Neural Coding of Interaural Time Differences with Bilateral Cochlear Implants: Effects of Congenital Deafness

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Neural Coding of Interaural Time Differences with Bilateral Cochlear Implants: Effects of Congenital Deafness

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Human bilateral cochlear implant users do poorly on tasks involving interaural time differences (ITD), a cue that provides important benefits to the normal hearing, especially in challenging acoustic environments, yet the precision of neural ITD coding in acutely deafened, bilaterally implanted cats is essentially normal (Smith and Delgutte, 2007a). One explanation for this discrepancy is that the extended periods of binaural deprivation typically experienced by cochlear implant users degrades neural ITD sensitivity, by either impeding normal maturation of the neural circuitry or altering it later in life. To test this hypothesis, we recorded from single units in inferior colliculus of two groups of bilaterally implanted, anesthetized cats that contrast maximally in binaural experience: acutely deafened cats, which had normal binaural hearing until experimentation, and congenitally deaf white cats, which received no auditory inputs until the experiment. Rate responses of only half as many neurons showed significant ITD sensitivity to low-rate pulse trains in congenitally deaf cats compared with acutely deafened cats. For neurons that were ITD sensitive, ITD tuning was broader and best ITDs were more variable in congenitally deaf cats, leading to poorer ITD coding within the naturally occurring range. A signal detection model constrained by the observed physiology supports the idea that the degraded neural ITD coding resulting from deprivation of binaural experience contributes to poor ITD discrimination by human implantees.

Introduction

Increasing numbers of profoundly deaf patients are receiving cochlear implants (CIs) in both ears with the goal of restoring the benefits of binaural hearing, including accurate sound localization and improved speech reception in noise. Although some benefits are observed in bilateral CI users, they differ from those experienced by normal-hearing listeners in that interaural time differences (ITDs) provide little advantage. Sound localization with bilateral CI is based primarily on interaural level differences (ILD) (van Hoesel, 2004; Seебer and Fastl, 2008), whereas improvements in speech reception in noise result from attending to the ear with the best signal-to-noise ratio (Litovsky et al., 2006). Bilateral CI provide little binaural “unmasking,” which requires neural processing of ITD, and is important for understanding speech when multiple sound sources are widely distributed in space (Zurek, 1992).

Performance of bilateral CI users on basic psychophysical tasks is consistent with their sound localization and speech reception abilities. ILD discrimination is exquisitely fine (∼0.2 dB) using direct electric stimulation and comparable with normal hearing (1–2 dB) when listening through clinical processors (van Hoesel and Tyler, 2003; Laback et al., 2004; Grantham et al., 2008). Conversely, ITD sensitivity is typically poorer than normal and restricted to a narrow range of stimulus conditions. For the best performers, just noticeable differences (JNDs) in ITD are on the order of 50 μs for low-rate pulse trains, comparable with JNDs in normal-hearing listeners for similar stimuli (Laback et al., 2007). However, ITD JNDs with bilateral CIs are highly variable across subjects, reaching several hundreds of milliseconds in some listeners, and degrade rapidly for pulse rates above 300 pulses per second (pps) (Lawson et al., 1998; van Hoesel and Tyler, 2003; Laback et al., 2007; van Hoesel, 2007; Poon et al., 2009).

In contrast to the typically poor ITD discrimination exhibited by human CI listeners, coding of ITD by inferior colliculus (IC) neurons is essentially as precise in acutely deafened, bilaterally implanted cats as in normal-hearing cats (Smith and Delgutte, 2007a). An important difference between human psychophysics and animal neurophysiology is the extent of deprivation of binaural experience. The acutely deafened cats studied by Smith and Delgutte (2007a) had normal binaural hearing until experimentation, whereas human CI wearers typically experience long periods of auditory deprivation before receiving their first implant and, in many cases, an additional period of binaural deprivation before the second implantation. Such extended periods of deprivation, especially if they include the neonatal period, may degrade neural ITD sensitivity by inducing changes in brainstem neural circuits involved in ITD processing or preventing these circuits from developing normally.
As an initial test of this hypothesis, we recorded from single units in the IC of two groups of bilaterally implanted cats representing the maximum contrast in auditory experience: acutely deafened cats, who had normal hearing before the experiment, and congenitally deaf white cats, who had no auditory experience. We found approximately half as many ITD-sensitive neurons in the congenitally deaf animals compared with the acutely deafened animals and abnormalities in ITD tuning among the neurons that were ITD sensitive. Using a computational model, we show that these physiological differences are likely to have a major impact on psychophysical ITD discrimination.

