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Forward masking in different cochlear implant systems

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The goal of this study was to evaluate, from a psychophysical standpoint, the neural spread of excitation produced by the stimulation of different types of intracochlear electrode arrays: the Ineraid™, the Clarion™ S-Series on its own or with the Electrode Positioning System (EPS), and the Clarion™ HiFocus-I with the EPS. The EPS is an independent silicone part designed to bring the electrode array close to the modiolus. Forward masking was evaluated in 12 adult subjects (3 Ineraid™, 4 Clarion™ S-Series, 3 Clarion™ S-Series+EPS, 3 HiFocus-I+EPS) by psychophysical experiments conducted using trains of biphasic stimuli (813 pulses per second, 307.6 μs/phase). Masker signals (+8 dB *re*: threshold, 300 ms) were applied to the most apical electrode. Probe signals (30 ms, 10-ms postmasker) were delivered to more basal electrodes. Masked and unmasked detection thresholds of probe signals were measured. For both Clarion™ HiFocus-I subjects, measurements were conducted in both monopolar and bipolar stimulus configurations. No major differences were found in forward masking between the different intracochlear electrode arrays tested in the monopolar configuration at suprathreshold levels equivalent to those used in speech-coding strategies, but significant differences were found between subjects. A significant negative correlation also was found between the level of forward masking and the consonant identification performance. These measurements showed that the neural spread of excitation was more restricted in the bipolar configuration than in the monopolar configuration for HiFocus-I subjects. It was found that CIS strategies implemented without using apical electrodes, which showed high levels of masking, could improve consonant identification. [DOI: 10.1121/1.1610452]

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I. INTRODUCTION

Existing multichannel cochlear implant systems allow adult patients that became deaf after the acquisition of language to recover oral speech communication. Cochlear implants also allow congenitally as well as prelingually deaf children to adopt oral language as a main mode of communication (Svirsky *et al.*, 2000). Some of the efficiency of multichannel cochlear implants is thought to be due to the stimulation selectivity of each of the intracochlear electrodes. This stimulation selectivity contributes to the frequency selectivity of the acoustic sounds that cochlear implant systems transmit to the central auditory system. We expect that if each electrode excites a limited population of auditory-nerve fibers, the stimulation of each electrode should provide spectral information. At present, this selectivity and its effect on speech reception is poorly understood.

The excitation spread produced by electrical stimulation can be measured with forward-masking experiments (Shannon, 1983a, 1983b; Tong and Clark, 1986; Lim *et al.*, 1989; Cohen *et al.*, 1996; Chatterjee and Shannon, 1998; Throckmorton and Collins, 1999). Shannon (1983a) proposed that the level of masking obtained in forward masking was due in part to the overlap of the population of fibers excited by the masker with the population of fibers excited by the probe. Indeed, the nature of forward masking is the result of different neural masking phenomena. Neural masking can be due to refractory effects of the auditory-nerve fibers or to central effects. The refractory effects are due to the fact that (1) fibers cannot produce a spike in response to an excitation while not having partially recovered from the generation of a previous spike or that (2) fibers need stronger excitation to produce a spike in response to an excitation while not having completely recovered from the generation of a previous spike see (Miller *et al.*, 2001 for a review). If the delay between the masker and the probe is long enough (about 7 ms), the refractory effect should not influence the results of forward-masking experiments. Central effects remain poorly understood but they certainly play an important role in forward-masking results obtained with electric hearing (Shannon, 1990; Shannon and Otto, 1990; Nelson and Donaldson, 2001, 2002).

In order to analyze the stimulation selectivity (the spread of excitation) of different intracochlear electrode arrays, we measured forward masking at levels similar to those used in sound coding strategies. We measured forward masking at comfortable hearing levels (masker: 8 dB above threshold) in a group of 12 subjects. We conducted forward masking for subjects who received the Ineraid™ electrode array, the Clarion™ S-Series electrode array with or without the Electrode Positioning System (EPS), and the Clarion™ HiFocus-I electrode array with the EPS. The EPS is an independent silicone part designed to place the electrode array close to the modiolus. Forward masking was measured successively in different electrodes for a masker applied on the most apical electrode available for stimulation. For both us-

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ers of the Clarion™ HiFocus-I electrode arrays (for whom the masker level could be reached within the stimulator-compliance limits), we also measured forward masking for bipolar stimulus configuration. Forward masking was not measured in bipolar configuration for Ineraid subjects as this implant was not designed to support this stimulus configuration. In this case, the Ineraid implant would present only 5 channels and very large bipolar pairs (electrodes are 4-mm distant). They were also not conducted in Clarion™ S-Series because of compliance limits in these cases. It was not possible to reach suprathreshold levels (+8 dB) in bipolar configuration. In order to analyze the effect of forward masking on speech reception performances, all subjects were also subjected to consonant identification tests. In addition, the information gathered through these experiments was used to adapt CIS strategies (Wilson *et al.*, 1991) for subjects who showed apical electrodes with high levels of masking.

