Auditory Cortical Images of Cochlear-Implant Stimuli: Dependence on Electrode Configuration

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Bierer, Julie Arenberg and John C. Middlebrooks. Auditory cortical images of cochlear-implant stimuli: dependence on electrode configuration. J Neurophysiol 87: 478–492, 2002; 10.1152/jn.00212.2001. This study examines patterns of auditory cortical activity elicited by single-pulse cochlear implant stimuli that vary in electrode configuration, cochlear place of stimulation, and stimulus level. Recordings were made from the primary auditory cortex (area A1) of ketamine-anesthetized guinea pigs. The spatiotemporal pattern of neural spike activity was measured simultaneously across 16 cortical locations spanning approximately 2–3 octaves of the tonotopic axis. Such a pattern, averaged over 40 presentations of any particular stimulus, was defined as the “cortical image” of that stimulus. Acutely deafened guinea pigs were implanted with a 6-electrode animal version of the 22-electrode Nucleus banded electrode array (Cochlear). Cochlear electrode configurations consisted of monopolar (MP), bipolar (BP + N) with N inactive electrodes between the active and return electrodes (0 ≤ N ≤ 4), tripolar (TP) with one active electrode and two flanking return electrodes, and common ground (CG) with one active electrode and as many as five return electrodes. Cortical images typically showed a focus of maximum spike probability and minimum latency. Spike probabilities tended to decrease, and latencies tended to increase, with increasing cortical distance from that focus. Cortical images of TP stimuli were the most spatially compact, followed by BP + N images, and then MP images, which were the broadest. Images of CG stimuli were rather variable across animals and stimulus channels. The locations of cortical images varied systematically from caudal to rostral as the cochlear place of stimulation changed from basal to apical. At the most sensitive cortical site for each condition, the dynamic ranges over which spike rates increased with increased current level were restricted to about 1–2 dB, regardless of configuration. Dynamic ranges tended to increase with increasing cortical distance from the most sensitive site. Electrode configurations that produced compact cortical images (e.g., TP and BP + 0) showed the greatest range of thresholds within each cortical image and the largest dynamic range at cortical sites removed from the most sensitive site.

INTRODUCTION

Electrical stimulation of the auditory nerve through a cochlear prosthesis elicits a perception of sound in patients who have severe to profound deafness. The usual input to a cochlear prosthesis consists of environmental sounds processed to select features that carry speech information, then converted to patterned electrical signals that directly stimulate the auditory nerve. The performance of prostheses users in understanding speech varies greatly, although the reasons for the variation are not well understood.

One parameter of the speech processing strategy that is known to influence speech-recognition performance is the electrode configuration. Electrode configuration is the spatial arrangement of the electrodes carrying the current for each channel of a cochlear prosthesis. The configuration of the electrical stimulus determines in part the extent of current spread in the cochlea and the number of auditory nerve fibers activated (Kral et al. 1998; Spelman et al. 1995; van den Honert and Stypulkowski 1984, 1987). The design of many modern cochlear prostheses has been based on the hypothesis that a more restricted current field in the cochlea activates a more spatially restricted population of auditory nerve fibers, thereby providing a better definition of the place of stimulation and a greater number of independent channels of information to the brain. That hypothesis is supported by limited psychophysical evidence indicating that more restricted current fields permit more accurate discrimination of cochlear place of stimulation (i.e., better channel discrimination) (Henry et al. 2000; Townshend et al. 1987). A further hypothesis is that a greater number of independent information channels to the brain permits improved speech-recognition performance. That hypothesis is supported by evidence that, under certain conditions, subjects’ speech recognition correlates with their ability to discriminate neighboring cochlear-implant channels (Henry et al. 2000; Nelson et al. 1995; Throckmorton and Collins 1999). Also, speech recognition is sometimes improved by reprogramming speech processors to avoid activation of indiscernible implant channels (Zwolan et al. 1997).

Several lines of psychophysical evidence predict, therefore that more restricted current fields would result in improved speech recognition. Nevertheless, many cochlear-prosthesis subjects prefer electrode configurations that produce relatively broad current fields (Kileny et al. 1998; Zwolan et al. 1997), and some patients show better speech recognition using broader current fields (Pfingst et al. 1997). Technical limitations, limitations in subject time, and concerns about subject safety discourage exhaustive tests in humans of effects of electrode configuration on speech reception. For those reasons, we have developed an animal model for study of information transmission from the cochlear implant to the auditory cortex.
We implant six-channel cochlear implants in anesthetized guinea pigs and then record the patterns of cortical activity that are elicited by various configurations of cochlear stimulation. Our goals are to identify electrode configurations that optimize information transmission to the cortex, and then to apply new understanding to the design of speech processing strategies for human patients.

We refer to the distribution of cortical activity across cortical place and poststimulus time that results from a particular stimulus as the “cortical image” of that stimulus. In the present report, we describe the cortical images of stimuli that varied in electrode configuration, place of cochlear stimulation, and stimulus level. The results suggest that electrode configuration has an impact on the accuracy of cochlear-place representation and on the dynamic range of electrical stimulation. In the companion report, we quantify the accuracy with which cortical images code the place of stimulation and stimulus current level.

METHODS

Anesthesia and surgery

Data were collected from 10 healthy adult pigmented guinea pigs (500–900 g). In each animal, unilateral deafening, cochlear implantation, and cortical recording were performed in a single, approximately 16-h session. Animals were anesthetized with a subcutaneous injection of a mixture of ketamine hydrochloride (40 mg/kg) and xylazine (10 mg/kg). Additional intramuscular injections of a mixture of ketamine and xylazine were given as needed to maintain an anesthetic state. Core body temperature was maintained at 38°C with a thermostatically controlled heated pad. A tracheal cannula was inserted. A head holder was mounted to the skull anterior to bregma.

The left bulla was accessed using a postauricular approach. Deafening was achieved by puncture of the round window, withdrawal of a small amount of perilymph with a wick, then slow infusion of 60 μl of 10% neomycin sulfate (approximately 0.1 M) into the scala tympani; typically, >2 h passed between the neomycin infusion and the beginning of cortical recording. The intracochlear electrode array was a six-electrode scala tympani cochlear implant (provided by Cochlear, Englewood, CO), with a ground wire positioned in a neck muscle. Aside from the reduced number of electrodes, the implant was similar to the 22-electrode Nucleus implant that is used clinically. The implant electrodes were platinum iridium bands centered at 750-μm intervals. The electrodes were numbered from 1 to 6 from base to apex. The implant was inserted into the scala tympani through a cochleostomy approximately 1 mm beyond the round window. Data from the most basal one or two electrode(s) were excluded if the threshold for that electrode differed from a middle electrode by more than 6 dB. For that reason the number of working scala tympani electrodes varied among subjects; there were four in four animals, five in four animals, and six in two animals. Based on the characteristic-frequency-to-cochlear-place scale factor by Greenwood (1990), the 3.75-mm center-to-center distance spanned by a six-electrode array would correspond to 1.43 octaves of characteristic frequency.