Materials and Methods
Experiments were performed on 11 barbiturate-anesthetized cats of either sex, divided into two groups. Seven were congenitally deaf white cats raised at Johns Hopkins University. In these animals, the organ of Corti degenerates before the onset of hearing (West and Harrison, 1973; Heid et al., 1998), so that they presumably never hear. The other four were adult cats acutely deafened by coadministration of kanamycin (300 mg/kg, s.c.) and ethacrynic acid (25 mg/kg, i.v.) 1 week before implantation and experimentation (Xu et al., 1993). The congenitally deaf cats may represent a model for early onset (prelingual) deafness in terms of binural experience, whereas the acutely deafened cats model sudden-onset deafness occurring in adulthood. All procedures were approved by the animal care committees at the Massachusetts Eye and Ear Infirmary, the Massachusetts Institute of Technology, and Johns Hopkins University.

There is considerable variability among white cats in the degree and type of cochlear pathology (Ryugo et al., 1998, 2003). To ensure that the white cats used in the present experiments were profoundly deaf, auditory brainstem responses (ABRs) to either clicks or tone pips were measured at 4 and 8 postnatal weeks in the laboratory of D.K.R. (Ryugo et al., 2003). Only animals that showed no response in either ear at the highest level tested (−100 dB sound pressure level [SPL]) were used in the present experiments.

Surgery and cochlear implantation
Surgical and experimental procedures were performed under barbiturate anesthesia, either single injections of Dial in urethane (75 mg/kg, i.p.) or separate injections of Nembutal (37 mg/kg, i.p.) and urethane (300 mg/kg, i.p.). Supplemental doses were administered as needed to maintain areflexia to a toe pinch. Dexamethasone (−0.2 ml, i.m.) was given every 4 h to minimize brain swelling. The trachea was cannulated, and body temperature was maintained at 37°C by means of a feedback-controlled heating blanket. Heart rate, respiration rate, and expired CO2 were monitored continuously throughout the experiment.

The posterior and dorsal aspects of the skull were exposed, and the ear canals were cut to allow insertion of closed acoustic systems. The tympanic bullae were opened to expose the round window. Electrode arrays were inserted 5–6 mm and secured in place using dental cement. A craniotomy was performed over the occipital cortex, which was aspirated to gain access to the IC. A portion of the bony tentorium was removed to maximize the exposure.

Stimulus generation
Electrical stimuli were generated by a 16-bit digital-to-analog converter (National Instruments model PXI-6221) and delivered to the intracochlear arrays through custom, high-bandwidth current sources. Stimuli were presented in a wide bipolar electrode configuration, between the most apical and most basal electrode in the array (5.25 mm separation). This configuration is similar to the monopolar configuration used in clinical devices in that it stimulates neurons throughout the tonotopic axis, but the stimulus artifact is reduced compared with a monopolar configuration (Litvak et al., 2003). Like monopolar stimulation, the wide bipolar configuration produces spatial excitation patterns with a single maximum across the tonotopic axis of the IC (Smith and Delgutte, 2007a).

ABRs
ABRs were measured with both acoustic stimulation, to verify deafness, and electrical stimulation, to assess overall neural sensitivity. ABRs were measured between a bone screw inserted in the vertex of the skull and the stereotaxis apparatus (David Kopf Instruments model 1404). Signals were amplified (acoustic, 94 dB gain; electric, 66 dB), filtered (0.1–10 kHz) and sampled at 25 kHz using a 16-bit analog-to-digital converter (National Instruments model PXI-6221). Responses to 500 stimulus presentations were averaged. Electric stimuli were biphasic current pulses (50 μs/phase) presented at 21/s in wide bipolar configuration and alternated in polarity to cancel stimulus artifact. Acoustic stimuli were 50 μs condensation clicks presented at 21/s through a Realistic 40-1377 tweeter enclosed in a calibrated acoustic assembly inserted into the ear canal. No acoustically evoked responses were observed up to the highest levels tested (−100 dB SPL peak).

The binaural interaction component (BIC) of the electric ABR was measured. First, the relative level between the ears was adjusted to produce equal amplitude wave 4 responses to monolateral stimulation (Smith and Delgutte, 2007b). Wave 4 in cat corresponds to wave V in human ABRs and is generated by lemniscal inputs to the IC (Melcher and Kiang, 1996). After level adjustment, the sum of the responses to stimulation of each ear alone was subtracted from the response to bilateral stimulation to obtain the ABR. The BIC is typically a biphasic waveform (see Fig. 1), and its amplitude was measured between its trough and the following peak (arrows).

Single-unit recordings
Silicon substrate microelectrode arrays with 16 contacts (177 μm2 contact area, 100 or 150 μm spacing between contacts; NeuroNexus) were used for single-unit recordings from the IC. The electrode array was gradually advanced vertically into the IC from dorsal to ventral with a microdrive until the tip reached a maximum depth of −5 mm. In every animal, we sampled the central regions of the IC in which neurons are known to be ITD sensitive in response to electric stimulation (Smith and Delgutte, 2007b). When time permitted, more anterior and posterior regions were sampled as well. Every single unit that could be isolated was characterized, whether it was ITD sensitive or not.

