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Current focusing and steering:

Modeling, physiology, and psychophysics

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Abstract

Current steering and current focusing are stimulation techniques designed to increase the number of distinct perceptual channels available to cochlear implant (CI) users by adjusting currents applied simultaneously to multiple CI electrodes. Previous studies exploring current steering and current focusing stimulation strategies are reviewed, including results of research using computational models, animal neurophysiology, and human psychophysics. Preliminary results of additional neurophysiological and human psychophysical studies are presented that demonstrate the success of current steering strategies in stimulating auditory nerve regions lying between physical CI electrodes, as well as current focusing strategies that excite regions narrower than those stimulated using monopolar configurations. These results are interpreted in the context of perception and speech reception by CI users. Disparities between results of physiological and psychophysical studies are discussed. The differences in stimulation used for physiological and psychophysical studies are hypothesized to contribute to these disparities. Finally, application of current steering and focusing strategies to other types of auditory prostheses is also discussed.

Keywords

Cochlear implant; Tripolar focusing; Steering; Psychophysics; Physiology; Inferior colliculus

1. Introduction

Over several decades, refinements of cochlear implant (CI) technology and changes in candidacy criteria have created a state in which current CI candidates can expect to understand speech over the telephone after receiving an implant. Despite this high level of overall success, there remains a subset of CI users for whom speech recognition is poor. In addition, speech reception with a cochlear implant is diminished in the presence of background noise and competing acoustic signals, even for users who are able to perform well in the absence of competing sounds. Goals of current CI research are to identify factors that contribute to this decreased performance and to develop technologies and stimulation strategies that can overcome these deficiencies. For example, speech reception for CI users is related to their ability to perceive spectral shapes (Henry and Turner, 2003; Henry et al., 2005; Litvak et al., 2007b; Won et al., 2007). The ability to discriminate spectral shapes across the electrode array might be limited in part by current spread in the cochlea, and in part by the limited number of distinct stimulation sites in the intracochlear arrays used in contemporary implants and

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stimulation strategies. That is, spectral contrast may be limited by the broad current spread or spread of activation and by the limited dynamic range typical of monopolar stimulation. The number of distinct stimulation sites may be limited by the number of electrodes in the intracochlear array. Stimulation methods that can stimulate a larger number of distinct locations in the cochlea, and do so more specifically, should lead to improvement in spectral shape perception, and so result in better speech recognition. Consequently, methods to increase contrast by current focusing and to increase the number of stimulation sites by current steering have become of great interest in CI research.

Current steering and current focusing are each a subset of a broader category of field shaping strategies - the modulation of intracochlear electric fields by simultaneous application of current to multiple CI electrodes. The overall goal of field shaping is to precisely control the region of the auditory nerve (AN) array that will be activated at any given moment. The specific goal of current steering is to stimulate AN regions that are centered between physical CI electrodes, and that of current focusing is to stimulate AN regions that are narrower than those stimulated using traditional monopolar strategies. An example of a simple form of field shaping is used in bipolar and tripolar stimulation strategies.

Among the challenges to field shaping strategies are the increased power consumption resulting from concurrent stimulation of multiple electrodes, the requirement for multiple independently controllable current sources in implanted CI circuitry, and the possibility that shunting of current through the conductive perilymph in the scala tympani (ST) may make it impossible to reach comfort or threshold levels without exceeding either the compliance limit of current source circuitry or the safety limit of intracochlear electrodes. In a research setting, limitations on circuitry and power consumption can be overcome by using percutaneous connectors with multiple external current sources. Some contemporary implantable processors¹ have incorporated multiple independent current sources, and consequently enable use of field shaping strategies employing more than two electrodes. As further advances in electronic hardware increase compliance limits and improve battery capacity in CI processors, field shaping will become a realizable goal.

In this paper, we will review research exploring current steering and current focusing stimulation paradigms. We will begin with an overview of field shaping strategies by describing results from computer and physical models of steered and focused stimulation. Then we will review the results of physiological studies using steered and focused fields. And finally, we will describe the psychophysical and perceptual consequences of steered and focused stimuli in CI users, including their effects on speech reception.

2. Models of field shaping

As mentioned earlier, the goal of field shaping strategies is to accurately and precisely specify the region of AN neurons activated at any moment. Analysis and design of these strategies begins with modeling to determine current levels for intracochlear electrodes that are needed to achieve desired patterns of AN stimulation. This modeling has three steps: the first is determination of the distribution of the potential field within the cochlear tissues, the second is prediction of the location of neurons activated by this potential, and the third is validation of the predictions. After validation, the model can be used to aid in the design of stimulation strategies that will appropriately activate the desired AN region.

¹Namely, CI, CII and HiRes90K implants from Advanced Bionics, and Pulsar100K processor from MedEl. The Pulsar100K processor does not have the capability to stimulate with two sources with differing polarities.

2.1. Modeling methods

The distribution of electrical potentials within the cochlear tissues in response to arbitrary stimulation patterns can be predicted using simple parametric or network models (Jolly et al., 1996; Kral et al., 1998; O'Leary et al., 1985; Rodenhiser and Spelman, 1995; Suesserman and Spelman, 1993; Townshend et al., 1987; Van Compernelle, 1985; van den Honert and Kelsall, 2007) or by finite element modeling (Briaire and Frijns, 2000, 2005b; Frijns, 1995; Frijns et al., 1996; Hanekom, 2001, 2005; Rattay et al., 2001a,b; Whiten, 2007). The simplest models assume that the conductivity of cochlear tissues is uniform, and that current sources (i.e., the stimulating electrodes) are point sources within this homogeneous medium [e.g. (Litvak et al., 2007a)]. In response to stimulation from a single electrode, homogenous models predict isopotential contours that are spherical in shape (Fig. 1A). An example of a simple, homogeneous, finite element model is presented in Fig. 1.

On the left in Fig. 1A, a transparent rendition of the model is illustrated. The diagonal cylinder represents the scala tympani, the seven small spheres represent intrascalar electrode contacts placed in the center of the scala. The black spheres and lines to the lower right of the scala represent spiral ganglion cells and their peripheral and central processes. For the sake of simplicity, the peripheral and distal portions of the central processes are drawn orthogonal to the long axis of the scala and are parallel to the x -axis. The large sphere represents an isopotential contour surface generated by applying current to the active (light colored) contact for the simplest case where electrical resistivity is homogeneous throughout the modeled volume (e.g., the cochlear bone and the perilymph are equally conductive). The parallelogram represents one cross-sectional plane through the scala and spiral ganglion at the level of the activated electrode. In all panels, the level of the contour highlighted is chosen so that decreasing stimulus current by 6 dB would shrink the contour so that it just intersects the peripheral tip of the closest neuronal process; this is designated as the condition for threshold level of neuronal activation. Thus, the contour shown represents a stimulus level 6 dB above this minimum threshold.