II. METHODS

A. Subjects

All 12 subjects who participated in this study also participated in previous electrical interaction experiments [Boëx *et al.*, (2003)], Three subjects received the Ineraid™ implant (I03, I09, and I34), four subjects (C05, C06, C24, and C30) the standard Clarion™ S-Series electrode array, three subjects (Cp08, Cp14, and Cp18) the S-Series electrode with EPS, and two subjects (H26 and H29) received the HiFocus-I electrode with EPS. All subjects except subjects Cp08 and Cp18 were bilaterally, profoundly, and postlingually deaf. Subjects Cp08 and Cp18 suffered from severe deafness in the contralateral ear. Within this group different etiologies of deafness were found (Mondini, Ménière's syndrome, traumatic, streptomycin, multiple otitis, viral labyrinthitis, otosclerosis, congenital, or unknown). At the time of the study, subjects were between the ages of 39 and 73 years old (mean: 53 years and 10 months). These experiments were conducted within the ethical guidelines of the Declaration of Helsinki.

B. Experiment design

Detection thresholds were measured for the probe alone, the masker alone, and the probe in presence of the masker. This was carried out using a 3-alternative-forced-choice (3AFC) ''one up/two down'' adaptive procedure converging to a level where approximately 70.7% of stimuli are detected (Levitt, 1971). The subjects were asked to indicate in which interval they perceived the probe signal in addition to the masker. Feedback was provided. Each measurement started with a clearly detectable signal. One detection threshold measurement was obtained after gathering six reversals (descending and ascending segments). Signals were decreased initially by steps of about 15% of the initial amplitude. For the second and third reversals the steps were halved. One threshold was computed as the mean of the last four descending segments of the staircase procedure. The detection thresholds indicated in the present study were calculated as the mean of at least two threshold measurements. Forward masking was measured for biphasic stimuli (813 pulses per second, 307.6 μs/phase, Fig. 1). The masker (300 ms) was set at 8 dB above the detection threshold of the masker electrode. It was clearly perceived and comfortable for all subjects. This level of masker was expected to be high enough to allow the observation of the masking effect on a probe electrode distant more than 6 mm away from the masker. Masker was applied to the most apical electrode, and its effects on the detection thresholds of the probe signals on other electrodes were measured. The probe signal (30 ms) was presented 10 ms after the end of the masker. This delay was applied to limit refractory effects and residual polarization effects on the forward-masking measurements.

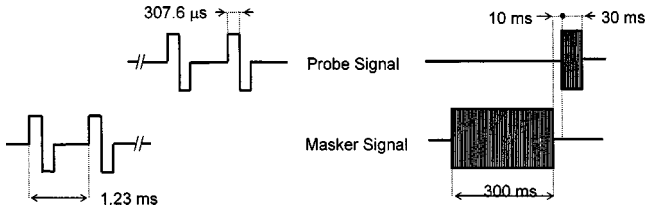

FIG. 1. Scheme of the masker and probe signals used in the forward-masking measurements. Stimuli were all biphasic (813 pulses per second, 307.6 μs/phase). Masker signals were 300-ms duration (zero rise–decay times). Probe signals were 30-ms duration (zero rise–decay times). Probe and masker signals were always first positive biphasic pulses. The probe signal was presented 10 ms after the setting off of the masker signal.

The electric stimuli were generated using the Clarion™ Research Interface (CRI, Wygonski *et al.*, 1999), which allows custom control of all stimulation parameters. The current amplitudes delivered through intracochlear electrodes were not exactly proportional to the requested currents over the entire electrical output range. They depended slightly on electrode impedance. These limitations can lead to measurement error in psychophysical experiments. Therefore, we devoted special attention to improve the control of the symmetry and the amplitude of current biphasic pulses. For this, we used special look-up tables based on calibrations. One table was built measuring, for each clinical unit, the effective current delivered by the current source of the Research Implantable Cochlear Stimulator (RICS). We defined from these tables the clinical units required to provide the desired current amplitudes. These tables were defined for each electrode impedance reported for all subjects in Boëx *et al.* (2003). Impedances were evaluated from the clinical setup impedance measurements[1] (SCLIN) using sine waves (1000 Hz) at very low levels. We also evaluated impedances of Ineraid™ electrodes using sine waves (1000 Hz) at very low levels. In monopolar configuration, maximum impedances and currents used in the present study were low (respectively, 18 kΩ and 60 μA). In these limits current sources behaved almost as perfect current sources. As an example, if impedance decreased from 18 to 13 kΩ (27% lower), stimulation amplitude would increase from 60 to 62 μA. In bipolar configuration, because the HiFocus-I subjects were SAS (Simultaneous Analog Strategy, Battmer *et al.*, 1999; Osberger and Fisher, 1999) speech strategy users and, because that strategy stimulates all electrodes, monopolar impedance of each electrode used could be considered and summed to define the bipolar impedances. For these subjects, maximum bipolar impedances and currents used in the present study