Signals from the microelectrode were amplified and bandpass filtered (300–3000 Hz; RA16; Tucker-Davis Technologies). Typically, signals from adjacent contacts were subtracted to minimize the amplitudes of stimulus artifacts and local field potentials. Subtracted signals were sampled at 200 kHz (National Instruments model PXI-6123), and custom software was used to blank residual stimulus artifacts and detect the times of action potentials by threshold crossing. Only well isolated single units were studied.

Biphasic electric pulses (anodic/cathodic, 50 μs/phase) alternately delivered to the left ear, the right ear, and both ears simultaneously at a rate of 10/s were used as search stimuli. Once a single unit was isolated, its response to the search stimulus was measured as a function of stimulus level in 1 or 2 dB increments to determine the threshold for stimulation of each ear. ITD sensitivity was then studied with pulse train stimuli.

Characterization of ITD sensitivity. Stimuli were low-rate (10–80 pps) periodic trains of biphasic current pulses (anodic/cathodic, 50 μs/phase). Neural responses were measured as a function of ITD at stimulus levels 1–6 dB re single pulse threshold. ITD was varied either statically or dynamically. Static-ITD pulse trains were 300 ms in duration, with a 300 ms silent interval between presentations. The ITD of each pulse was constant within a train but varied across presentations, typically from −2000 μs to +2000 μs (contralateral-leading) in 200 μs steps. Each ITD was typically presented 10–20 times in random order. Dynamic-ITD stimuli (Smith and Delgutte, 2007a) were continuous pulse trains with a rate of 40 pps. Every pulse was presented at a different ITD to create a “binaural pulse beat” stimulus. Specifically, ITD increased from −2000 to +2000 μs in 200 μs steps over 0.5 s and then decreased back to −2000 μs in 0.5 s; the whole 1 s cycle was typically repeated 20 times with no silent interval.

Data analysis
Temporal discharge patterns. Temporal response patterns to pulse stimuli were characterized using period histograms computed from responses to
static-ITD stimuli. For each ITD, spike times relative to the onset of the preceding stimulus pulse were binned with 0.5 ms resolution. The first pulse cycle in each 300 ms train was excluded from the analysis. Period histograms for a range of ITDs were added together to obtain enough spikes for quantification of the temporal discharge pattern. Period histograms were only pooled across ITDs for which the first spike latency was less than the median plus a small tolerance (2.5%). This selection procedure ensures that pooling does not smear the temporal discharge pattern. It is effective because we vary ITD by systematically delaying the stimulus in the lagging ear, keeping the leading-ear delay constant.

Period histograms exhibited a precisely timed early response (less than ~25 ms latency), a poorly timed late response (more than ~25 ms), or both. These components were quantified by fitting each period histogram with a sum of two Gaussian functions with different latencies. The mean latencies were constrained to nonoverlapping intervals (<25 ms and >25 ms) to capture the early and late responses, respectively. A response component (i.e., early or late) was judged to be present if the area under the corresponding Gaussian contained at least 20 spikes. The mean and SD of the Gaussian fit to the early response were used to quantify the latency and jitter, respectively, of pulse-evoked spikes.

Quantification of ITD sensitivity. ITD sensitivity of single-unit rate responses was quantified using a signal-to-noise ratio (SNR) metric based on ANOVA. For static-ITD stimuli, the spike count was summed over the entire 300 ms stimulus duration for each repetition. For dynamic-ITD stimuli, spike counts for each ITD were summed for the ascending and descending part of each 1 s ITD cycle. In either case, the ITD SNR was defined as follows:

\[
\text{ITD SNR} = \frac{\text{Variance in spike count attributable to variation in ITD}}{\text{Total variance in firing rates}}.
\]  

(1)

ITD SNR is the fraction of the variance in neural spike counts accounted for by the variation in stimulus ITD, as opposed to random variability across stimulus trials. It ranges from 0, indicating no ITD sensitivity, to 1, indicating perfectly reliable ITD coding (meaning that the spike counts for each ITD would be identical on every trial). Details of the ITD SNR computation are provided in the supplemental data (available at www.jneurosci.org as supplemental material).

An F test was used to determine whether the dependence of firing rate on ITD was statistically significant (p < 0.025). Summary data are shown for the stimulus condition (stimulus level and pulse rate) that maximizes the ITD SNR. Most of these summary plots combine data obtained in response to static-ITD and dynamic-ITD stimuli because the two responses were generally similar in cases when both were available from the same neuron.