On the right in Fig. 1A are several isopotential contours calculated across the cross-sectional plane (parallelogram) shown on the left. The heavy black lined circle represents the isolevel contour shown in the transparent model. Additional potential contours are represented by the light black circles. The potential is indicated by shades of grey. The hatched annulus indicates the cross-section of the cochlear bone. The white circle (located at x & y = 0) represents the activated electrode. The small black circle (with the heavy black line through it) represents a spiral ganglion cell located at this cross-sectional level. The highest potential is located adjacent to the activated electrode, and potentials drop monotonically as distance from the electrode increases.

The second modeling step is determining AN activation patterns that result from the potentials induced by electrical stimulation. The simplest of these models assume that neuronal activation will occur when the intrascalar potential field exceeds a pre-defined threshold, while the most complex use compartmental models of SG neurons which incorporate ionic channels. Compartments are included that correspond to cell soma and spike initiation zones, as well as myelinated and unmyelinated processes. An intermediate approach has been proposed by Rattay (1990), who determined that the "activating function", corresponding to the second derivative of the potential directed along the neuron, is appropriate for characterizing the effective stimulation of neural processes, and has been employed in many models (Bruce et al., 1999a,b,c; Litvak et al., 2007a). However, the first derivative of the potential directed along the neuronal process may be more appropriate for modeling activation either in the vicinity of the soma, which is the presumed site of activation with a cochlear implant (Whiten, 2007), or of peripheral processes (Warman et al., 1992). Examples of isolevel contours of the first

derivative, directed along peripheral processes of model neurons, of the potential field in Fig. 1A are shown in Fig. 1B.

On the left in Fig. 1B, an isolevel contour of the gradient of the potential field in Fig. 1A is shown (gradient taken in the direction of the x -axis). The light-grey “knob” protruding from the cylindrical scala tympani shows the radial and axial extents of the gradient isolevel contour. In this figure for the sake of simplicity, the scala tympani and the isolevel contour are rendered opaque and the spiral ganglion cells are omitted. On the right in Fig. 1B, the model elements are represented as shown A except that the cochlear bone cross-section is omitted and the shades of grey indicate gradient values. The highest gradients (both positive and negative) are adjacent to the activated electrode, and in the direction parallel to the x -axis, the gradients decrease monotonically as distance from the activated contact is increased. The potential gradient was computed along the radial trajectory of the peripheral processes (parallel to the x -axis) by taking centered differences in the x -dimension using the MATLAB (MathWorks, Natick, MA) *gradient* function. Note that these iso-gradient contours are typically sharper than the iso-potential contours.

Although the homogenous model is desirable because it is mathematically tractable, introduction into the model of appropriate inhomogeneities corresponding to different impedances of various cochlear tissues improves accuracy of predictions (Whiten, 2007). For example, introducing a tubular volume of lower conductivity to model (Fig. 1C) the bony wall that separates excitable neural elements of AN neurons from the intrascalar space increases the predicted spread of the potential field along the axis of the scala tympani and has been shown to be essential for accurate prediction of intra-scalar potentials recorded from cochlear implant users (Whiten, 2007).

Effects of increasing the resistivity of the cochlear bone in the example model are illustrated by comparing panels 1A and 1B with panels 1C and 1D, in which the resistivity of the cochlear bone has been increased by a factor of 10. Comparison of the isopotential contour shown in the left of panel 1C with that in panel 1A shows that increasing the cochlear bone resistivity results in a greater spread of the potential in the axial direction. Comparison of the protruding “knob” of the gradient isolevel contour in panel 1D with that in 1B shows a similar increase along the axial dimension.

The cross-section showing isopotential contours on the right side of 1C demonstrates that the contours become “bunched” within the volume of the cochlear bone when the bone resistivity is 10 times larger than the resistivity of other fluids and tissues. This is reflected in the large gradients observed within the cochlear bone in the cross-section shown in panel 1D. The non-monotonic change in potential gradient along the x -axis produces a local maximum of potential gradient within the volume of the cochlear bone. This suggests that response thresholds would be lower for neurons with peripheral processes traversing this bony wall. Further increasing the bone resistivity to 100 times that of perilymph (Whiten, 2007) would further increase axial spread of the potential and also the gradient within the bone.

Addition of greater detail in cochlear tissue impedances and geometry, as well as impedances and geometry of the CI array, generates models of greater complexity, and presumably greater predictive accuracy. It is important to note that while the results of modeling studies are interesting and can provide valuable insight, validation of models, especially of the more complex models, remains an important step.

2.2. Full-array field shaping paradigms

In theory, the goal of the modeling effort as applied to field shaping has been to determine a stimulation configuration which will produce an arbitrary desired pattern of activation in the

auditory nerve. Potentially, such a configuration could involve stimulation on all intra-cochlear electrodes. However, given the complexity of the neural activation models described above, and lack of knowledge as to the cochlea of individual subjects, such use of the computational models has been prohibitive. Consequently, various methods have been proposed for determining the optimal configuration for an individual subject.

One interesting approach was proposed by White and colleagues (Van Compernelle, 1985; Townshend and White, 1987). In this approach, psychophysical measurement of threshold shift is used to determine the degree of interaction between adjacent electrodes. Once this is determined, it is possible to determine the compensation currents necessary to cancel the interaction. Although in theory such stimulation should result in greater channel independence, compensation currents computed this way were unfortunately too great to be realizable in devices available at that time (Van Compernelle, 1985).

Several investigators have proposed a method of shaping the intra-scalar potential profile based upon measurements of potentials made using intracochlear electrodes themselves. In this technique, potentials are recorded at one electrode while current is applied on each of the other electrodes. If every combination of stimulation and recording electrode is used, it is possible to invert the resulting matrix of transimpedance measurements to determine currents that result in a potential profile that is identically zero everywhere except at the stimulation electrode (Townshend and White, 1987; Rodenhiser and Spelman, 1995; Hartmann and Klinke, 1990; Ross, 2006; van den Honert and Kelsall, 2007). Theoretical support for such field shaping strategies is predicated upon linear summation of field potentials, i.e., that the field resulting from simultaneous stimulation of two or more electrodes is simply the sum of the fields created when each electrode is stimulated individually. Accuracy of transimpedance measurements or estimates is critical to achieve accurate predictions (Van Compernelle, 1985), and can be complicated by a number of factors including residual polarization of the stimulated electrode. Recently van den Honert and Kelsall (2007) have described a method of estimating transimpedance elements that reduces the influence of residual polarization and appears to work well in practice. Empirical measurements of potential field spread in human subjects can be made only using intrascalar electrodes, and implementation of stimulation strategies are subject to safety and CI current and compliance voltage limitations. In animal studies, potential fields can be measured using additional electrodes that are introduced into the scala tympani or the modiolus, as well as on CI electrodes, and hardware limitations are less restrictive.