were low (24 kΩ and 130 μA or 33 kΩ and 100 μA). In these limits current sources behaved almost as perfect current sources. As an example, if impedance decreased from 24 to 18 kΩ (27% lower) or from 33 kΩ to 24 kΩ (27% lower), stimulation amplitude would increase from 130 to 139 μA or from 100 to 109 μA, respectively.

To conduct these measurements in Ineraid subjects, we connected directly the channel output of the RICS provided with the CRI to the Ineraid™ electrodes via their percutaneous plug.

C. Electrode types

Three different types of electrode arrays have been tested.

The Ineraid™ electrode array has six intracochlear platinum electrodes. Each electrode is a platinum 0.5-mm-diameter ball. The distance between the centers of electrodes is about 4 mm. In monopolar configuration, the stimuli were applied between one intracochlear electrode and an external electrode used as a far-field ground electrode, placed under the temporalis muscle. Electrodes were numbered from the most apical electrode to the most basal pair from 1 to 6. The masker signals were applied to electrode 1 (most apical) and the probe signals were successively applied to electrodes 1, 2, 3, and 4. Thus, the nominal distance between the probe and the masker electrodes was 0, 4, 8, and 12 mm, respectively.

The preformed Clarion™ S-Series electrode array (Kessler, 1999) was implanted either alone or with the Electrode Positioning System (EPS). The Clarion™ S-Series electrode array has eight pairs of electrodes (one medial, one lateral). Each electrode is an iridium–platinum (90:10) 0.3-mm-diameter ball. The distance between each pair is 2 mm. Each pair consists of one medial and one lateral electrode positioned radially, 0.6 mm distant from each other. Electrode pairs are numbered from the most apical electrode pair to the most basal pair from 1 to 8. In monopolar configuration, the stimuli were applied between a medial electrode and the base of the implanted receiver–stimulator case used as a far-field ground in the Clarion™ S-series system. The masker signals were applied to medial apical electrode 1 and the probe signals were applied successively to medial electrodes 1, 2, 4, 6, and 8. Thus, the nominal distance between the probe and the masker electrodes was 0, 2, 6, 10, and 14 mm, respectively.

The Clarion™ HiFocus-I electrode array has 16 electrodes. Each contact is rectangular in shape (0.4×0.5 mm) and made from pure platinum. The distance between the centers of electrodes is 1 mm. This array was implanted with the EPS. Electrodes are numbered from the most apical electrode to the most basal from 1 to 16. In the monopolar configuration, stimuli were applied between one electrode and the base of the implanted receiver–stimulator case used as a far-field ground in the Clarion™ HiFocus system. The masker signals were applied to electrode 2 and the probe signals were applied successively to electrodes 2, 4, 8, 12, and 14. Thus, the distances between the probe and the masker electrodes were 0, 2, 6, 10, and 12 mm, respectively. In the bipolar configuration, stimuli were applied between two adjacent electrodes, with the more basal electrode of the pair used as the ground electrode. The masker signals were applied between electrodes 2 and 3. The probe signals were applied successively between electrodes 4 and 5, 8 and 9, 12 and 13. Thus, the nominal distance between the probe and the masker electrode pairs was 2, 6, and 10 mm, respectively.

D. Consonant identification tests

Speech reception was evaluated through closed-set medial consonant tests (Pelizzone *et al.*, 1993) and initial consonant tests. Each test consisted in the presentation of 56 tokens using the 14 French consonants /b, d, f, g, k, l, m, n, p, r, s, t, v, and z/, in the form ''aBa,'' ''aDa,'' etc. Tokens were presented in a random order and feedback was not provided. Medial consonant tokens were spoken by one male speaker. Initial consonant tokens were spoken by two female and two male talkers.