Quantification of ITD tuning. For neurons with significant ITD SNRs, rate–ITD curve shapes were categorized using the four templates of Smith and Delgutte (2007a): peak (positive-going Gaussian), trough (negative-going Gaussian), biphasic (difference of two Gaussians), or sigmoidal (cumulative Gaussian). The data were fit to each of the four templates using the MATLAB function lsqcurvefit (MathWorks). Each rate–ITD curve was assigned the category of the template that yielded the smallest sum-of-squared errors. The best fit was generally quite good (95% of the fits had \( r^2 > 0.5 \), median of 0.88).

Figure 8A illustrates metrics of ITD tuning computed from the best fits to peak-shaped rate–ITD curves and peak portions of biphasic curves. The best ITD is the ITD corresponding to the peak of the fitted curve. The half-width is the difference in ITD between the two points of the fitted curve having 50% of the peak amplitude. The ITD of maximum slope (ITD_{max}) is the point on the fitted curve where the rate is most sensitive to changes in ITD.

Neural population model of ITD discrimination

To assess the functional implications of the deficits in neural ITD sensitivity observed in congenitally deaf cats, a computational model (Hancock and Delgutte, 2004) that predicts normal-hearing ITD discrimination based on physiological properties of IC neurons was extended to the bilateral CI case. Figure 10A shows a block diagram of this detection theoretic model. The model for the CI case comprises a two-dimensional grid of model neurons, each of which has a Gaussian-shaped rate–ITD curve. The SD of the Gaussian (sharpness of ITD tuning) varies systematically along one axis of the grid (corresponding to the tonotopic axis in the normal-hearing case), according to a lognormal distribution fit to the physiological data. The ITD of maximum slope varies along the other axis of the grid according to a normal distribution.

The model simulates a two-alternative forced-choice ITD discrimination experiment by comparing model neural firing rates in response to a reference ITD and a test ITD. Model rates are first summed along the columns of the grid (i.e., across neurons differing in sharpness of tuning). This summation was found to be essential for predicting the dependence of ITD JND on reference ITD in the model for normal hearing (Hancock and Delgutte, 2004). Then, for each column \( i \), the summed rates \( r_i \) are used to compute a standard separation \( D_i \) (analogous to \( d' \) in psychophysics):

\[
D_i = \frac{r_i(ITD_{\text{ref}}) - r_i(ITD_{\text{test}})}{\sqrt{\frac{1}{2}\left[\sigma^2_r(ITD_{\text{ref}}) + \sigma^2_r(ITD_{\text{test}})\right]}}.
\]  

(2)

The firing rate variance across trials, \( \sigma^2_r \), is assumed to be proportional to the mean firing rate (Hancock and Delgutte, 2004). The individual standard separations are combined optimally (Green and Swets, 1988), assuming statistically independent firing rates across columns, to get the overall standard separation \( D \):

\[
D^2 = e \sum_i D_i^2.
\]  

(3)

The test ITD is adjusted to find the value yielding \( D = 0.77 \) (equivalent to 71% correct). The difference between this test ITD and the reference ITD is taken as the predicted ITD JND. The model has only one free parameter, the “detection efficiency” \( e \) (Eq. 3), which is an overall scale factor on the JNDS predicted by the model. For all simulations, \( e \) was held constant at the value that produces accurate predictions of ITD discrimination performance for broadband noise in normal hearing (Hancock and Delgutte, 2004). Thus, the model for the CI case is completely constrained by the IC physiology in deaf animals on the one hand and the normal-hearing performance on the other hand and has no free parameter.

Human psychophysics

In normal-hearing listeners, the ITD JND is best for stimuli near the midline and degrades as the reference ITD is moved away from the midline (Mossop and Culling, 1998); whether this trend also applies with CI is unknown. To provide psychophysical data with which to test the neural population model for the CI case, ITD JNDS were measured as a function of reference ITD in two Advanced Bionics CI implant subjects, both with above-average discrimination performance. Table 1 summarizes the auditory experience of these subjects, including their age at the time of the experiment, the duration of deafness before receiving the first cochlear implant, the length of time between implantations, and the

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Table 1. Human subject characteristics
duration of experience with bilateral implants. Custom interface hard-
ware bypassed the wearable speech processors to control the implanted
current stimulators directly. Human subject procedures were ap-
proved by the Internal Review Board of the Massachusetts Eye and Ear
Infirmary.

The stimuli were periodic trains of biphasic pulses (cathodic/anodic,
27 μs/phase) presented at a rate of 50 pps to a single binaural electrode
pair. Electrode pairs were selected to maximize binaural image fusion, interaural pitch match-
ing, and ITD sensitivity (Poon et al., 2009). During each trial, the subject was presented with
two 300 ms stimuli separated by a 300 ms silent interval. The ITD of the first stimulus was always
the reference ITD, which remained fixed for an entire adaptive threshold run. The
ITD of second stimulus was either incremented or decremented by an amount ΔITD. The
subject’s task was to report whether the second stimulus was heard to the left or right of the
first using a keyboard. Feedback was provided after every trial.