2.3. Simplified field shaping paradigms

Although full-array field shaping paradigms promise to be useful, they have a number of drawbacks. First, these paradigms require independent control of a large number of electrodes, which could result in increased complexity of implant electronics, and put increased demands on the rates of the data transfer between the external processor and the implant. Second, the full-array field shaping configuration that results in minimum potential spread within the scala tympani does not necessarily result in minimal spread of activation of auditory nerve fibers.

Vanpoucke et al. (2004) showed that the intra-cochlear spread observed in human subjects can be very closely approximated by a “leaky tube” model. If this is the case, then current spread in the cochlea can be limited by using only electrodes adjacent to the center electrode. In this paper, we will therefore focus on configurations that use four or fewer intracochlear electrodes, in combination with external ground (Fig. 3). We will refer to configurations using three electrodes as partial tripolar configurations. In general, we define a partial tripolar configuration as one in which current I is placed on the center electrode, while currents $\sigma(1 - \alpha)I$ and $\sigma\alpha I$ are simultaneously applied in opposite polarity to the apical and basal electrodes respectively. The coefficient σ is referred to as the *tripolar compensation coefficient* [as in Litvak et al. (2007a)] and controls the “degree” of focusing, while α is the *steering*

coefficient. The configuration of $\sigma = 1$, $\alpha = 0.5$ will be referred to as the *full tripolar configuration*. Note that the term “quadrupolar” has also been used to describe the full tripolar configuration, while the term “tripolar” has also been used to describe the partial tripolar configuration (e.g. Jolly et al., 1996).

Fig. 2 shows the predictions of the potential and gradient amplitudes in the vicinity of the nerve fibers in response to various partial tripolar stimulation configurations. Potentials and gradients were computed using the finite element model described in Fig. 1. Stimulus current levels are shown in the left column: I_n indicate currents on intracochlear electrodes 2-4, and I_X indicates current on the extracochlear (return) electrode. The gradients (right column) are of particular interest because they are believed to describe effective stimuli to the neural tissue (Whiten, 2007). Monopolar configuration (panel A) corresponds to the tripolar compensation coefficient of $\sigma = 0$. In this configuration, applying stimulus current equal to twice the minimum threshold current results in potential or gradient levels that exceed the threshold potential (θ) over a broad region of the simulated cochlea. For example, in the panel showing the gradient strength for monopolar stimulation, neurons that lie within the region where the strength curve lies to the right of θ would experience a supra-threshold gradient, and therefore be activated.

Neural spread can be decreased if tripolar compensation coefficient σ is increased to 0.5 (panel B). However, the current level needed to reach threshold must be increased (in this case, by a factor of 1.3). Greater narrowing in activation will result if compensation coefficient σ is increased to 1 (full tripolar configuration, panel C). However, with this configuration, one also sees the presence of negative side-lobes in the gradient function. Since cochlear implant systems typically use stimulation pulses which have both cathodic and anodic phases of equal amplitude, these side lobes can be of perceptual significance (Litvak et al., 2007).

By adjusting the steering coefficient α , it is possible to “steer” the peak of the gradient function (e.g. $\alpha = 0.75$, $\sigma = 1$, panel D). Presumably, such “steering” could be used to provide increased spectral information to the CI user.

Two other field shaping configurations will be considered in this paper. The first is the bipolar configuration (panel E). This configuration is predicted to be less focused than the tripolar and some partial tripolar configurations, and also has a large side lobe in the vicinity of the compensation electrode. However, the bipolar configuration is of interest because it is readily available in most common cochlear implant systems, and has been studied extensively. Note that in our model, if the side lobes are considered, the bipolar configuration does *not* result in narrower excitation patterns compared to monopolar stimulation (panel A); however, other models do indicate an advantage for bipolar stimulation (e.g. Whiten, 2007). Finally, a monopolar steering configuration (panel F) allows for shifting of the peak of excitation, but does not attempt to sharpen the excitation pattern. Because this configuration has maximum current requirement that is similar to the monopolar, it can be readily implemented using contemporary cochlear implants.

One question that has been of theoretical interest with regards to field shaping has been whether configurations that produce more constrained fields will also produce narrower neuronal activation patterns once the current has been adjusted to be of equal loudness (Pfingst et al., 1997). In principle, the answer to this question will require knowledge of what aspects of neural excitation account for loudness. Litvak and colleagues (2007a) used the assumption that loudness is related to the total number of spikes on all active neurons. They showed that if one assumes the intrasite variance in single-fiber thresholds of humans to be similar to that recorded in cats (van den Honert and Stypulkowski, 1987), the neural activation patterns resulting from constrained configurations should be somewhat narrower than monopolar stimulation, so long as the side-lobes associated with the focused configuration remain below neural thresholds.

In summary, theoretical and physical modeling studies indicate that field shaping strategies should be effective in stimulating narrow regions of the auditory nerve, and should also be effective in stimulating regions that are not “centered” on intracochlear electrodes. To determine whether the responses to stimuli created by current steering and shaping strategies differ from one another, we turn to studies of central auditory system responses and studies of perceptual differences. Recent advances in these areas are described in the following sections.

3. Physiology

3.1. Physiological studies of current focusing and current steering

Several physiological studies of field shaping have been conducted with the specific goal of determining the tonotopic location and distribution of auditory nerve fibers activated by different intracochlear electrical stimulation strategies. Most of these studies have concentrated on current focusing; only a few have examined the effects current steering (Bonham et al., 2006; Miyoshi et al., 1997, 1999, 1996).

Moreover, most focusing studies have limited themselves to comparing monopolar (MP), bipolar (BP) and tripolar (TP) stimulation using symmetric biphasic pulses; very few have examined the effects of other stimulation strategies such as partial tripoles (Kral et al., 1998). Nevertheless, the distribution of cochlear activity evoked by MP, BP and TP stimulation has been studied using a variety of methods including directly measuring activity evoked in the auditory periphery as well as measuring responses in the central nervous system.

Studies using direct measurement of the spread of evoked activity in the auditory nerve have employed several techniques. The most direct are those that have measured responses of single fibers distributed across the AN array after stimulation at single intracochlear sites (van den Honert and Stypulkowski, 1987), and, conversely, measurement of a single fiber's response to stimulation at a series of sites distributed along the scala tympani (Hartmann and Klinke, 1990; Kral et al., 1998; Liang et al., 1999). The results of these studies are consistent and can be easily summarized. Generally, monopolar stimulation evoked activity across a broad (some times very broad) cochlear region, however, the current required to achieve activation was low. In contrast, bipolar stimulation evoked activity across a narrower region, but required use of higher current levels. Finally, tripolar stimulation evoked activity with the narrowest distribution, but required the largest current.