Speech tests were conducted without visual cues. Subjects were seated 1 m from the loudspeaker (Fostex™ UP203 S) in a sound-proof chamber (IAC 1201A). Tests were played from the Turtle Beach™ Pinnacle™ Pro Series sound card. Sound levels were adjusted with an EMB™ P 300 amplifier. The overall level of the tokens was about 75 dB peak SPL A. Scores were expressed in percent of correctly identified consonants.

E. Speech processor

The Ineraid subjects had been using a CIS strategy for at least 4 years (Geneva Wearable Processors, Pelizzone *et al.*, 1995, 1999). The Clarion subjects who received the S-Series electrode array (with or without the EPS) had been using the standard Clarion™ CIS strategy. The Clarion subjects who received the HiFocus-I electrode array had been using a SAS strategy. The Clarion subjects had been using their implant for at least 1 year and for less than 4 years (mean:2 years).

Subjects C05, Cp08, Cp14, Cp18, and C30 tested different CIS strategies, implemented with reduced numbers of stimulation channels. All strategies were implemented using the clinical Clarion™ (SCLIN) platform. Each time a channel was switched off, the same input frequency range was shared across the remaining electrodes and the rate of stimulation maximized.

III. RESULTS

A. Forward-masking experiments

The masker signals were presented on the most apical electrode that could be stimulated. Masker levels were set at 8 dB above the masker threshold. The masker amplitudes used for each subject are reported in Table I.

In Fig. 2, the thresholds for the probe alone and the masked probe conditions are plotted as a function of the distance between the masker and the probe electrodes for monopolar stimulation and in Fig. 3 for bipolar stimulation. The nominal distances (abscissa) between the probe electrode and the masker electrode were determined from the electrode designs. The probe thresholds (triangles pointing upwards) were generally lower than the masked probe

TABLE I. Amplitudes of masker signals set 8 dB above the threshold of the masker electrode in the monopolar and bipolar stimulus configurations. Suprathreshold levels (8 dB) could not be reached (because of compliance limits) in the bipolar configuration for S-Series subjects. Neural masking measurements were not conducted in bipolar configuration for Ineraid subjects.

Subjects	Monopolar masker signal (μA)	Bipolar masker signal (μA)
I03	55	···
I09	31	···
I34	68	···
C05	26	···
C06	25	···
C24	36	···
C30	32	···
Cp08	22	···
Cp14	19	···
Cp18	60	···
H26	21	79
H29	23	209

thresholds (triangles pointing downwards). In some cases, the thresholds for the probe alone and masked probe conditions were similar (e.g., subject H29 at 12 mm, subject H26 at 10 mm in monopolar configuration), indicating a negligible masking at those distances. In bipolar configuration, subject H26 did not show any differences between his probe thresholds and his masked probe thresholds, indicating an absence of forward masking.

Thresholds and masked thresholds could be very different across subjects, even for the same type of electrode array (e.g., subject Cp18 in comparison to subjects Cp08 or Cp14). Masking was computed as the ratio of masked probe thresholds to probe alone thresholds (masked probe threshold/probe threshold alone) in the monopolar (Fig. 4) and in the bipolar configurations (Fig. 5).

In the monopolar configuration, the largest masking was observed when the probe and the masker electrodes were both on the same most apical electrode (0 mm). The masking at 0 mm could be very different across subjects implanted