The ITD JND was measured using a two-
down one-up adaptive procedure converging
to 71% correct performance (Levitt, 1971).
The threshold was initially calculated as the mean ΔITD for the last 8 of 14 total reversals
and then scaled by √2 to yield JNDS compara-
tible with those that would be obtained using a
standard two-interval, two-alternative forced-
choice paradigm (Hartmann and Rakerd,
1989). (Our procedure differed from the stan-
dard in that the first interval always contained
the reference ITD.) ITD JNDS were measured
at four reference ITDs: 0, 215, 646, and 1077 μs.
For each subject, two or four runs were
averaged to obtain the final JND for each refer-
ce through each reference ITD (except for the 1077 μs reference, in
which there was only one run).

Results

ABR binaural interaction component
The binaural interaction component of the ABR (Dobie and Norton, 1980; Le-
vine, 1981) specifically reflects the activity of neural populations whose activity is
modulated by binaural stimulation, in-
cluding those responsible for processing and encoding ITD. Figure 1, A and B,
shows BIC waveforms measured in re-
sponse to biphasic electric pulses from
one acutely deafened cat and one congeni-
tally deaf cat, respectively. The two sets of
waveforms are broadly similar, with a
prominent negative deflection at 2.3 ms
latency, followed by a positive deflection
at 3 ms. However, the peak BIC amplitude
is larger for the acutely deafened cat than
for the congenitally deaf cat (note the dif-
ferent vertical scales in Fig. 1A, B). Figure
1C shows BIC amplitudes as a function of stimulus level in congenitally deaf and acutely deafened
cats. BIC amplitudes were measured between the
notch and the following peak (arrows). C, BIC amplitudes as a function of stimulus level in congenitally deaf and acutely deafened
cats. Gray shading shows the range of BIC amplitudes measured in acutely deafened cats by Smith and Delgutte (2007b). D, Monaural wave 4 amplitudes are also smaller in congenitally deaf cats. Amplitudes shown are average of responses to stimulation
of each ear alone. E, BIC latencies are similar between congenitally deaf and acutely deafened cats, with the exception of one cat.
F, Monaural wave 4 latencies tend to be shorter in congenitally deaf cats. Values shown are averages for stimulation of each ear
alone. In B–F, stimulus levels are expressed relative to the monaural ABR thresholds, defined as the stimulus amplitude required
to evoke 1 μV wave 4. Monaural wave 4 amplitudes were always equalized during binaural stimulation by application of an
appropriate ILD.

Figure 1. The BIC of ABR is reduced in congenitally deaf cats. A, B, Example BIC waveforms from one acutely deafened cat (A)
and one congenitally deaf cat (B). Each trace corresponds to a different stimulus level. BIC amplitude was measured between the
notch and the following peak (arrows). C, BIC amplitudes as a function of stimulus level in congenitally deaf and acutely deafened
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amplitude at each level (t test, \( p = 0.017 \)). Thus, the ABR data suggest there is a specific deficit in binaural processing in addition to an overall decrease in ABR amplitude, which may reflect a combination of decreased synchrony in neural activity and partial loss of spiral ganglion neurons in the congenitally deaf cats (Heid et al., 1998).

Wave 4 latencies were slightly shorter in congenitally deaf cats than in acutely deafened cats (Fig. 1F). BIC latencies were generally similar between the two groups, with the exception of one congenitally deaf cat, in which the BIC was delayed by \( \sim 0.6 \) ms with respect to wave 4 (Fig. 1E). This animal did not yield enough single units to determine whether their responses were unusual compared with those of other deaf white cats.

Overall, these results suggest that, although congenital deafness does not preclude binaural interactions in the auditory brainstem, deprivation of auditory experience does impair binaural circuitry to a greater extent than monaural responses. These findings are consistent with the report that the ABR BIC has an abnormal latency on the first day of bilateral implant use in congenitally deaf children (Gordon et al., 2007).

**ITD sensitivity of IC neurons**

Consistent with the BIC results, we found that ITD coding by IC neurons is degraded in congenitally deaf cats compared with acutely deafened cats. Figure 2 illustrates typical qualitative differences between the two groups of animals. Responses to a 20 pps pulse train varied in ITD are shown for one IC neuron from an acutely deafened cat (top) and one from a congenitally deaf cat (bottom). In the neuron from an acutely deafened cat, the spikes are precisely locked to each stimulus pulse (Fig. 2A), and the firing rate is strongly modulated by ITD (Fig. 2B), as indicated by the large, highly significant value of the ITD SNR (0.69, \( p < 0.001 \)). The ITD tuning curve is peak shaped with a best ITD of 200 \( \mu \)s and shows a preference for contralateral-leading stimuli. This neuron has no spontaneous activity, as is the case for most IC neurons in acutely deafened cats.

