiment. Pulses were 32 μ s/phase with a 0- μ s interphase gap and an 8.98- μ s masker-probe offset.

stimulus more efficient (e.g., Shepherd and Javel, 1999). That is, the IPG allows more time for neurons to discharge in response to the first phase before the charge is removed by the balancing phase; the result is that less current is required to achieve threshold. A characteristic of the Cochlear system is the presence of timing gaps within and between the biphasic pulses. The time between the opposing phases of a biphasic pulse is the IPG; 8 μ s was utilized in this study because Cochlear's present-generation clinical strategy, Advanced Combination Encoder (ACE), has a fixed 8- μ s IPG. Second, there is a gap between the offset of the first biphasic pulse (masker) and the onset of the next biphasic pulse (probe), labeled offset in Fig. 1(B). During this period, the electrodes are shorted to each other to dissipate any residual charge in the neural tissue. The CI24R internal is limited to an 8- μ s gap; therefore, this was the masker-probe offset used across all Cochlear participants in the present study.

For Advanced Bionics subjects, stimuli were 32- μ s/phase, with a 0- μ s interphase gap, and a 0- μ s masker-probe offset [Fig. 1(C)], presented in monopolar mode. The Advanced Bionics device does not typically have an IPG between the two phases of a pulse, and the temporal offset is eliminated between the masker and probe if the fastest rate is used in its clinical strategies. This allows the Advanced Bionics device to stimulate at high rates. Preliminary data for this study, however, warranted investigation of a condition in which there was a temporal offset between the masker and probe pulses similar to that used with Cochlear devices (8 μ s). Therefore, a subset of seven Advanced Bionics subjects (C8, C29, C36, C39, C41, C42, and C43) completed a second condition, called the "Offset" experiment, in which there was an 8.98- μ s temporal offset between the masker and probe pulses [Fig. 1(D)]. Due to timing restrictions in BEDCS, 8.98 μ s was the closest value to 8 μ s the software would allow.

D. Procedures

Pulse trains were presented to adjacent mid-array electrodes for all subjects. Middle electrodes were chosen to avoid the effects of potentially poorer nerve survival for basal electrodes (Nadol, 1997; Pfingst et al., 2011) and variations in insertion depth for apical electrodes across subjects. For Cochlear subjects, electrodes 11 and 12 served as the probe and masker, respectively. For Advanced Bionics subjects, electrodes 9 and 8 served as the probe and masker, respectively. In both cases, the masker was delivered to the more apical electrode in the pair. Note that the basal-to-apical electrode numbering is opposite between the two manufacturers.

For the experimental procedure, subjects sat in an acoustically controlled sound booth in front of a touchscreen monitor connected to the investigator's computer. Loudness estimates were performed first to establish the subject's behavioral dynamic range for the masker and probe electrodes separately, for the two different rates of stimulation. Pulse trains of 300 ms were presented in an ascending manner in five-current-level (CL) step sizes for Cochlear

subjects. For Advanced Bionics subjects, stimulus levels were increased using log-based increments using the following formula:

$$b = a10^{n/20},$$

where a was the starting current level, n was the step size factor (set to 3 in this study), and b was the next current level (substituted for a in the next iteration). Subjects indicated loudness estimates using a visual numeric rating scale with the following definitions: 1 = just noticeable, 5 = comfortable, 7 = loud but comfortable, and 9 = maximum loudness level.

Next, quiet thresholds were obtained for the probe signal alone and for the masker alone for the two different rates. A 3IFC procedure was used. One interval randomly contained the stimulus, and the other two intervals were silent. Each interval was 400 ms in duration with the 300-ms stimulus temporally centered. The three intervals were each separated by a 500-ms inter-stimulus interval. The starting presentation level was set individually between the subject's "1" and "5" rating from the loudness estimates to ensure the stimulus was audible. Thresholds were determined using a three-down, one-up adaptive procedure that estimated 79.4% correct (Levitt, 1971). For Cochlear subjects, the adaptive procedure used a 10-CL step size for the first reversal, 5-CL steps for the second reversal, and 2-CL steps for the last six reversals. For Advanced Bionics subjects, the step size for the first reversal was 3 dB, followed by 1.5 dB for the second reversal, and 0.75 dB steps for the last six reversals. The final six reversals were averaged to obtain the mean threshold of each adaptive track. The final threshold value for each condition was the average of three adaptive tracks. The subject was instructed to choose the interval that contained the stimulus. Correct-answer feedback was provided.