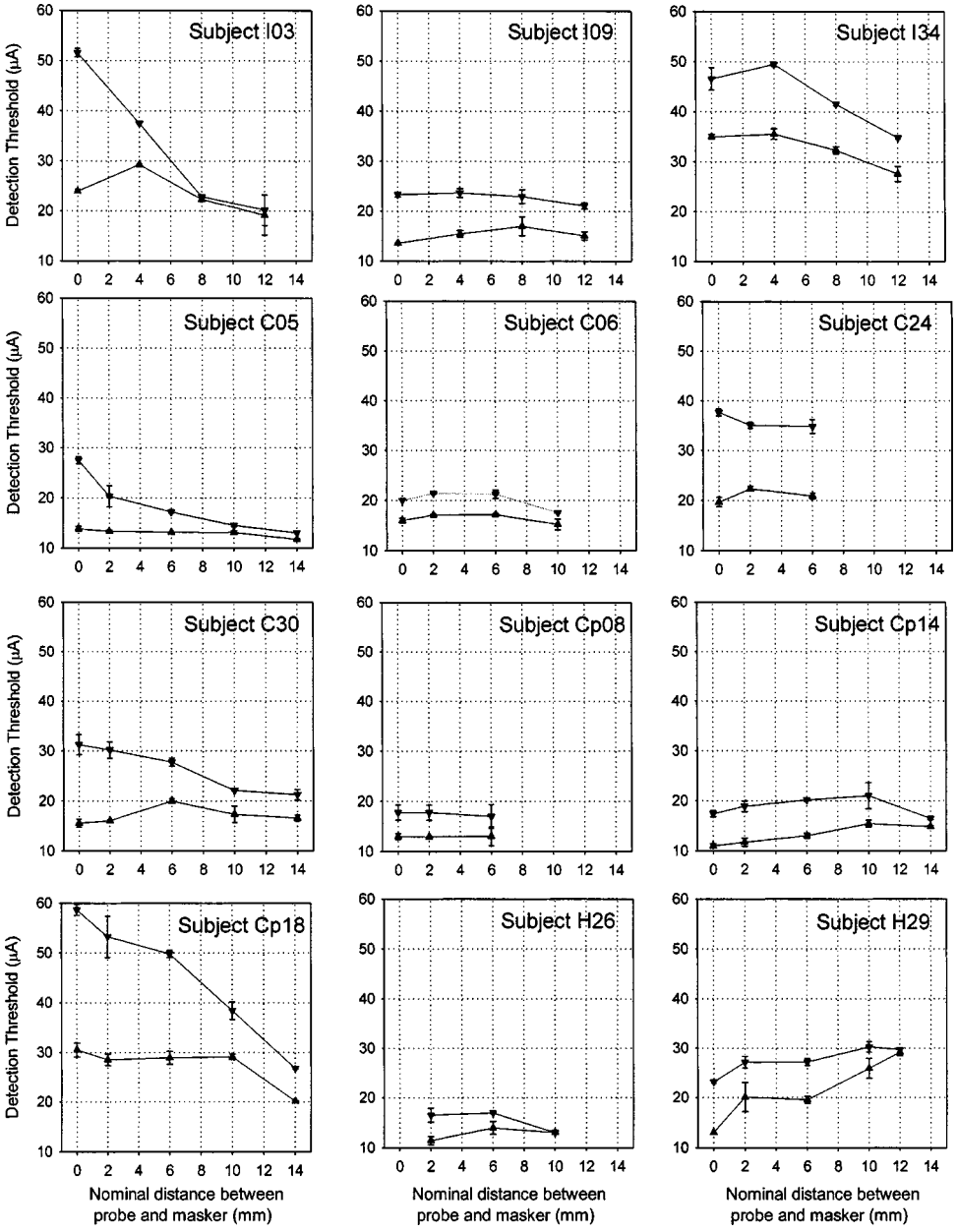

FIG. 2. Probe thresholds measured with and without masker in monopolar configuration. In all graphs, subject probe thresholds (μA) are identified by triangles pointing upwards. Masked probe thresholds (μA) are identified by the triangles pointing downwards. The nominal distance between the masker and the probe electrodes tested is indicated in mm on the abscissa.

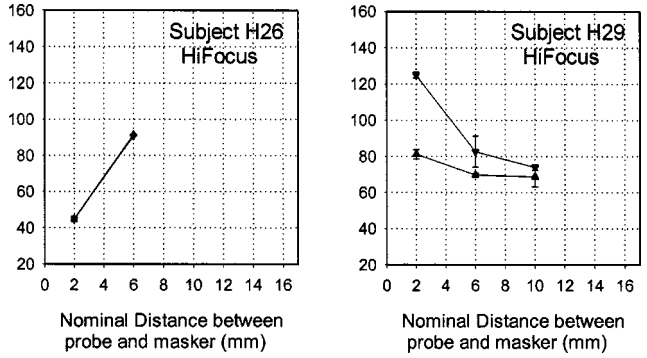

FIG. 3. Probe thresholds measured with and without masker in bipolar configuration. Subject probe thresholds (μA) are identified by triangles pointing upwards and masked probe thresholds (μA) are identified by triangles pointing downwards. The nominal distance between the masker and the probe electrodes tested is indicated in mm on the abscissa.

with the same electrode array (e.g., Cp08, Cp14, and Cp18). Overall, masking decreased as the distance increased between probe and masker electrodes (Table II), but it did not decrease identically for all subjects. For some subjects, masking did not decrease up to a 6-mm distance between probe and masker electrodes (Cp08,Cp14,C06,I34). With other subjects (C05,C24,H29), masking decreased significantly at 2 mm ($P<0.001$) in the monopolar configuration.