Studies of current focusing using less direct measures of cochlear activity spread include measurement of the amplitude of the electrically evoked compound action potentials (ECAPs) as a function recording electrode distance from the stimulating electrode (e.g., Cohen et al., 2004; Dillier et al., 2002; Finley et al., 1997; Frijns et al., 2002; Miyoshi et al., 1997, 1999, 1996). Activity spread has also been estimated using forward masking procedures of ECAPs by measuring the ECAP amplitude to a probe signal on one electrode while varying the separation between it and a masker on another electrode or pair of electrodes (e.g., Abbas et al., 2004; Brown et al., 1996; Cohen et al., 2004, 2006; Miller et al., 2003; White et al., 1984). The vast majority of ECAP studies have been conducted in humans, but this technique is extremely attractive, since in principle it can be used in both human and animal subjects to provide a bridge between these two bodies of data (Miller et al., 2004). Most of the ECAP studies (including neural response telemetry, NRT, and neural response imaging, NRI) that are directed at comparing electrode configurations have compared stimulation in monopolar mode vs. bipolar stimulation at different electrode sites. The results of these studies suggest that the centroid of the activity moves with the site of stimulation, a result that is consistent with the shift in pitch percepts with stimulation site in CI users. They also indicate that monopolar stimulation evokes more widely distributed activity than bipolar stimulation. However, in some studies, the results indicate that ECAPs demonstrate broader excitation patterns than suggested

by psychophysical forward masking patterns (Cohen et al., 2004), whereas other studies suggest that interpretation of intracochlearly recorded ECAPs may not reflect the auditory nerve compound action potential (Briaire and Frijns, 2005a).

Studies of the effects of current focusing in the CNS rely on measurement of activity spread across tonotopically organized areas in the central auditory system. These studies include 2-deoxyglucose labeling and fos-like immunoreactivity (Brown and Benson, 1992; Ryan et al., 1990; Saito et al., 1999), measurement of slow-wave evoked potentials (Hartmann et al., 1997; Klinke et al., 1999; Kral et al., 2000), and measurement of activity in neurons or small groups of neurons using microelectrodes in the inferior colliculus (e.g., Black and Clark, 1980; Bonham et al., 2007, 2006; Merzenich et al., 1977; Rebscher et al., 2001; Snyder et al., 2004, 2007, 1990) and in the auditory cortex (e.g., Bierer and Middlebrooks, 2002; Middlebrooks and Bierer, 2002; Raggio and Schreiner, 1999). Most of these studies infer the tonotopic organization of the central areas from previous acoustic studies in normal hearing animals, whereas others use either acoustic stimulation (Black and Clark, 1980; Merzenich et al., 1977) of the contralateral cochlea or acoustic mapping prior to deafening and electrical stimulation (Snyder et al., 2004, 2007; Middlebrooks and Snyder, 2007). The results of these current focusing experiments in the central auditory system confirm those in the periphery. Monopolar stimulation evokes broadly distributed activity and has a low threshold, bipolar stimulation evokes more narrowly distributed activity but has a higher threshold and tripolar stimulation evokes the narrowest activity but has the highest threshold.

As opposed to current focusing there are very few physiological studies of current steering. Miyoshi et al. (1997, 1999, 1996) used compound action potentials recorded using intracochlear electrodes to estimate the distribution of activity evoked by currents steered to one or another adjacent electrode. In the studies described below we will briefly summarize the of results of Bonham and colleagues (2007, 2006) in which responses to stimulation using current steering and focusing configurations were recorded in the inferior colliculus.

3.2. Current steering and current focusing measured in the inferior colliculus (IC)

3.2.1. Methods—The experimental methods have been described in detail elsewhere (Snyder et al., 2007). Briefly, the right inferior colliculus of anesthetized guinea pigs was exposed by aspiration of the overlying cortex. A multichannel silicon recording probe was inserted using a micromanipulator along a trajectory parallel to the tonotopic axis of the central nucleus of the inferior colliculus (ICC) so that the best frequencies of the 16 recording sites spanned an appropriate frequency range. The recording probe was fixed in place using agar and dental acrylic. The head was rotated and the left bulla was opened and the round window visualized. The cochlea was deafened with an intrasclerous injection of neomycin sulfate, and a stimulating electrode custom-designed for the guinea pig (Rebscher et al., 2007) was inserted into the scala tympani through the round window. A silver wire was placed in the skin near the left ear canal and served as the extracochlear return electrode. Multi-unit neuronal activity was recorded from the IC while the cochlea was electrically stimulated using an isolated eight-channel current stimulator. All procedures were approved by the UCSF Institutional Animal Care and Use Committee.

Stimuli were isolated charge-balanced current pulses with two phases. The initial cathodic phase (100 μ s) of each pulse was presented on one (for current focusing experiments) or two (for current steering experiments) predefined source electrode(s). The return path for the current was determined by the experimental goal (current focusing or steering) and the tripolar compensation coefficient, σ , and included any or all of the following: the extracochlear silver wire and the two intracochlear electrodes adjacent to the source electrode(s). The second, charge-recovery, phase of each stimulus pulse was a factor of 10 longer in duration and a factor of 10 lower in amplitude (i.e., stimulus pulses were pseudomonophasic) to decrease the

likelihood of cathodic stimulation of the auditory nerve array near sites corresponding to the intracochlear return electrodes during the recovery phase. For focusing experiments, pulses were presented using values of σ ranging between 0 and 1 and stimulus levels covering a range from below minimum threshold to above threshold. For steering experiments, pulses were applied simultaneously to two adjacent electrodes; the sum of the currents applied to these two electrodes varied from below threshold to above threshold, and the proportion of the total current applied to each electrode (corresponding to the parameter α) varied from 0 to 1; adjacent intracochlear electrodes provided the return path for these currents.

3.2.2. Results—A comprehensive evaluation of our observations using current steering and current focusing stimulation will be reported in a subsequent paper. This paper will present observations made from one representative current steering experiment and from one representative current focusing experiment.

3.2.2.1. Current steering: Spatial response profiles (SRPs) recorded in the inferior colliculus during one experiment using current steering stimulation are shown in Fig. 3. Each panel shows the SRP for a different total cathodic current. In each panel, the proportion of the cathodic current applied to intracochlear electrode #4 is shown along the abscissa; the balance of the cathodic current is applied to electrode #5. The return path for the stimulating current was divided between two adjacent intracochlear electrodes (#3 and #6). In each panel the ordinate indicates the depth in the IC (corresponding to the axis of characteristic frequency), and the response strength is indicated by color. Note that each column of each panel corresponds to a different unique combination of stimulus currents that was applied to electrodes #4 and #5. In each column, the location of the activity peak (asterisk) was determined using spline interpolation. The interpolation procedure identified the peak of the evoked response when the stimulus was suprathreshold, or identified the site with the greatest spontaneous activity when the stimulus was subthreshold.