The temporalis muscle was reflected and the skull exposed on the right side. A small hole was made in the dura over the primary auditory cortex. The recording probe was then inserted through the dural opening. The cortical surface was covered with agarose (20 mg agarose per ml of Ringer solution). All procedures were in accordance with policies of the University of Michigan University Committee on Use and Care of Animals.

Stimulus generation

Experiments were controlled by an Intel-based personal computer interfaced with Tucker-Davis hardware (Tucker-Davis Technology, Gainesville, FL). Stimuli were generated using custom software written in MATLAB script (Mathworks, Natick, MA). A D/A converter controlled a custom-made optically isolated constant current source with a capacitor-coupled output. Experiments were conducted in a sound-attenuating chamber.

Electrical stimuli consisted of single biphasic, charge-balanced pulses. Phase durations were 200-μs/phase pulses, except when stated otherwise. The polarity was initially cathodal at the active electrode. Stimuli were presented at a rate of 1 pulse per 700 ms.

Various electrode configurations of the electrical stimulus were employed. Electrode configuration refers to the physical configuration of electrical sources and sinks (i.e., active and return electrodes) in a cochlear prosthesis. In the monopolar (MP) configuration the active electrode was a single intra-scalar electrode, and the return was through a wire positioned in a neck muscle. In the bipolar configuration the active electrode was one intra-scalar electrode and the return was a second, more apical, intra-scalar electrode. In bipolar configurations, BP + 0, BP + 1, BP + 2, and BP + 3, there were 0, 1, 2, or 3 inactive electrodes separating the active and return electrodes, respectively, resulting in a center-to-center spacing of 750–3,000 μm from active to return electrodes. The BP + 0 configuration usually is referred to as “BP” in the implant literature. In the tripolar (TP) configuration the active electrode was a single intra-scalar electrode, and the return consisted of the two adjacent electrodes, each carrying one-half of the return current. Based on electric-field models and physical measurements (Kral et al. 1998; Spelman et al. 1995), we presume that the cochlear extent of electrical stimulation at a constant current level ranked from diffuse to focal in the order MP, BP + 3, BP + 2, BP + 1, BP + 0, TP. We also tested the common ground (CG) configuration that consisted of a single intra-scalar active electrode and a return that comprised the remaining N intra-scalar electrodes, each carrying I/N of the current.

In this report, a “channel” refers to an active electrode and its compliment of return electrode(s). The number of channels that was available for testing varied among configurations because of the varying number of return electrodes. For instance, given six cochlear electrodes there were six MP and CG channels but only four TP channels. The channel number corresponds to the number of active electrode; in keeping with the clinical convention, we number bipolar channels by the more basal electrode. For all electrode configurations, stimulus current levels were varied over a range from below threshold to 10 to 20 dB above threshold in 1-dB steps. During tests of each electrode configuration, current levels and stimulus channels were interleaved from trial to trial. Every combination of current level and channel number was presented once in random order, then every combination was repeated in a different random order until each stimulus combination was tested 40 times. The interleaving of current levels and stimulus channels minimized the effects of any long-term drift in neural sensitivity or in the number of active units contained within a multi-unit cluster. Stimuli in various electrode configurations were not interleaved, but the order of testing of various configurations was varied among animals.

Multi-channel recording and spike sorting

The activity of cortical neurons was recorded with silicon-substrate thin-film multi-channel recording probes (Center for Neural Communication Technology, Ann Arbor, MI) (Drake et al. 1988; Najafi et al. 1985). Each probe had 16 recording sites along a single shank at intervals of 100 μm (center to center). The shank was 15 μm thick, 4.55 mm long, and tapered in width from 100 to 15 μm over the 1.5-mm segment containing the recording sites. The multi-channel probe permitted simultaneous recording of spike activity from all 16 sites. The impedance at each site was 1.5–4 MΩ.
The recording probe was positioned with a micromanipulator. The probe penetrated the cortex from dorsocaudal to ventrostral, with the wide axis of the probe perpendicular to the cortical surface (i.e., with the width of the shank roughly parallel to the radial cell columns). The probe penetrated the somewhat-convex cortical surface 3–4 mm from the site of interest in area A1 and was advanced so that the 1.5-mm array of recording sites was in the middle cortical layers, roughly parallel with the cortical surface. We adjusted the depth of the probe so that spike activity was recorded at all 16 sites. Presumably, that corresponded to the depth of cortical layers III and IV, which were most likely to be active in the anesthetized condition. Histologically, layers II, III, and IV of cortical area A1 typically appear fused and are difficult to distinguish in Nissl-stained material (Rose 1949). Those layers occupy roughly 600 μm of cortical depth in guinea pigs (unpublished observation). From geometric considerations, the 16 recording sites at any probe placement would have been restricted to a range of ±500 μm whenever the angle of the penetration was within ±10° of the desired angle parallel to the cortical surface, a condition that was fairly easy to meet. Physiologically, the stimulus threshold and response latency at each of the 16 recording sites tended to vary with cochlear place of stimulation, but the ranges of minimum-to-maximum thresholds and latencies tended to be similar across all 16 recording sites in each animal. That observation further supported the inference that all recordings were from similar cortical layers.

Probe penetrations were oriented roughly parallel to the tonotopic gradient along which the frequency tuning of neurons changes most rapidly. In the guinea pig area A1, neurons sensitive to high frequencies (basal cochlea) are situated dorsocaudally, and low-frequency neurons (apical cochlea) are situated ventrostral (Arenberg et al. 2000; Hellweg et al. 1977; Redies et al. 1989; Wallace et al. 2000). Prior to detailed study at any recording-probe location, tuning properties of rostral, middle, and caudal cortical sites were estimated by observing responses to BP + 0 stimuli on the most apical and basal stimulus channels. This analysis allowed us to verify proper probe placement relative to the cochleotopic map in area A1. If the reverse cochleotopic order was detected, indicative of the dorsocaudal field (area DC) (Redies et al. 1989), the probe was retracted and placed further rostral in area A1. Based on our previous study that used acoustic stimulation (Arenberg et al. 2000), we estimate that best frequencies of units sampled at 16 recording sites of a single probe placement spanned 2 to 3 octaves. We collected data from one recording-probe placement in each animal.