Finally, psychophysical thresholds were measured for the probe in the presence of the interleaved masker to determine the effect of masker level on probe threshold for the two different rates. Probe thresholds were measured in the following four masker conditions: (1) masker at *sub threshold*, which corresponded to the current level set at 0.88 dB (Advanced Bionics subjects) or 5 CL (Cochlear) below the masker-alone (quiet) threshold from the first adaptive procedure described above; (2) masker at *threshold*, using the threshold value from the masker-alone condition from the adaptive task described above; (3) masker at the *comfortable* level, corresponding to the subject's "5" rating from the loudness estimates; and (4) masker at the *loud but comfortable* level, corresponding to the subject's "7" rating. The same adaptive procedure used for the masker and probe quiet conditions was used to obtain masked probe thresholds. For this, the masker pulse train was presented in all three intervals while the probe pulse train was interleaved with the masker in only one interval [see Fig. 1(A)]. Within an adaptive track, the masker level was held constant at one of the four levels indicated above, while the probe level was adapted to threshold. Subjects were instructed to choose the interval that sounded different. Subjects were provided with feedback regarding the interval that contained the probe.

Test order was randomized for rate and level within and across subjects.

E. Analysis

Statistical analyses were run using Statistical Analysis Software (SAS) (Cary, NC). For each subject group, a two-way repeated-measures analysis of variance was used to assess the effects of stimulus *level* [probe quiet (Q), masker at sub-threshold (ST), masker at threshold (MT), masker at “comfortable” (MC), and masker at “loud but comfortable” (ML)] and *rate* (1700 and 3400 pps/ch) on probe threshold (in clinical units) and probe threshold shift (dB). The simulation technique (Westfall *et al.*, 2011) was used to adjust for multiple comparisons because it is the recommended method for repeated measures and is less conservative than the Bonferroni adjustment.

III. RESULTS

The objective of this study was to measure masking and potential facilitative effects on probe thresholds for interleaved pulse trains. First, the raw data (probe thresholds) are presented in each device’s respective clinical units (CL for Cochlear and μA for Advanced Bionics) in Figs. 2, 3, and 4. Next, the data are presented as the amount of probe threshold shift in dB (re: 1 μA) in Figs. 5, 6, and 7 to allow for better assessment of potential facilitation effects and for comparisons to the literature.

The individual and group raw data for the 15 Cochlear subjects are displayed in Fig. 2. Filled circles and open squares represent probe thresholds for 1700 pps/ch and 3400 pps/ch, respectively. Probe thresholds are plotted as a function of masker level for the following categories: masker at sub-threshold (ST), probe in quiet (no masker present; Q), masker at threshold (MT), masker at “comfortable” (MC), and masker at “loud but comfortable” (ML). Note that because categorical loudness levels were used (so as to relate to clinical programming levels), it is not appropriate to consider the x axis as a purely linear scale. It is also for this reason that the slope of the growth of masking was not determined. For subject F18, loudness levels for ML could not be obtained for 1700 or 3400 pps/ch due to voltage compliance limits, so this subject was excluded from the group analysis. In general, the individual graphs show similar trends among the subjects with lower probe thresholds for 3400 pps/ch than for 1700 pps/ch, and higher probe thresholds for the higher masker levels. There were three subjects (R10, FS19, and N11) who had flatter functions than the other subjects, suggesting limited masking effects even at high levels. The mean data for the group (excluding F18) are shown in a separate plot on the bottom right corner of Fig. 2.

Results for the group data indicated a significant effect of level ($F_{(4, 10.3)} = 36.48, p < 0.001$) and rate ($F_{(1, 14)} = 198.06, p < 0.001$) on mean probe thresholds, with no significant interaction between factors ($F_{(4, 10.9)} = 2.49, p = 0.11$). *Post hoc* results for level showed no significant difference between ST and Q ($p = 0.18$) or between ST and MT ($p = 0.97$). All other level comparisons were significant ($p \leq 0.01$). The mean thresholds for each rate are detailed in Table II. As expected,

post hoc testing showed mean thresholds were significantly lower for 3400 pps/ch (mean = 116.7 CL) than for 1700 pps/ch (mean = 134.5 CL; $p < 0.001$), likely due to temporal integration.

Figure 3 displays the individual and group data (in μA) for the 14 Advanced Bionics subjects. The data are plotted as in Fig. 2. The Advanced Bionics participants tended to show greater variability in terms of level and rate effects than those observed for the Cochlear participants. Several subjects exhibited relatively flat functions and overlap between thresholds for the two rates (C24, C33, C39, and C40), or functions that crossed (C8, C29, C39, C41, and C44). Several subjects exhibited an unexpected “rollover” effect, where masking became less effective at higher levels (C8, 1700 pps/ch; C29, both rates; C39, 1700 pps/ch; C43, 1700 pps/ch; and C44, 1700 pps/ch).

The group data showed a significant main effect of level ($F_{(4, 44)} = 12.79, p < 0.001$) and rate on mean probe threshold ($F_{(1, 16.7)} = 11.19, p = 0.004$), with a significant interaction effect ($F_{(4, 61.9)} = 3.04, p = 0.02$). Mean thresholds are shown in Table II. Although there was a significant interaction between rate and level, *post hoc* results showed similar level comparisons for both rates. Specifically, all level comparisons were significant ($p < 0.05$) for both rates, except for ST versus MT and MC versus ML ($p > 0.05$). In general, probe thresholds were significantly lower for 3400 pps/ch than for 1700 pps/ch (189.2 μA versus 226.6 μA , respectively), consistent with the results for Cochlear subjects and expected temporal integration effects.