Overall masking computed from all electrode masking

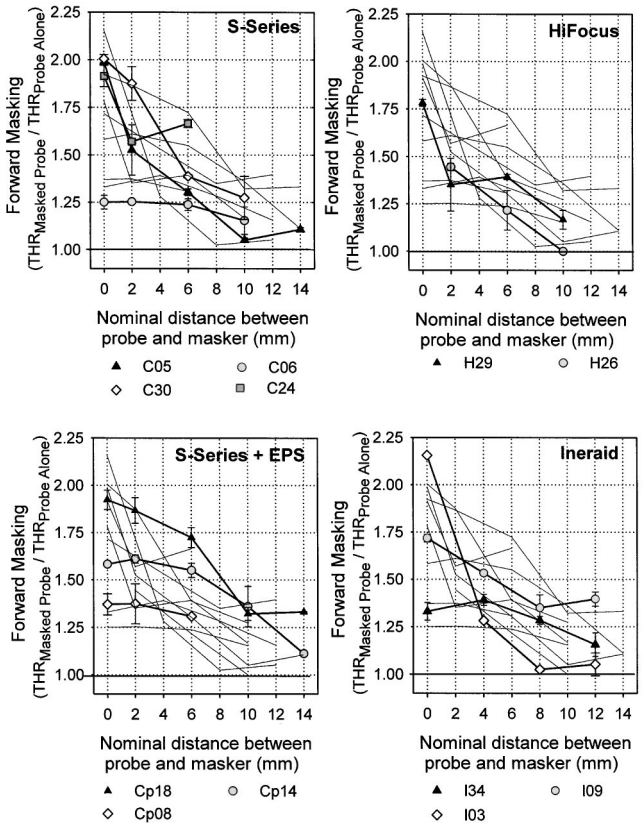

FIG. 4. Forward-masking measurements (ratio of masked probe thresholds to probe alone thresholds) obtained in the monopolar configuration. In all graphs the thin lines without symbols represent the curves obtained for all 12 subjects. In each graph, the forward masking obtained by subjects implanted with the specified electrode array is indicated by specified symbols and solid lines. Error bars are the standard errors of the mean.

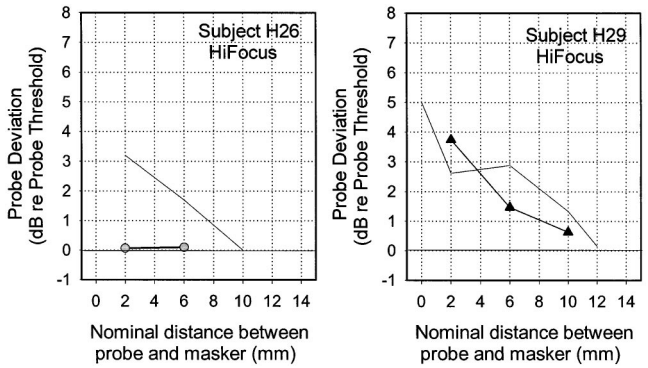

FIG. 5. Forward-masking measurements (ratio of masked probe thresholds to probe alone thresholds) obtained in the bipolar configuration for both HiFocus-I subjects. Masked threshold could not be reached (because of compliance limits) at 0-mm masker–probe separation (not measured for subject H26). The thin lines without symbols represent the curves obtained in the monopolar configuration by the same subject. Error bars are the standard errors of the mean.

was not significantly different for subjects implanted with the Ineraid than for subjects implanted with the S-Series with or without the EPS. Overall masking was not significantly different for subjects implanted with the S-Series alone or with the EPS.

Subject H26 (HiFocus-I) did not show masking in the bipolar configuration. In subject H29, masking in bipolar configuration was slightly higher at 2 mm and was slightly lower than in monopolar configuration for more distant probes.

B. Consonant identification scores

We analyzed the correlation between medial consonant identification scores and masking data. Consonant identification scores were expressed in percent of correct responses and were computed for at least six consonant tests obtained within two sessions. All subjects used the CIS sound coding strategy, except H26 and H29 who used a SAS strategy (seven channels in both subjects). All Clarion CIS-user scores were obtained with the standard eight channel Clarion™ CIS strategy. We used masking data obtained at 4 mm to consider masking for the same distance[2] between probe and masker electrodes for all different types of electrode arrays. To calculate masking at a distance of 4 mm between probe and masker electrodes we made a linear interpolation based on data obtained at 2 and 6 mm in Clarion subjects. The consonant identification scores with masking are shown in Fig. 6. We obtained a regression coefficient of -0.60 (R, Pearson product moment; $p<0.04$) which indicates that there was a statistically significant negative correlation between masking and consonant identification. The latter accounted, however, for only 36% of the variance.