Signals from the recording probe were amplified with a custom 16-channel amplifier, digitized at a 25-KHz rate, sharply low-pass filtered below 6 kHz, resampled at a 12.5-KHz sample rate, and then stored on the computer hard disk. The first 8 ms after stimulus onset were excluded from analysis because that period often was contaminated by the stimulus artifact and by brain-stem evoked potentials. Unit activity was isolated from the digitized signal off-line using custom spike-sorting software (Furukawa et al. 2000). We sometimes encountered well-isolated single units, but most recordings were of unresolved clusters of a small number of units. Spike times were stored at 20-μs resolution for further analysis of multi-unit spike patterns. Recordings at particular sites were excluded from further analysis if units did not respond to any stimulus with an average of ≥1 spike/trial or if the spike rate averaged over one presentation of all the stimulus conditions changed by more than a factor of two over the entire recording period.

Data analysis

Spike rates at each recording site were normalized by the following procedure. At each recording site, we averaged the spike rate across 40 trials for each stimulus channel, current level, and electrode configuration. From the distribution of mean spike rates, we took the 5th and 95th percentile rate as the spontaneous rate and maximum rate, respectively. The normalized spike rate at each recording site was computed by subtracting the spontaneous rate and dividing by the maximum rate minus the spontaneous rate. By normalizing in this manner we emphasized stimulus-driven changes in activity rather than absolute spike numbers across channels. Within each electrode configuration, the stimulus current level that elicited a normalized spike rate of 0.25 was taken as the threshold for each recording site. Similarly, the stimulus current level that elicited a normalized spike rate of 0.75 defined the saturation level for each recording site. Figure 1 gives an example of the normalized spike rate for one recording site as a function of stimulus level in response to a monopolar configuration stimulus. The filled circles represent levels at which normalized spike rates were 0.25 (threshold level) and 0.75 (saturation level). The range of stimulus levels corresponding to normalized spike rates of 0.25–0.75 were used to estimate the dynamic range of a site. These rather conservative criteria for threshold and saturation level yielded dynamic ranges that emphasized the steepest portion of the neural rate-level functions, providing a measure that was fairly immune to variation in the spontaneous and maximum rates. In pilot studies, estimates of threshold and saturation levels based on a maximum-likelihood analysis sometimes produced thresholds at normalized rates of <0.25 (and saturated levels at more than 0.75), but trial-by-trial variability often forced the threshold-normalized-rate criterion to near 0.25.

The distribution of cortical activity across all recording sites and across time was referred to as a cortical image. The cortical image of any particular stimulus was derived from simultaneous recordings at 16 cortical sites. The activity across all sites reflected the response to the same stimulus, averaged across 40 trials. The threshold of a cortical image for a particular stimulus was defined as the threshold level at the most sensitive site recorded across the 16 recording sites. The centroid of the cortical image was defined as the spike-rate-weighted center of mass calculated from all the sites at which the firing rate was above threshold; the centroid computation collapsed spike rates across all time bins. The centroid computation differed from that used previously (Arenberg et al. 2000) in that the previous computation only considered spike rates that were ≥50% of the maximum. In many cases, the region of supra-threshold sites extended to, and presumably beyond, one or both ends of the recording probe. In those cases, the cortical images were simply truncated; that is, there was no attempt to estimate the extent of activation beyond the recording probe. Centroids computed from such truncated cortical images presumably were biased somewhat toward the location of the center of the recording probe. Also, cortical image widths were somewhat underestimated in those cases.

FIG. 1. Rate-vs.-level function. The abscissa and ordinate represent current level and normalized spike rate, respectively. The filled circles represent the threshold (25%) and the saturation (75%) points that were used to compute the dynamic range. Data are from 1 recording site from animal 0001.

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RESULTS

We obtained detailed measurements of patterns of cortical activity from 10 probe placements in 10 animals. Stable recordings of small clusters of units were obtained from 14 to 16 sites at each probe placement, a total of 158 sites. We first review the sensitivity of cortical neurons to stimuli that varied in stimulus level, cochlear place of stimulation, and electrode configuration. Next, we characterize the distribution of spike activity across cortical place and poststimulus time (cortical images of cochlear implant stimulation). Finally, we examine dynamic ranges across the array of cortical neurons.

Thresholds for electrical stimulation

The threshold at the most sensitive cortical recording site in response to a particular stimulus was taken as the threshold for that stimulus. Thresholds depended on phase duration, electrode configuration, basal-to-apical cochlear location, and animal. Figure 2 shows the distributions of thresholds across those independent variables. Individual data points are shown for the BP + 3 condition (Fig. 2B) and the most basal channel (Fig. 2C). Otherwise, each box plot represents the distribution across 11–45 cases. Conditions of phase duration 200 μs/phase, configuration BP + 0, and the most apical electrode were used as references. The median threshold for those reference conditions for 10 animals was 134 μA, ranging from 79 to 355 μA. The dependence of threshold on phase duration is shown in Fig. 2A. For each electrode configuration, cochlear channel, and animal, the threshold at each phase duration was expressed relative to the corresponding threshold for the 200-μs/phase stimulus. Threshold was higher for the short phase duration and lower for the long phase duration relative to the 200-μs/phase stimulus (P < 0.001, 2-way ANOVA). Thresholds increased roughly with the logarithm of the phase duration, approximately 4.7 dB/doubling, somewhat less than the 6-dB/doubling rate expected from perfect integration of electrical current. The decrease in threshold from 200 μs/phase to 1,000 μs/phase tended to be less than that from 40 μs/phase to 200 μs/phase, consistent with greater leakage of electrical charge for the longer phase durations. Aside from changes in thresholds, changes in phase duration did not produce any consistent change in patterns of cortical activity. Therefore phase duration was held constant at 200 μs/phase for all remaining analyses. The dependence of threshold on electrode configuration is shown in Fig. 2B. For each stimulus channel and animal, the threshold for each electrode configuration was expressed relative to the corresponding threshold for the BP + 0 configuration. Thresholds tended to increase in order of presumed decreasing spread of cochlear excitation, from MP to BP + 0 (P < 0.001) to TP configurations (P < 0.025). Also, thresholds increased with decreasing extent of bipolar stimulation for BP + 2 to BP + 1, and BP + 1 to BP + 0 (P < 0.001). The small number of BP + 3 channels that were available for testing renders statistical comparisons meaningless. Nevertheless, we often observed a lower threshold for the BP + 3 configuration than for the MP configuration, and thresholds were significantly lower for BP + 2 than for MP configurations (P < 0.001). With increasing separation of the BP electrodes from 0 to 3, thresholds decreased at a rate of approximately 3.4 dB/channel. The dependence of threshold on basal-to-apical channel location is shown in Fig. 2C. For each electrode configuration and animal, the threshold for each basal-to-apical channel location was expressed relative to the corresponding threshold for the most apical stimulus channel. Thresholds increased with increasing distance relative to the most apical channel: the differences between distances 0 and 1, 1 and 2, and 2 and 3 were significant (P < 0.005).