One of the primary differences between the two groups was that rollover effects at higher masker levels were observed for several Advanced Bionics subjects, but not in any of the Cochlear subjects. To investigate whether this difference was caused by the differences in default masker-probe offset intervals between devices, seven Advanced Bionics subjects participated in a second experiment in which an 8.98- μs temporal offset was added between each masker-probe pair to approximate the condition used with Cochlear subjects [see Fig. 1(D)]. Figure 4 displays the data for these seven subjects; results are plotted as in Figs. 2 and 3. The longer temporal offset resulted in probe thresholds for C8 that were similar to what was hypothesized (higher thresholds with increasing masker level; compared with this subject’s results in Fig. 3). For C29, the rollover was still present for 1700 pps/ch. The rollover effect was reduced for 3400 pps/ch compared to that observed in Fig. 3; however, the results were not consistent with the expected hypothesis. The longer temporal offset did not alter the results substantially for three subjects (C36, C41, and C42) when compared to the 0- μs data in Fig. 3; however, the thresholds were altered for C43 and C39.

The analyses revealed no significant effect of level ($F_{(4, 4.95)} = 2.88, p = 0.14$) or rate ($F_{(1, 5.01)} = 1.77, p = 0.24$), and no significant interaction between factors ($F_{(4, 5.97)} = 1.23, p = 0.39$) for the 8.98- μs offset data. Mean thresholds are listed in Table II. In summary, for subject C8, a longer temporal offset between the masker and probe pairs resulted in probe thresholds that were consistent with the hypothesis for level (higher thresholds for higher masker levels). For

subject C29, the longer offset reduced the rollover effect for 3400 pps/ch, but not for 1700 pps/ch.

To better evaluate masking and potential facilitation effects, the data from Figs. 2–4 were replotted as probe threshold shift in dB (re: 1 μ A). First, thresholds in CL (Cochlear) and μ A (Advanced Bionics) for each trial were converted to dB and then averaged to obtain the mean threshold in dB for each condition. Next, the mean unmasked (quiet) threshold was subtracted from each mean masked threshold to yield the probe threshold shift in dB. Figure 5 displays the individual and group data for the 15 Cochlear participants (compare with Fig. 2). Because probe

threshold shift for each individual masker condition represents a difference measure between two mean values, it is not possible to plot standard deviation bars. The horizontal dashed line represents the probe quiet threshold (no threshold shift); thresholds above and below the line indicate masking and facilitation effects, respectively. There were cases in which the ST- or MT-level maskers resulted in threshold shifts slightly below zero (e.g., R7, R10, and FS19). The largest negative shift was -1.31 dB for R7 at ST for 1700 pps/ch. To estimate whether these instances were truly facilitation, the standard deviation of the three adaptive tracks (in dB) were calculated for the probe-alone and for

FIG. 2. Individual and group raw results for the 15 Cochlear participants. Subject number is shown on each plot. The group mean is plotted in the bottom right corner. Bars indicate ± 1 standard deviation. Data for F18 could not be obtained for the ML condition due to voltage compliance limits. Probe thresholds (CL) are plotted as a function of masker current level (re: loudness rating). Current levels for ST refer to probe threshold with the masker at sub-threshold (masker set 5 CL below masker threshold), Q refers to the probe threshold in quiet, MT is masker at threshold, MC is masker at a comfortable level, and ML is masker at a loud-but-comfortable level. Filled circles indicate probe thresholds for 1700 pps/ch; open squares are for 3400 pps/ch.

FIG. 3. Individual and group probe thresholds (± 1 standard deviation) for the 14 Advanced Bionics subjects. Thresholds (μA) are plotted as in Fig. 2 [the sub-threshold (ST) masker is 0.88 dB below masker threshold].

the masked conditions for these subjects. In most cases, the standard deviation across trials was generally 1–2 dB. Therefore, the instances in which difference values were below zero appear to be within the measurement variance across individual trials, and likely not true facilitation.

The results revealed a significant main effect of level ($F_{(3, 51.1)} = 56.03$, $p < 0.001$), but no significant effect of rate ($F_{(1, 21.5)} = 1.87$, $p = 0.19$), and no significant interaction

($F_{(3, 25.9)} = 0.35$, $p = 0.79$). The mean threshold shifts are detailed in Table II. For the main effect of level, *post hoc* testing indicated no significant differences between ST and MT ($p = 1.0$) or MC and ML ($p = 0.31$); all other level comparisons were significant ($p < 0.001$). The level effects generally supported the hypothesis that probe threshold shifts increase with masker level. The hypothesis that 3400 pps/ch would produce smaller probe threshold shifts than 1700 pps/

FIG. 4. Individual and group probe thresholds (± 1 standard deviation) for the seven Advanced Bionics subjects who completed the “Offset” experiment, consisting of an 8.98- μs delay between masker and probe pulses. Probe thresholds (μA) are plotted as in Figs. 2 and 3.