In contrast, in the neuron from a congenitally deaf cat, there is significant spontaneous activity (4.1 spikes/s), as commonly observed in this group of animals. During stimulation by the 20 pps pulse train, each pulse suppresses the spontaneous activity for \( > 50 \) ms, followed by a poorly timed rebound of activity before the next pulse (Fig. 2C). The end of the pulse train is also followed by a period of rebound activity exceeding the spontaneous rate and lasting \( > 100 \) ms. The mean firing rate is not obviously modulated by ITD (Fig. 2D), consistent with the small, statistically insignificant value of ITD SNR (0.15, \( p = 0.95 \)).

Figure 3A shows the distributions of ITD SNR for the two groups of animals. In acutely deafened cats, ITD SNR is distributed almost uniformly across the entire range from 0 (poor ITD coding) to almost 1 (highly reliable ITD coding). In contrast, in congenitally deaf cats, there are many more neurons with low ITD SNRs and fewer with high ITD SNRs. The median ITD SNR is significantly lower in the congenitally deaf group (0.19 vs 0.45, \( p < 0.001 \), Wilcoxon’s rank-sum test). We classified as “ITD-sensitive” neurons for which the ITD SNR is significantly greater than 0 at the \( p = 0.025 \) level. Only 48% (55 of 114) of neurons in congenitally deaf cats were ITD sensitive by this criterion compared with 84% (83 of 99) in acutely deafened cats (Table 2). Thus, approximately half as many IC neurons are ITD sensitive in congenitally deaf cats compared with acutely deafened cats.

The fraction of ITD-sensitive neurons in our samples from both groups of animals might be influenced by biases in the dis-
tributions of neurons along the cochleotopic axis of the IC. For example, because the intracochlear electrode array is inserted only a short distance into the cochlea past the round window, it may be hard to find responsive neurons in low-frequency, dorsal regions of the IC, which receive predominant inputs from the cochlear apex. This dorsal region is the primary target of projections from the medial superior olive (MSO) and thus might contain a higher proportion of ITD sensitive neurons. However, Figure 3 shows that ITD SNR does not depend on the depth of the microelectrode recording site relative to the dorsal surface of the IC, in either group of animals. Furthermore, there are no major differences in the distribution of recording depths between the two groups of animals. Hence, the observed difference in prevalence of ITD sensitivity between the two groups of animals is not likely to reflect differences in the IC regions sampled.

Relationship between ITD sensitivity and other response properties

Neurons from the two groups of animals differ in other response properties besides ITD sensitivity, including temporal discharge patterns and spontaneous rate (Table 2). Because this is, to our knowledge, the first report of single-unit responses from the IC of congenitally deaf cats, we describe these differences in this section and also examine whether variations in these response properties correlate with degraded ITD sensitivity.

Three types of temporal discharge patterns to low-rate pulse trains were observed in the IC of both groups of animals (Fig. 4). Most commonly, each stimulus pulse evoked a short-latency, well-timed “early” response comprising one or two spikes (Fig. 4A). In other neurons, each stimulus pulse produced a short-latency period of inactivity, followed by a long-latency, poorly timed “late” response (Fig. 4B). Late responses were frequently preceded by suppression of spontaneous activity, as in the neuron from a congenitally deaf cat in Figure 2. Although late responses could occur in the absence of spontaneous firing (so that suppression was not directly observable), they likely represent recovery from suppression or rebound from inhibition rather than direct excitatory effects. Early and late responses sometimes occurred together (Fig. 4C).

The vast majority of neurons in acutely deafened cats showed tightly pulse-locked early responses, whereas neurons from con-
congenitally deaf cats showed a mixture of excitatory and suppressive effects (Fig. 5A).

Almost every neuron in acutely deafened cat had an early component in its response (either alone or in combination with a late response) compared with only two-thirds of the neurons in congenitally deaf cat (94 vs 66%). Conversely, late responses rarely occurred in acutely deafened cats but were present half the time in congenitally deaf cats (13 vs 53%). The difference in the incidence of response shapes between the groups of cats was highly significant (p < 0.001, χ² test). The increased incidence of poorly timed late responses in congenitally deaf cats is consistent with the lower ABR wave 4 amplitude in this group of animals, although wave 4 represents the synchronized activity of ascending inputs to the IC rather than IC activity itself (Melcher and Kiang, 1996).