Rate-versus-current functions and channel sensitivity at single cortical sites

Most spike rates increased monotonically with increasing current level. Responses were considered nonmonotonic if an increase in current resulted in a decrease in the maximum spike rate to <75% of the maximum rate. Across all 2,128 combinations of prosthesis configurations, channels, and cortical

**FIG. 2.** Distribution of relative thresholds across phase durations, electrode configurations, and cochlear channel locations. Individual data points are plotted for the BP + 3 configuration (middle panel) and the most basal channel (right panel). Otherwise, there were 11–45 cases for each condition, and the distribution is represent by a box plot. The horizontal lines represent 5th, 25th, 50th, 75th, and 95th percentiles. The plus symbols represent outlying data points. A: distribution of thresholds for 40, 200, and 1,000-μs/phase stimuli relative to the threshold at 200 μs/phase across animals, configurations, and stimulus channels. B: distribution of thresholds for various electrode configurations relative to the threshold for the BP + 0 configuration, across animals and stimulus channels. C: distribution of thresholds for various stimulus channels relative to the threshold for the most apical stimulus channel, across animals and electrode configurations. CG, common ground; TP, tripolar; BP, bipolar; MP, monopolar.
We refer to the characteristic spatiotemporal distribution of cortical spike activity elicited by any particular stimulus as the cortical image of that stimulus. Figure 4 presents cortical images of as many as six cochlear implant channels (columns) in four electrode configurations (rows). Panels are labeled according to configuration and active channel number. For instance, “BP + 0_3” indicates the BP + 0 configuration with 3 as the active electrode. In these examples, each stimulus was presented at 2 dB above threshold. The vertical axis of each panel represents the location along the cortical recording array relative to the most caudal recording site, the horizontal axis represents time after stimulus onset, and colors represent the normalized spike probability. In the example shown in Fig. 4, the cortical images of BP + 0, TP, and CG stimulation each contained a restricted focus of activity at which sites responded with highest probability and shortest latency. The cortical images of most MP stimuli were truncated at the caudal end of the recording array. For MP, BP + 0, and TP electrode configurations, the focus of activation shifted from caudal to rostral in the cortex as the stimulated cochlear implant channel shifted from basal to apical. In the CG configuration, the cortical images of the most apical channels (4, 5, and 6) were fairly restricted and showed a caudal-to-rostral progression for increasingly apical channel number. The images of more basal CG channels (1, 2, and 3) were wider and showed little or no cochleotopic progression. Note that even in the case of the TP configuration, which showed the most restricted cortical images, there was overlap between neuronal populations that were activated by cochlear channels that were as far as three channels apart.

An example of cortical images obtained from a second animal is presented in Fig. 5. This example shows more prominently a systematic variation of first-spike latencies within each image. That is, in most images one can see a cortical site showing a minimum latency with latencies increasing with distance from that cortical site. Again, the cortical images of MP, BP + 0, and TP stimulation shifted monotonically from caudal to rostral for successively more apical cochlear stimuli. As in the previous example, cortical images of apical channels in the CG configuration (4, 5, and 6) were fairly restricted and showed a caudal-to-rostral progression, whereas the image of the most basal channel (3) was wider and centered further rostral than would be predicted based on the trend of the other channels.

Cortical images increased monotonically in width in response to increasing current levels. Figure 6 presents cortical images of stimuli presented with a fixed active channel and various current levels. Each panel represents the cortical image of a particular electrode configuration (rows) and stimulus level (columns) relative to threshold. Images of MP, BP, and CG expanded to span nearly the entire recording array at the...
7-dB level, whereas the image of the TP stimulus remained restricted within the recording array even at high current levels. Note that images of this relatively apical channel tended to expand preferentially toward the more caudal cortical sites, which is the direction of the representation of more basal stimuli.

In many cases, cortical images for more apical electrodes tended to be narrower than for basal electrodes for any given configuration and current level relative to threshold. That trend was difficult to quantify because of the tendency of cortical images of basal channels to extend off the caudal end of the recording array. Nevertheless, the trend can be seen in the examples shown in Figs. 4 and 5 and in other examples (not illustrated).

As the extent of cochlear stimulation was broadened from that of TP to MP, the width of the activated cortical area increased. Cortical image widths were quantified by computing the area under plots of normalized spike rate versus cortical place, then dividing the area by the maximum normalized spike rate. The result was the width of a rectangle of unit height with area equal to the area under the spike-rate function. Figure 7 represents the image width versus stimulus level for six animals. Each symbol represents a different electrode configuration. Five of the six panels represent responses to electrical stimulation of one active stimulus channel (either channel 3 or 4 as indicated in each panel). Within each electrode configuration, image width increased as the stimulus level was increased; in some cases the image width saturated at a value determined by the length of the recording probe. Image widths also broadened as the electrode configuration was changed from TP to BP + N to MP. Among the BP + N configurations, there were instances in which image widths broadened with increasing N, but that trend was not observed consistently. The widths of CG images were relatively narrow in two animals (0001 and 0010), similar to those of TP, and were relatively broad in two other animals (0030 and 0031), similar to those of MP. The bottom right panel represents data from a previous paper that described responses to acoustical stimuli (Arenberg et al. 2000). In the acoustical cases, images of tones and noisebands expanded considerably as the stimulus level was increased as well as when the bandwidth was increased from that of a tone to a 1-octave narrowband noise.

It is difficult to compare the image widths of acoustical and
electrical stimuli because of the difficulty in equating the stimulus levels between them. Nevertheless, the acoustical data appear to show a saturated image width that is somewhat smaller than the saturated width for any of the electrical data. Moreover, it appears that none of the electrode configurations reliably activated a cortical area as restricted as that activated by an acoustical tone.