C. Electrode selection for the CIS strategy

Subjects Cp18 and C30 showed high levels of forward masking at 2 mm (electrode 2) and for subject Cp18 interpolated forward masking was also high at 4 mm. We concluded that for these subjects, electrodes 1 and 2 stimulated partly the same population of fibers. Hence, we proposed to test a

TABLE II. Average monopolar masking (masked probe threshold/probe threshold alone) for masker to probe distances ranging from 0 to 14 mm. These average changes were computed from all available electrode measurements. Standard errors (SE) are indicated.

Nominal distance between masker and probe electrodes	0 mm	2 mm	4 mm	6 mm	8 mm	10 mm	12 mm	14 mm
Number of subjects	11	9	3	9	3	7	3	3
Mean changes	1.73	1.54	1.40	1.42	1.22	1.19	1.20	1.18
(SE)	(0.09)	(0.07)	(0.07)	(0.06)	(0.10)	(0.05)	(0.10)	(0.07)

seven-channel CIS strategy that eliminated the use of electrode 1. We suggested further to subject Cp18 to test a six-channel CIS strategy that eliminated electrode 2 in addition to electrode 1. Subject Cp18 tested each strategy for 2 weeks. His initial consonant identification scores increased (nonsignificantly) from 50.86% (5.52) to 53.5% (5.2) with the seven-channel processor and significantly increased to 58.25%$(4.27; P<0.05)$ with the six-channel processor (Fig. 7). He chose then to adopt the six-channel CIS processor permanently. Subject C30 tested the seven-channel strategy for 2 weeks. Her initial consonant identification scores increased (nonsignificantly) from 40.4% (6.8) to 46.4% (3.21) with the seven-channel strategy. When she tested a CIS strategy implemented with six channels, the initial consonant scores were not significantly different, varying from 40.4% (6.8) to 45.8% (3.19). She chose then to adopt the seven-channel strategy permanently.

We also tested CIS strategies implemented with seven channels for subjects C05 and Cp08. The latter, differently from the former two subjects, presented lower masking at 2 mm. Their initial consonant identification scores decreased significantly $(P<0.05)$, even after 2 weeks of daily use. They reverted to their initial standard eight-channel strategy.

IV. DISCUSSION

We did not find differences in monopolar masking between the groups of subjects using the three types of electrode arrays under study (Ineraid™, Clarion™ S-Series, and HiFocus-I). We noted, however, significant forward-masking differences across subjects (e.g., Cp18 and C08). Important similar variations across subjects were also observed by Chatterjee and Shannon (1998) and by Cohen *et al.* (in bipolar configuration, 1996). For example, subject C06 presented a low level of forward masking, indicating that the tested electrodes did not stimulate significantly overlapping neural populations. Some subjects like subject Cp18 showed important masking at 0 mm which did not decrease significantly when the probe was switched from electrode 1 to electrode 2. In his case, if forward masking was due in part to the extent of the population of fibers excited, this would suggest that the population of fibers excited by electrode 1 was similar to the population of fibers excited by electrode 2. In agreement with this hypothesis, consonant tests showed that subject Cp18 performed better when the CIS-speech processor was fitted without using the two most apical electrodes, while subject Cp08 did not. In addition, we also observed that subject Cp18 assigned about the same 100-Hz tone pitch [acoustic to electric pitch-comparison experiments in Boëx *et al.*,

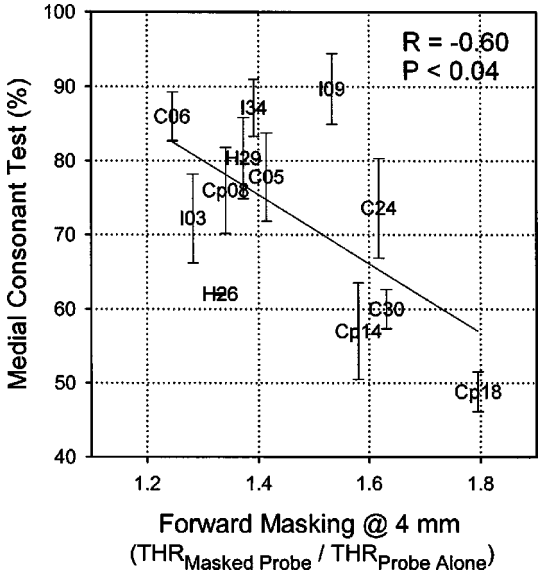

FIG. 6. Medial consonant identification scores are expressed in percent of correct responses versus forward masking measured at 4 mm for Ineraid subjects and calculated at 4 mm from 2- and 6-mm data for Clarion subjects. Scores are the mean percentage of correct responses computed on six consonant tests (6×56 tokens), collected in two different test sessions. Error bars are standard deviations.