FIG. 5. Probe threshold shift (in dB) plotted as a function of masker level for the 15 Cochlear participants. Threshold shift was calculated by subtracting the unmasked threshold in dB from the masked threshold in dB. Because this was a measure of threshold shift, probe quiet thresholds (Q) are not plotted. Filled circles indicate probe thresholds for 1700 pps/ch; open squares are for 3400 pps/ch. Subject F18 was excluded from the group graph because of voltage compliance limits for the highest masker level (ML). Data points above the dashed line at zero indicate masking effects, and values below the line indicate facilitative effects.

ch was not supported, as both rates exhibited similar threshold shifts at each masker level.

Probe threshold shift in dB for the Advanced Bionics participants are displayed in Figs. 6 and 7 for the 0- μ s and 8.98- μ s temporal offsets, respectively. For the 0- μ s data (Fig. 6), no subjects exhibited facilitation for low-level maskers. For two subjects (C8 and C29), masked probe thresholds for the high-level maskers were lower than the unmasked probe thresholds. There were two additional subjects (C39 and C44) who exhibited a rollover effect for 1700 pps/ch at MC and/or ML that approached zero (probe quiet threshold). For the group data in Fig. 6, there was a significant effect of level ($F_{(3,39.3)}=9.12$, $p<0.001$), rate ($F_{(1,16.7)}=9.44$, $p=0.007$), and a significant interaction ($F_{(3,49.3)}=5.42$, $p=0.003$). For 1700 pps/ch, none of the masker-level comparisons were significant ($p\leq 1.09$) due to the large standard deviations at high levels. For 3400 pps/ch, all level comparisons were significant ($p\leq 0.002$) except for ST versus MT ($p=0.95$) and MC versus ML ($p=0.28$). Mean threshold shifts are shown in Table II. The mean data were generally consistent with the hypothesis that greater threshold shifts would be observed for higher masker levels, although several subjects exhibited less effective masking with increased level. The hypothesis that 3400 pps/ch would produce smaller probe threshold shifts than 1700 pps/ch was not supported; in fact, the opposite trend was exhibited.

For the 8.98- μ s offset data (Fig. 7), two subjects exhibited facilitation (for low-level maskers) or rollover effects

(for high-level maskers); subject C43 had facilitation for low-level maskers (ST and MT), and C29 exhibited rollover effects for high-level maskers (MC and ML). The analyses revealed a significant effect of level ($F_{(3,20.5)}=14.39$, $p<0.0001$), rate ($F_{(1,7.52)}=7.77$, $p=0.03$), and a significant interaction effect ($F_{(3,22.4)}=4.74$, $p=0.01$) on mean threshold shift. Mean threshold shifts are shown in the bottom two rows of Table II. *Post hoc* testing for 1700 pps/ch indicated a significant difference in threshold shift between ST and ML ($p<0.001$) and MT and ML ($p=0.01$); no other level comparisons were significant. For 3400 pps/ch, there were significant differences between ST and MC ($p=0.02$), ST and ML ($p<0.001$), MT and MC ($p=0.04$), and MT and ML ($p<0.001$); all other level comparisons were not significant ($p\geq 0.98$). Again, the level effects generally supported the hypothesis that greater threshold shifts would be observed for higher masker levels, and the hypothesis that 3400 pps/ch would produce smaller probe threshold shifts than 1700 pps/ch was not supported.

IV. DISCUSSION

The objective of this study was to evaluate masking effects and potential facilitation for interleaved pulse trains using present-generation CI systems. Psychophysical thresholds were evaluated using two stimulus rates (1700 and 3400 pps/ch) across four masker levels ranging from sub-threshold to “loud but comfortable.” It was hypothesized that probe

FIG. 6. Probe threshold shift (in dB) plotted for the 0- μ s masker-probe offset condition for the 14 Advanced Bionics participants. Data are graphed as in Fig. 5.

thresholds would increase with masker level and that sub-threshold maskers (and potentially threshold-level maskers) would produce facilitation. Further, it was expected that the faster rate would result in less effective masking than the slower rate due to neural desynchronization. In general, larger threshold shifts were observed with increased masker level, consistent with the hypothesis; however, there was little

evidence of facilitation for threshold or sub-threshold maskers. The hypothesis regarding rate differences was not supported. An unexpected finding was that probe levels were reduced below the probe quiet thresholds at the *highest* masker levels for two Advanced Bionics subjects. To the authors' knowledge, similar findings have only been reported in one published abstract (Donaldson *et al.*, 2005).

FIG. 7. Probe threshold shift (in dB) plotted for the 8.98- μ s masker-probe offset condition for seven Advanced Bionics participants. Data are graphed as in Figs. 5 and 6.