Topography of cortical images

In all electrode configurations, the focus of maximum cortical activity varied in location according to the cochlear place of stimulation. We quantified the location of the focus of activity by the centroid (see METHODS section). Figure 8 represents centroids as a function of the cochlear stimulus channel for the BP + 0 configuration with each panel representing data obtained from one animal. Each symbol represents a particular current level. For low current levels, the centroid shifted roughly linearly from caudal to rostral as the cochlear location of the stimulus was changed from basal to apical. The plots of centroid versus channel tended to flatten at increasing current levels, largely because of a tendency of centroids of basal channels to shift rostrally. That tendency was due, in turn, to expansion of the image to and presumably beyond the caudal end of the recording array. If increases in current level caused images to grow symmetrically or even with a caudal bias, we would have recorded the expansion of only the rostral edge of the image and would have computed a rostral shift in centroid. Note that the cortical images of more apical channels, in animals 0001, 0030, and 0031, which tended to be restricted within the recording array, tended to show a slight caudal shift in centroid with increasing current level, as seen in Figs. 4 and 5.

The dashed lines in Fig. 8 represent results from our previous study that used tonal acoustical stimulation in normal-hearing guinea pigs; the slope is the median from 11 probe placements in 10 animals (Arenberg et al. 2000), and the vertical position is arbitrary. Sound frequencies were converted to cochlear place using the function derived by Greenwood (1990). The median scale factor from the acoustical data were 131-μm cortex per mm cochlea (range: 77- to 271-μm cortex per mm cochlea). For nine animals we computed the scale factor for cochlear implant stimulation for BP + 0 stimuli at 2 dB above threshold. The median electrical stimulation scale factor was 130-μm cortex per mm cochlea, ranging from 46- to 226-μm cortex per mm cochlea, a distribution similar to that obtained with acoustical data.

Figure 9 presents examples from one animal of cortical images of bipolar configurations that varied in spatial extent. The top two rows show conditions in which the basal electrode of the configuration was held constant and the spatial extent of the stimulus was increased by shifting the apical electrode increasingly more apical. Conventionally, channels are numbered according to the number of the basal electrode, so all the stimuli represented in the top two rows have the same channel number. As expected, increases in the spatial extent of the stimulus led to a slight increase in the width of the cortical images. In addition, there was a conspicuous rostral shift of the cortical images; that is, the centroids of the cortical images tended to follow the apical (return) electrode. In the bottom two rows of the figure, the apical electrode was held constant in location and the basal electrode was changed. In that case the location of the centroid remained constant. In a separate experiment (not illustrated), we tested the effect of reversing the polarity of the stimulus for a BP + 3 configuration, i.e., presenting the cathodal phase of the biphasic pulse first to the apical electrode. Reversing the polarity showed essentially no change in the location of the centroid of the cortical responses; the location of the apical electrode determined the location of the centroid, regardless of the polarity.
Figure 10 plots the cortical centroid versus the cochlear stimulus channel number for stimuli of various electrode configurations. Each symbol represents a different electrode configuration, and all stimuli were presented at 2 dB above threshold; again, the dashed line is based on responses to acoustical stimulation from the previous study (Arenberg et al. 2000). Each column of panels represents responses for one animal. The top row of panels represents responses from MP, TP, and CG configurations, and the middle row represents BP + N configurations with the cochlear channel defined conventionally by the basal electrode of the pair. The change in centroid location as a function of cochlear stimulation channel is monotonic for TP and BP + N configurations but not always for MP and CG. The relatively shallow slope of the MP function reflects the tendency of cortical images of MP stimuli to extend to the limits of the recording array and the resulting tendency of centroids to lie near the center of the array. The CG function tended to be nonmonotonic with the most basal channel centroid lying near that of the most apical channel centroid. The lowest row also represents BP + N configurations but with the
cochlear channel defined by the apical electrode of the pair. Plotted in that way, the functions for different configurations overlapped closely, indicating that the location of the apical channel dominated the location of the cortical centroid.

Dynamic range

Human cochlear implant subjects have a limited dynamic range of comfortable listening levels, typically a range of $<10$ dB of electrical current (Pingst et al. 1997). We explored features of cortical responses that might influence dynamic range. Figure 11 plots for one guinea pig the stimulus levels needed to elicit responses at various criterion spike rates. The abscissa and ordinate represent current level (dB re. 1 mA) and cortical place (mm), respectively. Contours represent current levels that elicited normalized responses of 0.25, 0.50, and 0.75. Thus the thickness of the black and dark gray region represents the dynamic range of current levels over which spike rates increased from 25 to 75% of the maximum rates at each cortical site. Rows and columns of panels represent electrode configurations and cochlear stimulus channels, respectively. The format of these plots is somewhat similar to that of spatial tuning curves defined by (Snyder et al. 1990). The contour plots differ from spatial tuning curves in that they show the thresholds, midpoints, and saturation points for each recording site, whereas spatial tuning curves plot only the threshold. Across configurations, one can see differences in the ranges of thresholds and in the dynamic ranges at individual cortical sites. Note that the contour plots for the CG configuration are quite variable among stimulus channels. Across all configurations, there was a tendency for apical channels to show more restricted contour plots than basal channels. Previous studies that used spatial tuning curves measured the width of the spatial tuning curve at 6 dB above threshold (Raggio and Schreiner 1999; Snyder et al. 1990). The dashed lines in this figure represent the 6-dB levels for these data. In most cases, stimuli 6 dB above threshold activated units across most or all of the sites on the recording probe.

The thresholds for MP stimuli were fairly uniform across the 16 cortical recording sites. The range of thresholds between the most sensitive and least sensitive sites averaged 3.13 ± 1.7 dB (mean ± SD) across subjects. Ranges of thresholds across cortical recording sites were wider for the other configurations, averaging 6.65 ± 3 dB for BP 0, 7.58 ± 3.57 dB for TP, and 4.68 ± 3.75 dB for CG.