At the lowest total stimulus current level (0.036 mA, Fig. 3, upper left panel), when 100% of the current was applied to electrode #5 (and 0% was applied to #4; $\alpha = 0$) a focus of response was observed at a depth of approximately 950 μm , which is interpolated to have a CF of 18 kHz in this experiment. When 100% of the current was applied to electrode #4 ($\alpha = 1.0$), the focus of activity was observed to be 750 μm , corresponding to 15 kHz. When the current was equally between these two electrodes (50% to each electrode; $\alpha = 0.5$), the stimulus was below the minimum threshold at all locations.

As the total stimulus current was increased (successive panels in Fig. 3), the range of α 's for which no response was observed decreased. When the total current exceeded 0.048 mA, the response was present for all values of α , although the total activity was always smallest for $\alpha = 0.5$. For current levels above 0.066 mA, the location of the peak response was observed to gradually shift from 950 to 750 μm as a greater proportion of the total current was applied to electrode #4. This gradual shift of the activity peak corresponds to a gradual steering of the stimulating field between adjacent physical electrodes within the cochlea. Thus, the center of the population of neurons that was activated when the stimulating current was *divided* between physical electrodes #4 and #5 lay between the centers of neuron populations activated when all the stimulating current was delivered to either electrode #4 or #5. Further, the center of activity appeared to shift gradually as the proportion of total current delivered to either electrode changed.

These observed differences in regions of activated neurons would create corresponding perceptual changes in perceived pitch in human listeners. Thus, current steering strategies might provide CI users with a more extensive palette of perceptual pitches that might approach the continuum of pitch available to normal hearing listeners.

3.2.2.2. Current focusing using partial tripolar compensation: Spatial tuning curves (STCs) recorded in the inferior colliculus during one experiment using partial tripolar compensation are shown in Fig. 4. In each panel of the figure, stimulus current is plotted along the abscissa, IC depth (corresponding to the axis of characteristic frequency) is plotted along the ordinate and the strength of response is encoded by color.

At the lowest stimulus level in each panel, there is no activity at any IC depth, indicating that the stimulus is below the minimum threshold. As the stimulus level is increased, response first becomes evident at a depth of approximately 400 μm , corresponding to a CF of approximately 9 kHz in this animal. In each panel, as the stimulus level is further increased, the response measured at 400 μm becomes stronger, and the responsive region grows and spreads along the tonotopic axis of the IC. This increase is qualitatively similar to one observed in normal-hearing animals in response to an acoustic tone of increasing level (Snyder et al., 2007).

Comparing responses between panels (a) and (i), it can be seen that the maximum spread of response along the tonotopic axis is substantially larger when the stimulus is in the monopolar configuration ($\sigma = 0$) than when it is in the full tripolar configuration ($\sigma = 1$). It can also be seen that the range of stimulus level between the minimum threshold (indicated by red circles) near 400 μm and distant recording sites (e.g., at 1000 μm) decreases substantially as σ is reduced. Finally, for a given stimulus level above the minimum threshold, the spread of the response along the tonotopic axis increases as the σ is decreased (indicated by red lines). These qualitative phenomena are general features of our IC current focusing experiments. Two specific factors that differ among our focusing experiments are the rate of threshold decrease and rate of width increase as functions of the parameter σ . In this example, the minimum threshold gradually decreased from 100 to 20 μA as the stimulus configuration was changed from tripolar to monopolar. Also in this example, the spread of the response measured at 2 dB above the minimum threshold increased gradually from 300 to about 500 μm , and then increased rapidly to greater than 1500 μm . A quantitative summary of variations in these measurements across experiments is beyond the scope of this short review.

In summary, responses measured in the inferior colliculus to stimulation using current steering and focusing configurations are consistent with predictions made by models. Dividing stimulus current between two nearby intracochlear electrodes can evoke responses along the IC tonotopic axis that lie between response evoked by stimulating either electrode alone. Reducing the spread of the electric potential (and potential gradient) by full or partial tripolar compensation can reduce the spread of responses evoked along the IC tonotopic axis.

4. Psychophysics

4.1. Evidence of focusing with bipolar and tripolar stimulations

In experimental animal models, stimulation using “focused” configurations appears to reliably produce neural activation patterns that are more specific than those produced by monopolar stimulation. It is less clear, however, whether such stimulation also produces narrower excitation patterns in human cochlear implant users. Although animal and human experiments have been reported to be contradictory (e.g. Chatterjee et al., 2006), care must be taken in comparing the two. In fact, the degree of mismatch appears to be less than has previously been reported.

Several techniques have been used for estimating excitation patterns in human listeners. One such technique is forward masking (e.g. Houtgast, 1972; Moore, 1978). In this technique, thresholds to narrow-band, spatially constrained probe stimuli are measured in isolation, and also when they are preceded by a masker. A common interpretation of forward masking patterns is that the probe threshold is elevated when the masker excites part of the neural population

excited by the probe, but is unchanged if the neural populations excited by the masker and the probe do not overlap. Thus, probe threshold as a function of masker-to-probe spectral separation is considered to be indicative of the degree of neural population overlap between the stimuli. In normal hearing listeners, acoustic forward masking produces psychophysical tuning curves that are comparable to physiological tuning curves in the auditory nerve; however, the psychophysical methods generally overestimate the sharpness of physiological tuning curves (Moore, 1978; Nelson and Freyman, 1984; Relkin and Turner, 1988; Turner et al., 1994). Forward masking procedures have an advantage over simultaneous masking procedures in that the former are not confounded by lateral suppression (Houtgast, 1972; Moore, 1978). Nevertheless, forward masking measurements in CI users can be expected to produce tuning curves that are similarly comparable to electrical tuning curves measured in the auditory nerve.

In cochlear implant listeners forward masking techniques have been applied to estimate activation spread of electrical stimuli. For both monopolar and bipolar configurations, greatest elevations in probe threshold generally occur near the masker electrode. When normalized relative to the peak, the overall masking patterns are largely independent of the level of the masker. This result is in variance with physiological studies, which show an increase in excitation width with increasing stimulation level. However these psychophysical studies also reported large variability across subjects. Some subjects exhibited forward masking patterns with a pronounced peak in a 1-2 mm region around the probe, while others had masking excitation patterns that had a broad peak extending over 8 mm or more (Boex et al., 2003; Chatterjee and Shannon, 1998, 2006; Kwon and van den Honert, 2006a).

Several studies have directly compared monopolar and bipolar forward masking patterns using various bipolar configurations, which are described as $BP + n$, where n is the number of electrodes between the electrode pair. However, the narrowest bipolar configurations ($BP + 0$) could not be evaluated in many cases due to their high thresholds and compliance limits in commercial devices. For example, no study has reported on forward masking patterns in response to electrodes that were more narrowly spaced than 1 mm. This is unfortunate, as the narrowest configurations are likely to be the ones which are expected to produce the most spatially restricted patterns.