TABLE II. Group mean thresholds (top section) and threshold shifts (bottom section) for all masker levels and rates. Threshold units are listed under the Device column. CL, current level units; μA , microamps; dB, decibels re: 1 μA . AB, Advanced Bionics; ST, sub-threshold masker; Q, probe quiet thresholds (no masker); MT, masker at threshold; MC, masker at comfortable level; ML, masker at loud but comfortable level.

Device (units)	Rate (pps/ch)	ST	Q	MT	MC	ML
Thresholds						
Cochlear (CL); Fig. 2	1700	114.2	110.3	116.8	158.4	172.7
	3400	97.3	91.3	96.2	141.0	157.9
AB (μA); Fig. 3	1700	190.7	155.4	190.9	289.1	307.0
	3400	140.9	113.0	152.7	240.2	299.3
AB 8.98 offset (μA); Fig. 4	1700	210.6	220.3	230.6	295.1	346.1
	3400	163.5	149.8	171.7	258.5	327.4
Threshold shifts						
Cochlear (dB); Fig. 5	1700	0.61	—	0.96	7.76	9.34
	3400	1.40	—	1.23	8.44	9.88
AB (dB); Fig. 6	1700	1.98	—	2.06	5.39	5.12
	3400	2.14	—	2.72	6.86	8.80
AB 8.98 offset (dB); Fig. 7	1700	−0.03	—	0.94	2.43	4.47
	3400	0.51	—	1.08	5.25	8.60

A. Masker level effects

1. Supra-threshold masking

In general, higher masker levels typically produced larger threshold shifts than low-level maskers, consistent with studies that used long-duration maskers and short-duration probes (Zeng *et al.*, 2005; Kwon and van den Honert, 2009). As the masker level increases, the region of excitation expands, making the probe more difficult to detect. Although most subjects demonstrated some increase in masking from MC (comfortable) to ML (loud but comfortable), there was generally no statistically significant difference in the amount of masking between these two highest levels for the group data (see Figs. 5–7). Across all subjects, MC and ML levels were 68% and 85% of the masker dynamic range, respectively, for 1700 pps and 67% and 86% of the dynamic range, respectively, for 3400 pps. Similar to the present study, the individual data plots from Kwon and van den Honert (2009) showed variability across subjects in the amount of threshold shift at the highest levels. Their subjects demonstrated increased, decreased, or no change in threshold shifts across the highest levels, consistent with the present results. The literature shows that with electrical stimulation, the masker becomes less effective with increased level; this has been shown for forward masking (Chatterjee and Shannon, 1998; Nelson and Donaldson, 2002) and for interleaved maskers (Kwon and van den Honert, 2009). Masking effects in the present study were also reduced by the addition of a masker-probe offset (see Table II and Figs. 6 and 7), consistent with results reported by de Balthasar *et al.* (2003). In the present experiment, the masker and probe were delivered to adjacent electrodes with a very short temporal offset, so it is important to also recognize the role of pitch. Sequential stimulation of adjacent electrodes with short temporal offsets can create intermediate pitch percepts,

depending on the relative current amplitude delivered to each electrode (McDermott and McKay, 1994; Kwon and van den Honert, 2006a). For subjects with better neural survival, the interval containing the probe could potentially shift the pitch percept of the masker, making it sound different from the other two intervals. Pitch is also easier to differentiate at higher levels (McKay *et al.*, 1999; Pfingst *et al.*, 1999). As the masker level increases, its pitch percept becomes more distinct, potentially making any pitch shifts produced by the probe easier to identify. Thus, at higher masker levels, the probe level will not have to increase by as much as the masker level to be detected.

In the present data set, several subjects exhibited relatively flat masking functions (see R10, FS19, and N11 in Fig. 5 and C24, C36, and C39 in Fig. 6), indicating minimal masking effects overall, even at high levels. It may be that these subjects have better pitch discrimination abilities, which would allow for easier detection of the probe at low stimulus levels in the presence of a high-level masker. It has also been shown that electrode array design (e.g., perimodiolar versus straight) may affect measures of physiological channel interaction and pitch perception (Hughes and Abbas, 2006). In general, perimodiolar arrays (Cochlear) and electrode positioners (Advanced Bionics) are designed to keep the array close to the modiolus, reducing spatial spread of excitation that can be exacerbated with lateral-wall arrays. One subject with a flat-level function in this study had an electrode positioner (C24), and one had a perimodiolar array (R10). The remaining three subjects had mid-scalar arrays. It is therefore possible that a combination of factors including pitch discrimination ability and electrode design impacted the reduced masker level effects for individual participants.