Each configuration showed the narrowest dynamic range at the most sensitive cortical site (i.e., the site showing the lowest threshold for each stimulus channel). Dynamic ranges at the most sensitive sites (MSS) fell within a narrow range; mean dynamic ranges ranged from 1.12 to 1.83 dB across configu-

![FIG. 8. Centroids of cortical images for the BP + 6 configuration as a function of cochlear implant channel. Each panel represents centroid functions for 1 animal. Each symbol type represents 1 stimulus level relative to threshold.](image)

![FIG. 9. Cortical images of the BP + N configuration. The top 2 rows represent cortical images of BP + N stimuli at 1 and 3 dB above threshold. Electrode 2 was the basal (active) electrode, and the apical (return) electrode was shifted progressively more apical to produce increasing spatial extent (left to right). The bottom 2 rows also represent cortical images of BP + N stimuli at 1 and 3 dB above threshold. The apical (return) electrode was electrode 6, and the basal (active) electrode was shifted progressively more basal (left to right). Conventions as in Fig. 4. Data are from 0031.](image)
with SDs from 0.57 to 1.47 dB. Across all cortical sites, however, dynamic ranges showed greater difference among electrode configurations, with mean and SDs of 2.3 ± 1.4 dB for MP, 3.3 ± 2.1 dB for BP 0, and 3.6 ± 2.5 dB for TP.

Dynamic ranges at individual recording sites generally increased with increasing cortical distance from the MSS. Figure 12 represents the dynamic ranges for one animal as a function of the distance from the MSS. Symbols show mean ± SE for various electrode configurations. We compared the dynamic range at each cortical site with the dynamic range at the adjacent site 100 μM further from the MSS. Across all subjects, the dynamic ranges at sites further from the MSS were

**FIG. 10.** Centroids of cortical images for the various electrode configurations as a function of cochlear implant channel. Each column represents centroid functions for 1 animal, and rows represent centroid functions for various electrode configurations as indicated in the legends. Current levels were 2 dB above threshold. In the middle row, BP + N channels are identified according to clinical convention (i.e., by the basal electrode). In the bottom row, BP + N channels are identified by the apical electrode.

**FIG. 11.** Spatial tuning curves for various electrode configurations and cochlear stimulus channels. The abscissa and ordinate represent current level and cortical place, respectively. Each panel represents the responses to 1 electrode configuration (rows) and 1 stimulus channel (columns) across many current levels. Levels indicated by the black shading elicited responses between 25 and 50%, the dark gray levels elicited responses between 50 and 75%, and the light gray level elicited responses >75% of the normalized spike rate from each recording site. The dashed line represents the current level at 6 dB above threshold. Data are from 0031.
significantly larger ($P < 0.005$ for BP $+$ 0, TP, and CG; $P < 0.01$ for MP, paired $t$-test). The increase in dynamic range with increase in distance from the MSS was significantly greater for BP $+$ 0, TP, and CG than for MP ($P < 0.025$).

Cortical latencies

Cortical images often exhibited an increase in the first spike latency with increasing cortical distance from the centroid (see Fig. 5). We explored the possibility that the first spike latency might carry stimulus-related information independent of spike rates. The distribution of first spike latencies was generally skewed, having a long tail toward longer latencies, and the SDs of first-spike latencies tended to be larger for longer mean latencies. For that reason mean first spike latencies were represented by the geometric means. The first spike latency was computed for each recording site, averaged across each trial that showed a normalized spike rate $>0.25$. For each subject, electrode configuration, and recording site, we computed the correlation between mean first spike latency and normalized spike rate across all stimulus channels and current levels. The median correlation coefficient was similar across electrode configurations, ranging from $R = -0.73$ to $-0.92$, with an overall median of $R = -0.82$. That indicates that in many cases the first spike latency correlated closely with the spike rate. For one-half of the cases, however, the spike rate accounted for less than approximately 67% (i.e., $0.82^2$) of the stimulus-related variance in first spike latency.

**DISCUSSION**

The results of this study demonstrate that the cortical images of cochlear electrical stimuli presented with various electrode configurations vary in cortical location, width, and dynamic range. We evaluate those results in reference to previous work and discuss the cortical images of particular cochlear electrode configurations. Quantitative evaluation of the accuracy of coding of place of stimulation and current level is deferred to the companion paper (Middlebrooks and Bierer 2002).

**Tonotopic organization**

The primary auditory cortex exhibits a well-known tonotopic (or cochleotopic) organization in which neurons are sharply tuned for the frequencies of sounds and in which the best frequencies of neurons vary systematically across the cortex [e.g., Hellweg et al. 1977 (guinea pig); Lauter et al. 1985 (human); Merzenich et al. 1975 (cat)]. Although most studies have evaluated the representation of acoustical frequencies, it is interesting to note that the first physiological demonstration of this organization was based, not on acoustical stimulation, but on electrical stimulation of the cochlea (Woolsey and Walzl 1943). In that study, electrical stimulation was delivered to small groups of nerve fibers in the osseous spiral lamina of anesthetized cats, and cortical-surface evoked potentials were measured. That study demonstrated that nerve fibers from specific regions of the cochlea projected through multiple synaptic levels to corresponding regions of the auditory cortex in a systematic manner (Woolsey and Walzl 1943). More recently, functional optical imaging techniques also have been used to demonstrate cochleotopic organization of electrical stimuli in cat (Dinse et al. 1997) and guinea pig (Taniguchi et al. 1997).

The previous study most similar to the present one was by Raggio and Schreiner (1999), who examined single and multi-unit responses in cat auditory cortical area A1 to cochlear electrical stimulation. Results from that study were expressed as the topographical cortical distributions of thresholds. Typically, two parallel bands of low-threshold responses were found, separated by a band of high-threshold responses. Each low-threshold band exhibited a cochleotopic map of the cochlear place of stimulation. In the present study in the guinea pig, we consistently found a single area of low-threshold responses. The caudal-to-rostral cortical position of the minimum threshold shifted respectively with the basal-to-apical place of cochlear stimulation. We extended the previous observation to show that, at supra-threshold stimulus levels, the focus of maximal cortical activity (represented by cortical centroids) also followed a cochleotopic organization. The distribution of factors that scaled cochlear place onto cortical place in the present study closely overlapped the distribution of scale factors measured using acoustical stimuli (Arenberg et al. 2000).

We tested bipolar electrode configurations with longitudinal extents (from active to return electrodes) that ranged from 0.75 mm (i.e., BP $+$ 0) to 3.75 mm (BP $+$ 4). In some instances, the widths of cortical images increased with increasing electrode extent, but that was not seen in all animals. The electrical model presented by Spelman and colleagues (1995) predicted that a broad bipolar stimulus would produce two discrete foci of activity in the cochlea, which we presume would lead to discrete foci of cortical activation. Our results failed to confirm that prediction for bipolar extents as wide as 3.75 mm. The image of every bipolar stimulus consisted of a unimodal distribution of spike activity versus cortical place. Rebscher and colleagues (2001) show at least one example of a 4-mm bipolar configuration that produced two spatially discrete threshold minima in the central nucleus of the inferior colliculus in cat, but apparently that finding was an exception rather than the norm. Similarly, Raggio and Schreiner (1999) illustrate one...
case of a 6-mm bipolar configuration that produced two local minima in the cat area A1.