In general, bipolar maskers have not demonstrated narrower forward masking patterns than monopolar maskers (Boex et al., 2003; Kwon and van den Honert, 2006a; Shannon, 1983). Kwon and van den Honert (2006a) directly compared bipolar ($BP + 1$ and $BP + 2$) and monopolar configurations in four subjects. While in some subjects, a narrower peak was observed in the vicinity of the masker electrode pair using bipolar configurations [as in Shannon (1983)], in others the monopolar masker was narrower. In addition, monopolar and bipolar masking patterns frequently exhibited multiple peaks, with multiple peaks occurring more frequently using bipolar configurations. Distant bipolar (e.g. $BP + 7$) configurations resulted in masking patterns that were flatter, and frequently had two peaks (Chatterjee et al., 2006; Lim et al., 1989). A similar result was observed in the spatial tuning curves in inferior colliculus when bipolar electrodes were widely separated (Snyder et al., 2007).

One issue with several studies that compared monopolar and bipolar masking patterns has been the use of different probe stimuli when comparing across differing stimulus configurations. In particular, a monopolar probe was used when evaluating a monopolar masker, while a bipolar probe was used when evaluating a bipolar masker. Such comparisons may be inappropriate because the two probes may have different excitation growth patterns such that a threshold shift reported for monopolar probe may not be equivalent to a threshold shift reported for a bipolar probe. In addition, a broad probe may provide opportunities for “off-frequency” listening whereby the presence of the probe becomes detected by the subject listening to the

place in the cochlea that is some distance away from the probe electrode. Off-frequency listening can artificially sharpen the forward masking patterns when the probe with the broader configuration is used (O'Loughlin and Moore, 1981). One way to get around this problem is to use the same “narrow” probe stimulus regardless of the masker stimulus.

In summary, the psychophysical data have not indicated a consistent narrowing of the forward masking patterns in bipolar mode as compared to monopolar mode. However, more definitive experiments have yet to be performed.

Although bipolar configurations produce excitation patterns that are not consistently narrower than monopolar patterns, it is unclear whether the same is true for the tripolar and partially tripolar configurations. The physiological measurements in the guinea pig indicate that patterns of activation produced using tripolar and partial tripolar configurations can be narrower than those produced using either monopolar or bipolar configurations [Snyder et al. (2004, 1990) - IC; Bonham et al. (2005) - IC; Bierer and Middlebrooks (2004) - cortex]. Tripolar and partial tripolar configurations have been evaluated in human cochlear implant recipients in a number of recent reports (Bierer, 2007), including work in our laboratory. Saoji and Litvak (2007) evaluated masking patterns associated with monopolar and partial tripolar maskers. The masker stimuli were 350 ms pulse trains with 129 μ s phase durations and a stimulation rate of 1923 Hz. For the partial tripolar masker, the tripolar compensation factor σ was varied from 0.625 to 0.875. The probe was a 20 ms pulse train with the same phase duration, pulse rate and tripolar compensation factor as the masker. Prior to estimation of the masking patterns, the monopolar and the tripolar maskers were matched in loudness using the method of adjustments. In the masked condition, the probe was presented 4 ms following the masker stimulus. All research was approved by an independent private institutional review board.

A more detailed report of these experiments will be described in a subsequent publication. Fig. 5 shows the average difference in threshold between the monopolar and the tripolar masker (in μ A) for tripolar compensation factors of 0.625, 0.75 and 0.875. Although these results are preliminary, they suggest a sharpening of the excitation pattern near the masker electrode. Consistent with spectral sharpening, 3 out of 5 subjects reported that the focused stimuli produced tones that had greater pitch strength as compared to those produced with monopolar stimulation (Marzalek et al., 2007). However, some subjects also reported that the partial tripolar stimuli were significantly higher in pitch (Marzalek et al., 2007; Mens and Berenstein, 2005). It is unclear however whether the “pitch salience” and “pitch magnitude” are partially confounded in CI listeners. Since five of six subjects were able to achieve comfort levels with tripolar compensation factor of 0.75 (Litvak et al., 2007a), it appears that field focusing with partial tripolar compensation may be a practical possibility for a majority of cochlear implant recipients.

In summary, the preliminary data for the partial tripolar configuration suggest a sharpening of the forward masking pattern near the center electrode of the tripole.

4.2. Dynamic range of monopolar and “focused” configurations

Recordings from central auditory systems of experimental animal suggest that relative to the monopolar configuration, focused configurations such as bipolar or partial tripolar stay spatially constrained over a greater range of levels (e.g. Fig. 4). If a similar phenomenon occurs in human cochlear implant listeners, then it is likely that more focused configuration should result in a greater psychophysical dynamic range relative to monopolar. However, the evidence for such increase of dynamic range with focusing is weak. Bierer (2007) showed dynamic range that was comparable for monopolar, bipolar and tripolar configurations for several subjects; however, dynamic range for the narrow bipolar configuration used in that study could not be assessed due to compliance limitations. Mens and Berenstein (2005) reported a significant

increase in dynamic range for a range of tripolar configurations; however, the compliance limits were not carefully controlled in this study. Litvak et al. (2007a) reported dynamic ranges for several partial tripolar configurations. For most patients, dynamic range was approximately constant as a function of the tripolar compensation coefficient σ . However, two of the six subjects showed significantly increased dynamic range for more focal configurations (larger σ). Interestingly, both of these subjects also had a silastic positioner attached to their electrode arrays and had the smallest interaction between electrodes. Thus, the effect of configuration on dynamic range may partially depend on the proximity of the electrode array to target neurons.

4.3. Speech perception with “focused” configurations

Speech perception of cochlear implant recipients is strongly correlated with their ability to resolve spectral peaks (Henry and Turner, 2003, 2005; Litvak et al., 2007b; Won et al., 2007). Spectral resolution of cochlear implant listeners may be limited by broad excitation patterns of electric stimulation. Conceptually, and in simulations, narrower excitation patterns lead to better spectral resolution and correspondingly better speech understanding. It is natural to hypothesize, therefore, that narrower activation patterns will result in improved speech perception in CI users.

Speech perception with programs that use bipolar stimulation is generally poorer than or the same as that with programs that use monopolar stimulation [see Pfingst et al. (2001) for a recent review]. This finding is consistent with the observation that the bipolar configurations do not generally result in narrower forward masking patterns. In addition, “bipolar” programs have the drawback that they require larger current levels that result in decreased battery life, as well as increased fitting times due to greater variability in thresholds and upper comfort levels. Furthermore, in order to achieve sufficient loudness, bipolar programs use pulses of longer duration as compared with monopolar programs, and in the context of multi-electrode stimulation, require lower stimulation rates. Perhaps for these reasons, monopolar programs are preferred by CI users.