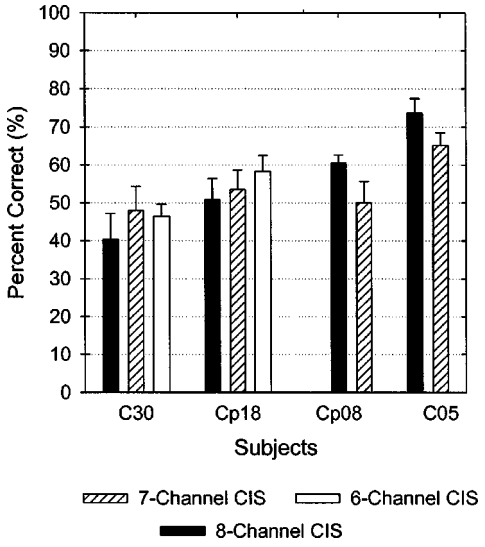

FIG. 7. Initial consonant identification scores are expressed in percent of correct responses for S-Series subjects who tested CIS strategies implemented with different numbers of channels. Each processor was used daily for at least 2 weeks. Scores are the mean percentage of correct responses computed on three consonant tests (3×56 tokens). Error bars are standard deviations.

(submitted)] to electrodes 1 and 2. In contrast, we observed that subject Cp08, who showed low masking levels between electrodes 1 and 2, assigned to electrode 1 a 383-Hz tone pitch and to electrode 2 a 505-Hz tone pitch. This observation is in agreement with the hypothesis that in the case of subject Cp08, electrode 1 and 2 stimulated different populations of fibers as indicated by the low masking levels observed. When we reduced the number of channels subject Cp18 showed improvements in his performance. Zwolan *et al.* (1997) made a similar observation when they reported improvements in speech perception for some subjects who tested speech-coding strategies that did not use nondiscriminable electrodes. The improvements we obtained discarding the most apical electrodes could also be in agreement with the proposition of Shannon *et al.* (2001) that signals should be delivered to pitch-matched electrodes to allow for the best speech reception, particularly for apical electrodes. The negative correlation shown in our data between forward masking and consonant reception suggests that the overlap of the populations of fibers excited may limit speech reception. Throckmorton and Collins (1999) also obtained significant correlations between average forward masking and some speech recognition test scores.

We compared forward masking between monopolar and bipolar configurations in both HiFocus-I subjects. In the bipolar configuration the forward masking was lower for both subjects, suggesting that the use of a bipolar configuration would significantly improve the selectivity of HiFocus-I stimulation. Improvement of selectivity offered by a bipolar configuration was observed previously for different intracochlear electrodes. Tong and Clark (1986) and Tong *et al.* (1987) also observed a decrease in masking with increases in distances between masker and probe electrodes in users of the Nucleus electrode array with bipolar configuration. Shannon (1983b) conducted forward-masking measurements for sinusoid stimuli in one subject, using different bipolar and monopolar electrode pairs placed at different positions in the cochlea and for different stimulus levels. He reported better spatial selectivity in the bipolar configuration than in the monopolar configuration. Lim *et al.* (1989) also conducted experiments showing that larger spatial extents of bipolar pairs resulted in spatially larger patterns of masking.

V. CONCLUSION

We did not find group differences in the monopolar selectivity of stimulation among the subjects wearing the Ineraid™, the Clarion™ S-Series, and the HiFocus-I electrode arrays. But, we found important differences between subjects implanted with the same types of electrode array. We found a statistically significant negative correlation between the level of forward masking and the initial consonant identification performance in our group of 12 users, implanted with different cochlear implant systems. One subject performed better when the most apical electrodes presenting high levels of masking were not used in CIS processors. Forward-masking experiments can help determine the selectivity of intracochlear electrode stimulation and select electrodes to be used in sound-coding strategies for each subject. This is even more relevant with present cochlear implants, which have more intracochlear electrodes.

In addition, we found low levels of masking in bipolar stimulation for both HiFocus-I subjects, indicating a better selectivity of stimulation in that configuration. Interestingly, both subjects had adopted the bipolar simultaneous analog strategy (SAS) since the beginning of their implant use. The use of more selective electrical stimulation channels should increase the number of effective spectral channels and should allow better implementation of simultaneous strategies.

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[1]Tables were built for our RICS; they could not be built for each subject-implanted stimulator as we could not measure current amplitudes as well as impedances on implanted electrodes with our present research interface. We were not aware of any reports on description of impedance changes with amplitude of stimulation in Clarion™ S-Series or HiFocus-I electrodes.

[2]This choice was made because masking could not be measured at 2 and 6 mm in Ineraid subjects and because we did not want to use the 0-mm data.

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