In the present study, the centroids of cortical activity in the guinea pig cortex were determined more by the cochlear position of the apical electrode of a bipolar pair than by that of the basal electrode (Figs. 9 and 10). The clinical convention for human implant users is to name a channel by the basal electrode of a BP + N pair. The finding of the current study suggests that, by that convention, stimuli with nominally identical channels would activate varying cortical sites depending on the spatial extent of the electrode configuration (i.e., the N). In the guinea pig cochlea, apical electrodes appear to fit more closely to the modiolus than do the basal electrodes. This proximity to the neural elements might produce a greater current density at the apical electrode of a bipolar pair, thereby causing that electrode to dominate the cochlear activation. This dominance of the apical electrodes was sometimes observed in longitudinal bipolar configurations in the cat (Rebscher et al. 2001).

A recent study in humans provides insight into the perceptual significance of the basal and apical electrodes of a bipolar pair. Pfingst and colleagues (2001) tested speech recognition using one set of channels in the BP + 0 configuration that spanned a central region of the cochlea and another set that was located further basal. The more central set provided significantly better speech recognition. With the BP + 6 configuration, better speech recognition was obtained with a set of channels that was located further basal. The nominally “active” basal channels of the BP + 6 stimuli were well basal to the active channels of the central set of BP + 0 channels, but the basal-to-apical spans of the BP + 6 stimuli were roughly centered on the central BP + 0 set. The finding in humans suggests that the cochlear location of the apical return electrode of a bipolar pair has a substantial impact on the percept elicited by a bipolar stimulus.

The CG configuration produced inconsistent cortical images. Cortical images of apical channels were narrow and showed a caudal-to-rostral progression with increasing channel number, whereas the images of the most basal channels were wider and were centered further rostral than would be predicted from the trend of the other channels. Again, the shape of the guinea pig cochlea and the size of the cochlear implant might lead to more effective stimulation by one or more of the apical return electrodes than by the basal active electrode. The finding of a nonmonotonic map of cochlear place onto cortical place is consistent with the psychophysical finding that human subjects produce nonmonotonic pitch-scaling functions using the CG configuration (Busby et al. 1994).

**Extents of cortical images**

Models of electrical fields in the cochlea predict that various electrode configurations will produce electric fields that vary among configuration in size and in current density, thereby producing activated neural populations that vary in size (Briare and Frijns 2000; Kral et al. 1998; Spelman et al. 1995). In the auditory nerve of the cat, studies have demonstrated that broader stimuli activate a broader distribution of auditory nerve fibers than restricted stimuli (Kral et al. 1998; van den Honert and Stypulkowski 1984, 1987).

In the central auditory system, previous studies that measured directly the spread of neural activation associated with various cochlear electrode configurations have been limited to measurements of neural thresholds. Snyder and colleagues defined spatial tuning curves (STCs) as the plot of multi-unit thresholds for a particular stimulus across recording locations in the central nucleus of the inferior colliculus (ICC) (Snyder et al. 1990). The width of ICC activation was represented by the width of the STC at 6 dB above the minimum threshold. The 6-dB width of the STCs was broader for either the longitudinal bipolar configuration (electrode separation of ≥ 4 mm) or monopolar configuration compared with the radial bipolar configuration (Rebscher et al. 2001; Snyder et al. 1990).

Raggio and Schreiner (1999) measured STCs in the auditory cortex of cats; STCs were plotted in two dimensions, parallel to and perpendicular to the cochleotopic axis of the cortex. Two electrode configurations were compared in cats that were acutely deafened and implanted. The radial bipolar configuration that was tested was presumed to produce extremely focal electrical fields, possibly more focal than our TP configuration. The longitudinal bipolar configuration that was tested was 6 mm in spatial extent, broader than the 3.75-mm extent of our broadest bipolar configuration. The widths of the cortical STC along the cochleotopic axis in the dorsal and ventral low-threshold regions were not significantly different between electrode configurations (Raggio and Schreiner 1999), although the two-dimensional cortical area was greater for longitudinal compared with radial bipolar configurations.

The present results showed a clear increase in the cochleotopic widths of cortical images as electrode configurations were varied from TP to BP + N to MP. At any current level relative to threshold, the image width for the TP configuration was the smallest, whereas the widths for other configurations tended to extend across the entire recording array at current levels more than a few decibels above threshold. Again, the width of BP + N cortical images did not grow consistently with increasing N as one might predict based on electrical field models (Spelman et al. 1995). A possible explanation for that observation is that the proximity to the modiolus of the apical electrode of the bipolar pair might have caused that electrode to dominate the cortical images, thereby reducing the influence of the longitudinal position of the basal electrode. It would be incorrect, however, to regard a bipolar configuration as equivalent to an active apical electrode and a distant diffuse return electrode, like an MP configuration, since the cortical images of MP configurations consistently were larger than the images of even the broadest bipolar configuration.

Cortical images of electrical stimuli consistently were wider than images of acoustical tones measured in normal-hearing guinea pigs (Arenberg et al. 2000). At current levels and sound levels at which images saturated in width, images of TP stimuli were equal or larger in width even than the images of 1-octave acoustical noise bands.

Cortical images of all electrode configurations tended to widen with increasing current level. Basal-channel images appeared to extend beyond the caudal end of the recording array. That limitation of our experimental methods resulted in a tendency of the calculated centroid to shift rostrally with increased stimulus level. Images of apical channels more often were confined within the dimensions of the recording array. Those images tended to widen more caudally than rostrally. One might speculate that a caudalward shift in the distribution
of cortical activity would result in an elevation in the associated pitch. We do not have pitch-perception data in the guinea pig. In humans, increases in current level sometimes have an impact on reported pitch. In a pitch-scaling study by Shannon (1983), the pitch reported by one subject increased systematically as the current level increased. In another study by Townshend and colleagues (1987) that used a forced-choice procedure, one subject consistently associated a current-level increase with a decreasing pitch, and the other subject reported an increasing pitch. The greater tendency of guinea pigs to show a caudalward shift in cortical image and, presumably, an upward shift in the presumed pitch in response to an increase in current level might result from inter-species differences in the spread of current. Difference in current spread, in turn, could result from inter-species differences in the geometry of the basal turn of the cochlea relative to the modiolus (e.g., Schreiner et al. 1997). We believe the present finding to be a consequence of the shape of the guinea pig cochlea that significantly tapers near the location of the most apical electrode (Briaire and Frijns 2000).