The presence of a late response in the discharge pattern was typically associated with degraded ITD sensitivity (Fig. 5B). Across both groups of cats, the median ITD SNR was 0.13 in units with only a late response versus 0.57 in units with only an early response. Thus, the lower prevalence of well-timed, pulse-evoked spikes clearly contributes to degraded ITD sensitivity in the congenitally deaf cat. However, even among early responding units, the median ITD SNR was still twice as large in acutely deafened cats as in congenitally deaf cats (0.64 vs 0.33).

A possible explanation for the difference in ITD sensitivity among early responding neurons could be differences in spike timing. We quantified mean spike latency and latency jitter for each unit using the mean and SD of Gaussian curves fitted to the timing. We quantified mean spike latency and latency jitter for early responding neurons could be differences in spike timing. We quantified mean spike latency and latency jitter for early responding neurons could be differences in spike timing. We quantified mean spike latency and latency jitter for early responding neurons could be differences in spike timing.
Figure 8. Congenital deafness alters ITD tuning metrics. Analysis applied to rate–ITD curves containing a peak. A, Illustration of metrics derived from Gaussian fits to rate–ITD curves. Best ITD, ITD of the peak response; Halfwidth, width of rate–ITD curve at 50% of maximum rate. ITD<sub>MS</sub>, ITD for which the slope of rate–ITD curve is maximal. B, Best ITD distribution is broader and lacks contralateral bias in congenitally deaf cats. C, Rate–ITD curves are broader in congenitally deaf cats. D, ITD<sub>MS</sub> distribution is not concentrated in naturally occurring range of ITD in congenitally deaf cats. B, D, Dashed lines, approximate naturally occurring ITD range in cat.

Figure 8A shows the distributions of best ITD and half-width of ITD tuning (i.e., broadly tuned neurons tend to have large best ITDs). The trends in Figure 9, A and B, are closely related because the main lobe of the rate–ITD curve for broadband stimuli widens with decreasing BF (Yin et al., 1986; McAlpine et al., 2001). This widening is apparent in the average rate–ITD curves for each BF quintile (Fig. 9D). The correlation between best ITD and half-width tends to align the rising slopes of the rate–ITD curves near 0 ITD and misalign them at more lateral ITDs. This alignment is key in accounting for the greater psycho-physical ITD acuity near the midline (Hancock and Delgutte, 2004).

It is, a priori, unclear whether the relationship between best ITD and half-width holds in the case of bilateral CI. In normal-hearing animals, the dependence of half-width on BF for broadband stimuli primarily reflects the effects of cochlear filtering (Yin et al., 1986). Furthermore, there is evidence that small interaural mismatches in the BFs of the inputs to the binaural coincidence detector neurons likely contribute to the dependence of best ITD on BF (Joris et al., 2006; Day and Semple, 2009). With cochlear implants, however, one might not expect to observe these relationships because cochlear mechanisms for frequency selectivity are bypassed.

Nevertheless, Figure 9C shows that a positive correlation between best ITD and half-width does exist in the IC of acutely deafened, bilaterally implanted cats (r<sup>2</sup> = 0.41, p < 0.001). For this purpose, we combined our sample (n = 31) of peak-shaped and bipolar rate–ITD curves with a larger sample (n = 77) from Smith and Delgutte (2007a). A similar analysis could not be conducted in congenitally deaf cats because our sample of ITD-sensitive neurons was too small to characterize the joint distribution of ITD tuning parameters. Figure 9E shows the average rate–ITD curves for each quintile of half-width, illustrating alignment of slopes near the midline despite the wide variation in half-widths. Conversely, ITD<sub>MS</sub> is uncorrelated with half-width, as shown in Figure 9G (r<sup>2</sup> = 0.0013, p = 0.70). These relation-
ships are used in the following section to constrain a neural population model of perceptual ITD discrimination.

**Psychophysics and neural population model of ITD discrimination**

An important question is the extent to which the changes in neural ITD coding observed in congenital deafness affect ITD perception. The answer depends on assumptions about the central processing of the ITD information available in the activity pattern of the population of IC neurons. For example, good ITD discrimination might still be achievable with a small number of ITD-sensitive neurons if performance is primarily determined by the most sharply tuned neurons in the population. To approach these questions systematically and make our assumptions explicit, we used a signal detection model for assessing the impact of congenital deafness on ITD discrimination. For this purpose, we adapted to the deaf case a neural population model that was shown previously to account for key aspects of ITD discrimination in normal-hearing listeners (Hancock and Delgutte, 2004).