Mens and Berenstein (2005) evaluated speech perception with two partially tripolar configurations. Consistent with results of forward masking studies, they found that speech performance using a partial-tripolar configuration with a tripolar compensation coefficient σ of 0.5 was equivalent to that using monopolar stimulation. Generally they found that more focused configurations produced a decrement in speech perception. However, several subjects in that study reported a substantial increment in the pitch with tripolar. In some cases, pitch shifts were equivalent to more than half of the total pitch range produced by the array. Such profound pitch shifts may have an adverse affect on speech perception. Certainly, greater time may be necessary for subjects to adjust to such profound changes in frequency-to-place alignment. Further, novel strategies may need to be developed that take advantage of focused stimulation.

4.4. Evidence of current steering in human subjects

When two electrodes are stimulated simultaneously, subjects frequently report hearing a pitch that is intermediate to the pitches resulting from stimulation using each electrode separately (Townshend et al., 1987; Wilson et al., 1993). Due to broad spread of currents through the scala tympani, it is possible to use electrodes separated by a distance of as much as 16 mm to evoke such intermediate pitches (Townshend et al., 1987). The resolution, identified as the number of intermediate pitches, achievable with simultaneous stimulation has been quantified in several studies (Donaldson et al., 2005; Firszt et al., 2007; Koch et al., 2007). Generally, it is found that on average 4-7 intermediate pitch sensations per pair of adjacent electrodes

(spaced between 0.85 and 1.1 mm, depending on electrode type) could be perceived using current steering, with a trend for basal pairs to produce fewer intermediate pitches.

It has long been speculated (Townshend et al., 1987) that such intermediate pitches can be used to convey additional spectral information to cochlear implant listeners. Wilson and colleagues evaluated a speech processing strategy (VCIS) where additional virtual channels were implemented using simultaneous stimulation of adjacent channels (Wilson et al., 1994a,b). Speech perception was tested without chronic experience with the VCIS processor. In that condition, they reported no statistical improvements in speech perception with the VCIS processor over the CIS processor. However, all three subjects preferred the 11-channel VCIS processor over the 6-channel CIS processor, reporting that VCIS processor was “richer” sounding (especially while listening to music). It is possible that more experience with the new processor would have improved their ability to use the added cues and would have improved speech perception.

Recently, Trautwein (2006) investigated performance with a speech processing strategy (Fidelity120) in which for each processing channel, one of nine virtual channels in between adjacent electrode pair was stimulated in each stimulation cycle. The stimulated virtual channel was chosen based on fine spectral analysis of the incoming sound, and the total amplitude of stimulation was determined by the total energy within the channel. Rather than *increasing* the number of channels, this approach sought to better represent within-channel fine structure. In studies with normal hearing listeners, such representation of sound has been shown to support good performance in noisy environments or in the presence of competing talkers (Nie et al., 2005; Stickney et al., 2005). In addition, for some subjects, such intermediate channels appeared to be perceived even in presence of “noise stimulation” on other channels (Kohler et al., submitted for publication). In our investigation, 34 subjects were converted to the new strategy for three months, after which their speech performance was evaluated and compared with base-line performance. In addition, a questionnaire was administered where subjects reported on the quality of the new strategy. As with VCIS, significant differences in reported quality were recorded, with 28 of 34 subjects strongly preferring the new strategy (Trautwein, 2007). While subject performance was generally improved, the overall performance gains were small. The largest improvement occurred for a speech-in-noise test (Az-Bio sentences +8 SNR; 60% versus 67% correct, $p < 0.05$), and the improvements were largest for the poorest performers. Furthermore, it cannot be excluded that the improvement may be related to other differences in processing between the baseline strategy and the new strategy. For example, the new strategy uses a high-resolution FFT-based frequency decomposition which results in small filter overlaps across channels. Such high-resolution analysis is important to minimize “incidental” across-channel “pitch steering” associated with non-simultaneous stimulation which occurs in CIS-like strategies (Kwon and van den Honert, 2006b; McDermott and McKay, 1994). In contrast, the baseline strategy has filter bands with greater overlaps. Although further evaluations of the new strategy are on-going, both the baseline strategy and the VCIS strategy may be limited by the broad current spread associated with monopolar stimulation. A combination of focused stimulation with steering may be necessary to further improve place perception of cochlear implant listeners.

5. Discussion and conclusion

5.1. Physiology and psychophysics of monopolar stimulation

One surprising result from studies of single- and multi-unit responses recorded from central structures in experimental animals has been the extremely broad excitation patterns in response to monopolar stimulation [e.g. Snyder et al. (2007)]. When specificity is observed with monopolar stimulation (e.g. Fig. 4, $\sigma = 0$), it is usually maintained only as long as the amplitude of stimulation is less than 3-4 dB above threshold of the most sensitive fibers. However, for

most human listeners, excitation patterns derived from forward masking do have a pronounced peak near the stimulation electrode even for relatively loud masker stimuli (Chatterjee and Shannon, 1998; Kwon and van den Honert, 2006a). In addition, extremely broad spread with monopoles is not consistent with successful use of monopolar stimulation in speech processing strategies (Pfungst et al., 2001).

While differences in species and methods must be considered (see above), one possible hypothesis is that differences in excitation patterns with monopoles seen in physiological and psychophysical experiments are due to the difference in stimuli used in physiological experiments (namely, low-rate pulse trains or single-pulses) and those used in forward masking experiments and speech processors of human subjects (typically high-rate (>200 Hz) pulse trains). For human subjects, thresholds to 20 Hz pulse trains are on average 7 dB higher than thresholds to 1000 Hz pulse trains (unpublished observations). Thus, given the relatively narrow range of electric stimulation, each pulse of a high-rate pulse train that is reported as loud by cochlear implant recipient may be only a few dB above single-pulse threshold. This hypothesis would predict that in animals, monopolar stimulation with higher-rate pulse trains may evoke patterns that are selective over a larger dynamic range. Conversely, for human listeners, forward masking patterns for single pulse maskers are predicted to be wider than those for comparably loud high-rate pulse trains.

5.2. Next steps

Field shaping techniques (which include “current steering” and “current focusing”) provide opportunity to alter the excitation patterns in response to electric stimulation. In this manuscript, we presented several lines of convergent evidence, obtained from computational models, animal studies and human subjects, that field shaping achieves some of its goals. First, tripolar and partial tripolar stimulation (at least with tripolar compensation coefficient $\sigma > 0.6$) results in excitation patterns that are narrower than comparable monopolar stimulation. Second, by manipulation of two simultaneously presented pulses, it is possible to systematically steer the peak of excitation patterns to a locations that are intermediate to those of the component electrodes.