2. Facilitation

Charge from multiple sub-threshold pulses can build up within the integrative time constant of the neural membrane, increasing the likelihood of an action potential. The time course for integration is generally less than 300–400 μs (e.g., Cartee *et al.*, 2000; Middlebrooks, 2004; Morsnowski *et al.*, 2006). In the present study, two pulses (each masker-probe pair) were delivered within 124 μs (Cochlear), 128 μs (Advanced Bionics, 0- μs offset), or 136.98 μs (8.98- μs offset). Additionally, the time between the offset of each probe pulse and the onset of the next masker pulse for the 3400-pps/ch stimulus was <300 μs (range: 170–230 μs ; see Fig. 1), which is also within the time frame for integration. If there was enough spatial overlap between the masker and probe electrodes for interaction to occur, then two pulses occurring at low levels within these short time periods should have resulted in facilitation. In contrast to the hypothesis, however, there was virtually no evidence of facilitation for low-level maskers across rates, devices, or conditions.

One consideration is that the present study used interleaved pulse trains with short temporal offsets, so the stimuli were not exactly interleaved. From a probabilistic standpoint, we would expect reduced probe thresholds in the presence of near-threshold maskers (ST, MT), regardless of whether the probe is exactly interleaved (effectively doubling the overall

stimulation rate) or offset so that the masker-to-probe and probe-to-masker intervals are not equally spaced (as in the present study). In either case, the number of stimuli is doubled in the presence of the masker, which should increase the probability of detection. The present results, however, do not support a probabilistic model because the thresholds for ST and MT (double the amount of stimuli) were essentially the same as the probe-alone thresholds, or in some cases, elevated.

Another consideration is that the short temporal offset design of the present study is similar to that used to achieve “virtual channels” or intermediate pitch percepts with interleaved pulse trains (e.g., McDermott and McKay, 1994; Kwon and van den Honert, 2006a). At supra-threshold levels, intermediate pitch percepts are produced with this type of stimulus by exciting a combination of neural populations: one group of neurons is stimulated by the first pulse/electrode, and an additional group of neurons is stimulated by the second pulse. The second group of neurons would be those that were either partially depolarized by the first pulse/electrode, or those that were outside the spatial range of the first pulse/electrode. This combined collection of stimulated neurons, if adequate in number, can produce pitch percepts that are distinct from those produced by either electrode alone because the spatial excitation pattern differs. If we consider the same mechanisms acting at lower levels, near threshold, the current level requirements for two electrodes should be lower than the current delivered to a single electrode to recruit the same number of neurons needed to produce threshold. However, the probe thresholds in this study for ST and MT stimulation were not lower than the probe-alone threshold, which does not support the idea that the dual-electrode stimuli were simply adjusted until threshold was achieved.

The present study used monopolar stimulation, which results in lower thresholds than for more focused stimulation modes such as bipolar or tripolar (e.g., Bierer and Faulkner, 2010). Current fields are smaller at lower levels, so there might not have been enough spatial overlap of excitation between adjacent electrodes to induce sufficient integration to result in a perceptual change in threshold. The single exception was subject C43 in the offset condition for 1700 pps/ch (see Fig. 7). As can be seen from the raw threshold data in Fig. 4, however, the probe quiet thresholds (Q) for this subject at 1700 pps/ch had a relatively large standard deviation. As a result, the negative threshold shift in Fig. 7 could simply be a result of large measurement variance.

An unexpected finding in this study was that probe thresholds were greatly reduced for two subjects (C8 and C29) when the masker was set at “comfortable” and/or “loud but comfortable” (see Figs. 3 and 6). These subjects were able to detect the probe in the presence of a high-level masker *below* the probe quiet threshold. This effect was observed at 1700 pps/ch for C8 and for both rates for C29. The pulse paradigm for the Advanced Bionics device consisted of a 0- μ s IPG and a 0- μ s masker-probe offset [see Fig. 1(C)], which is also the configuration used in clinical strategies for this device when stimulating at maximum per-channel rates. For subject C8, adding an 8.98- μ s masker-probe offset eliminated the rollover effect (see Figs. 4 and 7). For C29, this effect was

only eliminated for 3400 pps/ch. This outcome was only observed in Advanced Bionics subjects, and adding a temporal offset between the pulse pairs eliminated the rollover effect in two of three instances. It is therefore possible that consecutive biphasic pulses produced unconventional interactions for some recipients, which might be mediated by the amount of spatial overlap of stimulated neurons between the two electrodes.