Stimuli >1–2 dB above threshold in the MP configuration in the present study tended to activate nearly the entire population of neurons recorded by our multi-channel array, which spanned the present study tended to activate nearly the entire population associated with the basal turn of the cochlea relative to the modiolus (e.g., Chatterjee and colleagues 1987) that used a forced-choice procedure, one subject consistently associated a current-level increase with a decreasing pitch, and the other subject reported an increasing pitch. The greater tendency of guinea pigs to show a caudalward shift in cortical image and, presumably, an upward shift in the presumed pitch in response to an increase in current level might result from inter-species differences in the spread of current. Difference in current spread, in turn, could result from inter-species differences in the geometry of the basal turn of the cochlea relative to the modiolus (e.g., Schreiner and Frijns 2000).

Thresholds and dynamic ranges

Thresholds for cochlear electrical stimulation at the most sensitive cortical sites varied with phase duration, electrode configuration, and basal-to-apical cochlear place. The present results demonstrated that an increase in phase duration produced a decrease in threshold by approximately 4.7 dB per phase-duration doubling, slightly less than the 6 dB per doubling that would be predicted if there were complete temporal integration. That finding is comparable to behavioral threshold changes measured in guinea pigs, 5.3–5.9 dB per doubling (Miller et al. 1995, 2001), in cats, 4.4–5.9 dB per doubling (Smith and Finley 1997), in monkeys, 4.0–5.3 dB per doubling (Pfingst et al. 1991), and human cochlear implant users, approximately 4.2 dB per phase-duration doubling (Moon et al. 1995). When the presumed extent of cochlear stimulation was increased by changing the electrode configuration from TP to BP + N to MP, a systematic decrease in threshold of the most sensitive cortical site was observed. Such a systematic change in threshold as a function of electrode configuration has been described in previous psychophysical studies both in animals and in human cochlear implant users (Chatterjee 1999; Morris and Pfingst 2001; Pfingst et al. 1995). The study by Chatterjee (1999) examined psychophysical thresholds in human cochlear implant users for bipolar configurations having various basal-to-apical electrode extents (i.e., BP + N with various N). Thresholds measured in microamperes decreased exponentially with increasing N, implying a roughly linear decrease in threshold measured in decibels. A similar linear decrease was observed in the present physiological results. In the present study, a systematic increase in the threshold of the most sensitive cortical site was observed as the stimulus channel was changed from apical to basal. Human psychophysical thresholds tend not to show such a consistent dependence of longitudinal electrode position (Pfingst et al. 1997). We believe the present finding to be a consequence of the shape of the guinea pig cochlea that significantly tapers near the location of the most apical electrode (Briaire and Frijns 2000).

The dynamic ranges for cochlear electrical stimuli are markedly narrower than the dynamic range for hearing of sound. For instance, the dynamic range of current levels that human cochlear implant users can hear comfortably is often <10 dB (e.g., Pfingst et al. 1997). The narrow dynamic ranges that are measured perceptually presumably reflect the narrow dynamic ranges of single neurons. The present results showed that the dynamic range at the MSS of each cortical image was only 1–2 dB, regardless of electrode configuration. Those dynamic ranges were considerably smaller than the ranges of approximately 7 dB for single cortical units reported by Raggio and Schreiner (1994). Differences in dynamic ranges across studies are likely a result of differences in the methods. Raggio and Schreiner measured one cortical site at a time, so it is not clear where recording sites were located relative to the MSS. In the present study, we found that dynamic range tended to increase with distance from the MSS. Also, Raggio and Schreiner defined dynamic ranges as ranging from a fairly liberal measure of threshold (2 spikes per 30 trials) to saturation (the transition point of the rate-vs.-level function). This definition was more inclusive than that used in the present study, which defined the dynamic range as the middle 50% of the rate-versus-level function.

The present results show that dynamic ranges at single recording sites tend to increase with increasing cortical distance from the MSS, and that tendency is greater for TP, BP + 0, and CG configurations than for MP. One assumes that a listener’s psychophysical judgment depends on the responses of more than just the neurons at the MSS. That reason, dynamic ranges might be greater for TP, BP + 0, and CG configurations than for MP. That speculation is difficult to evaluate without knowledge of the extent of the cortical area that contributes to listeners’ judgments of stimulus level.

Finally, dynamic ranges might be influenced by the dispersion of cortical thresholds. If the range of cortical thresholds within the cortical image of a particular configuration was wide, neurons would be recruited to activity over a wide range of stimulus levels. That predicts that the more restricted electrode configurations (TP, BP + 0, and CG) would have a greater dynamic range than the MP configuration. Again, we do not know the size of the cortical area that contributes to a psychophysical judgment. Therefore it is not clear to what extent the high-threshold neurons at the edges of the cortical images of highly focal stimuli might contribute to perceived stimulus intensities and to dynamic range.
Concluding remarks

This study characterized the cortical images of single-pulse cochlear electrical stimulation. Cortical images shifted systematically from rostral to caudal in the primary auditory cortex as the place of cochlear stimulation (i.e., cochlear implant channel) was changed from apical to basal. The spatial extent of cortical images depended heavily on the configuration of the stimulating electrodes. The presumed sizes of cochlear electrical fields were reflected in the breadth of cortical images (i.e., diffuse cochlear electrical fields elicited spatially broad cortical images and focal cochlear electrical fields elicited spatially restricted cortical images).

Modern cochlear implants are designed such that each cochlear implant channel is presumed to activate different populations of auditory nerve fibers. Ideally, each cochlear implant channel delivers independent information. A subject’s ability to recognize speech through a cochlear implant is, presumably, limited by the accuracy of cochlear place representation in his or her auditory cortex and by the amount of channel overlap. We found that cortical images of focal electrical configurations, such as TP or BP + O, showed more precise cochleotopic location and less overlap than did images of more diffuse electrical configurations, such as MP. The companion study evaluates quantitatively the accuracy of the cortical representation of cochlear place of stimulation and of electrical current level for various electrode configurations (Middlebrooks and Bierer 2002).

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