Although these effects have been demonstrated on the psychophysical level, the challenge remains to properly use the focusing and steering techniques in a sound processing strategy. One processing strategy, using dynamic monopolar steering, has provided modest improvements in speech perception and music appreciation. However, the improvements may be limited by relatively large current spread due to monopolar electric stimulation. While current focusing appears to decrease current spread at least in the vicinity of the stimulation electrode, focused stimulation with today’s devices, which have a limited compliance range, requires use of much wider pulse widths and, consequently, lower stimulation rates. Strategies using stimulation carrier rates below about 300 Hz per channel may result in audible artifacts that can be deleterious to perception (personal observations), and offer limited opportunity to convey fine time structure. In addition, careful considerations of compliance limitations must be applied when fitting “focused” stimulation strategies. A combination of dynamic focusing and steering in appropriate channels may be necessary to drastically improve perception of cochlear implant users.

5.3. Applications to other auditory prostheses

Recently, either penetrating or surface electrodes have been used to stimulate either the cochlear nucleus (Colletti and Shannon, 2005) or the inferior colliculus (Lenarz et al., 2006) of human subjects, with the result of producing auditory sensations; in some cases, open-set speech recognition has been achieved by stimulation of the cochlear nucleus. In addition, direct stimulation of the auditory nerve using a penetrating electrode has been investigated in animals

(Middlebrooks and Snyder, 2007). It is of interest to consider whether field shaping techniques could be applicable to these auditory prostheses as well.

First, it must be considered that field shaping techniques are applicable whenever there is a substantial interaction between electric fields generated by the various electrodes in the vicinity of the target neural tissue. If we define a notion of electric distance between two points, which is roughly equivalent to impedance, then practically, this requirement implies that the inter-electrode distance must be substantially smaller than the effective “electric” distance to the neural targets. In the case of contemporary cochlear implants, the effective “electric” distance between intrascalar electrodes and AN neurons is substantially increased by the presence of high-impedance modiolar wall, whereas the effective “electric distance” between the electrodes is decreased by the low-impedance perilymph. This situation, although sub-optimal from the point of view of selectivity, is conducive to field shaping techniques. However, this would not be the case with penetrating electrodes, which are presumably located very close to the target neural tissue. For example, the field of a monopole produced by a small, e.g., point source, electrode approaches a Dirac delta function when it is measured near the location of the electrode. Placing a small electrode in apposition with target neurons would thus minimize response threshold, but also would minimize the spread of the field produced by current applied at a given level above the minimum threshold. In contrast, padded surface electrodes, e.g. those that are used in auditory brainstem implants, may provide some opportunity for field shaping. As with conventional cochlear implants, goals for field-shaping using those devices may include producing intermediate “virtual” electrodes to increase the number of effective channels, and sharpening, to decrease channel overlap. The latter might be particularly valuable where the electrode is somewhat distant from the tissue. Additionally, field shaping might be applied to control the “shape” of excitation of the structure, where spread of excitation might be directed along the surface of the tissue rather than along the depth of the tissue or vice versa. For example, bipolar stimulation across two surface electrodes will have the effect of reducing current spread both perpendicular to the electrode pair and into the tissue.

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Fig. 1.

Computational model describing monopolar isolevel contours for potentials (A,C) and potential gradients (B,D), and effects of homogeneous (A,B) and nonhomogeneous (C,D) tissue resistivity. Left of each panel shows 3-dimensional representation of potential or gradient (light-grey), elements of model (including tubular bone surrounding scala tympani) and plane of section for contours shown at right of each panel. See text for more detailed description of model and observations. Electrical potentials and potential gradients are derived using a finite element model (Open source software freeFEM3D and Gmsh: Geuzaine and Remacle, 2007; Del Pino and Pironneau, 2007). Contour lines are spaced at 2 dB intervals in A and C and at 4 dB intervals in B and D. Small irregularities in contours shown result from boundary conditions and limited spatial resolution in model. Model dimensions, given in arbitrary units, are: radius of cylindrical scala tympani = 1.75; thickness of tubular bony wall = 0.25; radius of seven spherical electrode contacts = 0.5; axial center to center spacing of electrode contacts = 2; x -coordinate of tip of the peripheral neuronal processes = 2.25. Note that dimensions and shapes of model elements are chosen for convenience and clarity rather than anatomical accuracy.

Fig. 2.

Strength of electrical potential (center column) and potential gradient (right column) predicted by the model of Fig. 1. Configurations were: monopolar (A), focused (partial) tripolar (B), tripolar (C), virtual channel using asymmetric tripole (D), bipole (E) and virtual channel using two monopoles (F). Bone resistivity was set to 10 times resistivity of perilymph and surrounding tissues, and potential and gradient strengths were evaluated at the tips of model peripheral processes ($x = 2.25$). Potential or gradient strength is plotted along the abscissa, and ordinate indicates axial location. Stimulus current used to compute strength in each panel was set to 6 dB above an arbitrary minimum threshold (dashed line). The total cathodic current required to reach threshold in each configuration relative to the monopolar configuration

threshold was: monopolar - 1.0 (0 dB), partial tripolar - 1.3 (2.2 dB), full-tripolar - 2.2 (7.0 dB), asymmetric tripolar - 2.1 (6.6 dB), bipolar - 1.9 (5.8 dB) and two monopolar - 1.2 (1.6 dB). For reference, shaded circles represent locations of intracochlear electrodes, shaded vertical bar represents bony modiolar wall and small circles represent AN neurons.

Fig. 3.

Spatial response profiles showing response in the inferior colliculus to virtual channel stimuli. Successive panels show IC response rate (color) to increasingly larger total applied cathodic current. Variation along abscissa corresponds to variation of virtual channel stimulus. Diagonal trajectory of response peaks (asterisks) for currents above 0.060 mA indicates that response was evoked in regions of IC neurons that were between those evoked by single-channel stimulation (see text for details). (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

Fig. 4.

Spatial tuning curves showing inferior colliculus response to stimulation using several values of tripolar compensation. Tripolar compensation factor is indicated at top left of each panel; 1.0 corresponds to full-tripolar stimulation (no extracochlear electrode current) using intracochlear electrodes #2, 3 and 4, 0.0 corresponds to monopolar stimulation using electrode #3. Red circles indicate current and IC depth for threshold level response. Vertical red lines indicate extent of response along tonotopic axis for stimulation 2 dB above minimum threshold, and demonstrate that spread of response along tonotopic axis evoked by more monopolar stimulation is greater (see text for details). The all-blue rows at top and bottom of each panel, beyond the physical ends of the 1.5 mm recording probe, were included to limit width calculations and do not represent measured data. Note that the range of stimulus current represented along abscissa is not identical for all panels. Color scale same as in Fig. 3. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

Fig. 5.

The average difference between forward masked thresholds for partial tripolar and monopolar maskers as a function of the masker-to-probe electrode displacement. The two curves show data for two different tripolar compensation coefficients σ [0.625 (diamond), and 0.75 and 0.875 (square)]. The error bars indicate the standard deviation of the data. Different subjects were used for the two curves. Although specific significance tests are difficult to apply with a small sample, it appears that on average there is a significant elevation of the masked thresholds for the tripolar stimulation.