Physiological studies have shown that symmetrical biphasic pulses are less effective than monophasic, pseudomonophasic (asymmetrical), or symmetrical biphasic pulses with an interphase gap (e.g., Shepherd and Javel, 1999). In the case of a cathodic-leading biphasic pulse [Fig. 1(C)], the cathodic phase depolarizes the neural tissue and then the anodic phase hyperpolarizes the tissue. There is a “vulnerable” period prior to action potential initiation in which the action potential can be suppressed by the second phase (Shepherd and Javel, 1999). Timing gaps in electrical stimulation, such as an interphase gap or an offset between biphasic pulses, allow the auditory nerve more time to initiate an action potential before the opposing phase is delivered (Shepherd and Javel, 1999; McKay and Henshall, 2003; Carlyon *et al.*, 2005; Ramekers *et al.*, 2014). However, the second phase does not completely eliminate the excitatory state produced by the first phase (Shepherd and Javel, 1999). Recent evidence suggests that the anodic phase is the most excitatory in the deafened human ear (e.g., Macherey *et al.*, 2010; Carlyon *et al.*, 2013). If two pulses are presented in succession with no temporal gap between them [as in Fig. 1(C)], the first phase of the second pulse (probe) could potentially reduce the effectiveness of the second phase of the first pulse (masker), assuming sufficient spatial overlap of the excitation patterns for each electrode. Thus, the flanking cathodic phases might have reduced the effectiveness of the anodic second phase of the masker, if the masker level was high enough to overlap with the excitation region of the probe. According to subject C8, the stimulus that evoked a rollover effect at the loud masker level sounded like a “buzz,” in contrast to the other conditions, which reportedly sounded like a “tone.” These qualitative differences are consistent with those reported by subjects in Landsberger and McKay (2005) for high rates of stimulation on a single electrode. Perceptual effects achieved with current steering to create “phantom electrode” stimulation involve phase inversion to change the shape of the current field used for stimulation (Saoji *et al.*, 2013). It is possible that interactions of the pulse phases at the level of the neural membrane in the present study created similar perceptual effects as those produced by “phantom” stimulation. It may also be that complex interactions between hyperpolarized and depolarized regions along the auditory nerve, such as collisions between orthodromic and antidromic propagation of action potentials (Rattay *et al.*, 2001a,b; Miller *et al.*, 2004) resulted in an atypical excitation pattern that changed the percept of the masker, making the presence of the probe more identifiable based on sound quality.

Only one other study has reported reduced probe thresholds (or a rollover) for higher masker levels, where the masker and probe stimuli were presented to the same electrode (Donaldson *et al.*, 2005). In that study, a rollover effect

was observed for each of five early-generation Advanced Bionics participants for various rates (100–2600 pps/ch) and masker levels (10%–90% of the dynamic range). There was no speculation regarding potential factors that contributed to the rollover effects in that study. Finally, [Tong and Clark \(1986\)](#) reported reduced thresholds with a “comfortable-level” masker; however, that was for wider electrode separations (beyond 2 mm or three electrodes) with interleaved pulses. As previously noted, current strength decays with distance; therefore current amplitudes can reach threshold or sub-threshold levels at the spatial edges of the current field, potentially resulting in facilitation.

One of the disadvantages of a 3IFC task in which the listener selects the interval that sounds different is that judgments can be based on any dimension across listeners or even across trials within listeners. As a result, listeners might have made judgments based on pitch, loudness, or quality. Each of these aspects might have different underlying neural representations, which could complicate interpretation of the data.

B. Effect of stimulation rate

It was hypothesized that 3400 pps/ch would yield less effective masking than 1700 pps/ch due to stochastic properties of the auditory nerve ([Hughes et al., 2012](#)). This was not observed for any of the group data in this study (see Figs. 5–7, group mean plots), in contrast to the results reported by [Kwon and van den Honert \(2009\)](#). For the Cochlear group, mean threshold shifts were virtually identical for the two rates, yielding no interaction between rate and level (see Fig. 5). The Advanced Bionics data showed a trend that was opposite of that hypothesized (see Fig. 6); however, this result was likely driven by the subjects with rollover effects at the highest masker levels (C8, C29, C39, and C44). An analysis completed with these four subjects removed resulted in threshold shifts that were similar across the two rates (data not shown).

Although ECAP data show differences in neural response patterns between the range of rates tested here ([Hughes et al., 2012](#); [Hughes et al., 2014](#)), it may be that both 1700 and 3400 pps/ch were sufficiently fast to produce a relatively stochastic state at high masker levels with the interleaved paradigm. When the probe is interleaved with a 1700 pps/ch masker pulse train, the total number of pulses per period is doubled to 3400 pps (although not periodically spaced in the present study). Assuming substantial overlap of excitation patterns at high levels, these neurons will effectively be stimulated at double the masker rate when the probe is presented. In [Kwon and van den Honert \(2009\)](#), the two fastest rates tested were 2000 and 6410 pps, and both rates exhibited similarly shallow growth of masking. These results suggest minimal effects of rate on masking for relatively fast-rate pulse trains.

At low stimulus levels, in contrast, there should be less interaction at the peripheral level between the masker and probe if there is sufficient spatial separation between excitation patterns for the adjacent electrodes. As a result, limited temporal integration of the masker and probe pulses would

be expected. If temporal integration were to occur, then the 1700 pps/ch probe interleaved with the 1700 pps/ch masker at threshold (MT) might approximate the 3400 pps/ch probe quiet (Q) threshold. As can be seen from the group mean plots in Figs. 2–4, this was not the case for the two offset conditions (Figs. 2 and 4), which suggests little to no temporal integration of stimuli between the masker and probe at low levels. In contrast, [McKay et al. \(2001\)](#) found the largest loudness summation effects for lower levels (50% of the dynamic range, as opposed to C-level) and for the faster stimulation rate (1000 pps, as opposed to 500 pps) for stimulation of adjacent electrodes. That study, however, did not test interactions at threshold.

To further investigate the general lack of rate effects on the amount of masking observed in Figs. 5–7, additional data were collected using 250-pps pulse trains for the masker and probe [otherwise similar to Fig. 1(B)] in a subset of six ears with Cochlear devices (R3, F1, F11, FS19, N5, and N11). These data are shown in Fig. 8. Note that the 1700-pps and 3400-pps data only include the same six ears for which the 250-pps data were collected. Physiological data for rates as slow as 250 pps (as used with Cochlear’s SPEAK strategy) demonstrate relatively complete refractory recovery between subsequent pulses ([Wilson et al., 1997](#)). As a result, we would expect less temporal integration and potentially less masking for slow rates. The top panel in Fig. 8 shows the mean masked (ST, MT, MC, and ML) and unmasked (Q) probe thresholds for each of the three rates; data are plotted similar to the group data plot in Fig. 2. As expected from known temporal integration effects, thresholds are highest for the slowest rate (250 pps) and lowest for the fastest rate (3400 pps). The bottom panel shows the data from the top panel expressed as probe threshold shift (in dB), similar to the group data plot in Fig. 5. Overall, the threshold shifts were smaller and the growth of masking was shallower for 250 pps than for the two faster rates, consistent with the expectation that slow rates produce less temporal integration and less masking. It would be worthwhile for future studies to explore masking effects for a wider range of rates than those assessed here.

C. Clinical implications

In contrast to previous studies that used forward masking (e.g., [Lim et al., 1989](#); [Chatterjee and Shannon, 1998](#); [Chatterjee, 1999](#); [Boëx et al., 2003b](#); [Cohen et al., 2003](#); [Chatterjee et al., 2006](#); [Kwon and van den Honert, 2006b](#); [Nelson et al., 2008](#); [Hughes and Stille, 2008](#)) or short-duration probes interleaved with longer-duration maskers ([Tong and Clark, 1986](#); [de Balthasar et al., 2003](#); [Zeng et al., 2005](#); [Kwon and van den Honert, 2009](#)), the present study was designed to assess channel interactions using a stimulation paradigm that is more clinically relevant. Results suggest that threshold- or sub-threshold-level stimulation does not result in facilitation. This is particularly relevant to MED-EL and Advanced Bionics devices, in which map T-levels are typically set to sub-threshold current levels. According to [Holden et al. \(2011\)](#), T-levels are set to 10% of the respective M-levels in the Advanced Bionics devices

FIG. 8. Top: Group mean probe thresholds in CL (± 1 standard deviation) are plotted as a function of masker current level (re: loudness rating) for three stimulation rates in a subset of six ears with Cochlear devices (compare to the group data plot in Fig. 2). Bottom: Data from the top panel are expressed as probe threshold shift in dB (re: $1 \mu\text{A}$) for each of the four masker levels (compare to the group data plot in Fig. 5).

because the average CI recipient has an electrical dynamic range of approximately 20 dB, which results in an M/T-level ratio of 10. If a recipient's electrical dynamic range was less than 20 dB, then setting T-levels to 10% of the M-level might result in soft sounds being presented at a level that is inaudible. If facilitation did occur, it might be beneficial for making sounds audible at levels that would otherwise be below the behavioral threshold (but above the default T-level setting) for a given electrode. Results also showed that levels that are representative of the upper edge of the map dynamic range (i.e., MC for Advanced Bionics M-levels and ML for Cochlear C-levels) produce masking effects for adjacent electrodes. It is important to recognize that the temporal offsets between masker and probe in the present study represented the minimum possible temporal offsets between pulses that can be produced with each device's maximum stimulation rate and maximum number of electrodes. Longer temporal offsets between pulses will occur for slower per-channel rates, fewer electrodes or maxima enabled in the clinical map, or for strategies that stagger firing order across

non-adjacent electrodes (e.g., Fidelity 120, HiRes Optima). A comparison between the 0- and 8.98- μs offset conditions in the Advanced Bionics subjects suggests that these interactions are reduced with longer temporal offsets between successive pulses. These findings suggest that temporal gaps between stimulation of subsequent electrodes should be utilized to reduce channel interactions.

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¹The default for T-levels in the Advanced Bionics software is 10% of the M-levels. For MED-EL, the default is zero charge units.

²The default rate for Cochlear's Advanced Combination Encoder (ACE) strategy is 900 pps/ch and Advanced Bionics' HiRes-S is 2900 pps/ch. For HiRes, the per-channel rate increases as electrodes are disabled; for example, if two electrodes are disabled, the rate increases to 3314 pps/ch.

³The exception is MED-EL's Fine Structure strategies, which utilize different rates on the most apical channel(s).

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