The normal-hearing model is a two-dimensional grid of neurons (Fig. 10 A), each of which is characterized by a rate–ITD curve whose shape is determined by two independently distributed parameters: BF and best interaural phase difference (best ITD × BF). The model predicts the sharper psychophysical ITD acuity on the midline compared with more lateral locations (Mossop and Culling, 1998) by summing the rate–ITD curves of model neurons across BF. This summation harnesses the alignment of the rising slopes of the rate–ITD curves across BF near the midline (and their misalignment at lateral locations) to produce composite rate–ITD curves that are much steeper near the midline than at lateral locations.

Although BF is undefined in deaf animals, the tight correlation between best ITD and half-width of ITD tuning that is observed in both normal-hearing (Fig. 9 B) and acutely deafened (Fig. 9 C) cats makes it possible to use half-width as a proxy for BF in the deaf model. However, the normal-hearing model is based on the responses of neurons with BF < 2 kHz, whereas the cochlear implant model is based on data spanning the entire cochleotopic axis (Fig. 3 B). Thus, an important assumption is that the principle of operation of the model generalizes across BF.

We constructed a population model for the bilateral CI case in which half-width and ITD$_{MNS}$ vary along the two independent axes of the grid of model neurons (Fig. 10 A) because these parameters are uncorrelated in the physiological data (Fig. 9 G). We started with a model for acutely deafened cats using the constraints imposed by the physiological data from this group of animals and then modified this model for the congenitally deaf case. In the acutely deafened case, all model neurons are ITD-sensitive, and half-width and ITD$_{MNS}$ are independently distributed according to the data of Figure 9 G. In every other respect, the bilateral CI model operates in identical manner to the normal-hearing model (see Materials and Methods). Importantly, the sole free parameter of the model, the detection efficiency $\varepsilon$ (Eq. 3), was chosen to produce accurate predictions of normal-hearing ITD discrimination for broadband noise (Hancock and Delgutte, 2004), making the model completely constrained in the CI case.

To obtain psychophysical data to directly test model predictions, we measured ITD discrimination for 50 pps pulse trains in two postlingually deaf human bilateral cochlear implant users (Table 1). The gray line in Figure 10 B shows the mean ITD JNDS as a function of reference ITD for these two listeners. JNDS are small near the midline (90 $\mu$s) but increase as the reference ITD increases, similar to the trend exhibited by normal-hearing listeners (Mossop and Culling, 1998). The acutely deafened model
The deficits in ITD coding observed in IC single units are broadly similar to those reported previously in multiunits from the auditory cortex of deaf white cat (Tillein et al., 2010), including reduced incidence of ITD-sensitive units and weaker contralateral bias in best ITD distributions. However, differences in methods and analysis techniques make it difficult to tell whether the cortex simply inherits its deficits from the IC or whether additional deficits arise at the thalamocortical level.

Our finding of significant effects of auditory deprivation on ITD coding at the level of auditory brainstem contrasts with studies of plasticity in cochleotopic organization and binaural pro-
cessing of ILD. In these studies, plasticity is either observed only at the thalamocortical level or is much more dramatic in the cortex than in the IC, whether the plasticity is induced by restricted cochlear lesions (Robertson and Irvine, 1989; Irvine and Rajan, 1994), monaural conductive hearing loss (Popescu and Polley, 2010), or neonatal deafening (Snyder et al., 1990; Raggio and Schreiner, 1999). Auditory experience may be particularly critical for maturation of the ITD-processing circuitry in auditory brainstem because the operation of this circuit places extraordinary requirements on the precision of spike timing. ITD processing may provide a particularly sensitive model system for studies of neural development, plasticity, and repair.

By what mechanisms might congenital deafness affect ITD coding in IC neurons? Because deprivation of auditory experience does not preclude precise spike timing to electric pulse trains in the auditory nerve (Shepherd and Javel, 1997; Sly et al., 2007), the deficits are likely to arise primarily in the brainstem. Congenitally deaf cats and mice show structural abnormalities in the end bulbs of Held, the specialized synapses between auditory nerve fibers and spherical bushy cells (SBCs) of the cochlear nucleus (Ryugo et al., 1997, 1998; Lee et al., 2003). Transmission at this critical synapse is altered in congenitally deaf mutant mice in vitro (Oleskevich and Walmsley, 2002). Disruption of this synapse specialized for precise transmission of timing information is likely to impair the encoding of ITD by neurons in the MSO and lateral superior olive (LSO), which receive excitatory inputs from SBCs.

Interuption of auditory inputs by cochlear ablation also results in abnormalities in the calyces of Held, the giant synapses formed by axons of cochlear nucleus globular bushy cells on medial nucleus of the trapezoid body (MNTB) cells (Jean-Baptiste and Morest, 1975). The inhibitory projections from MNTB are critical to ITD sensitivity of LSO neurons (Joris and Yin, 1995) and also play a role in ITD tuning in MSO (Brand et al., 2002). The normal maturation of inhibitory synapses in MSO depends on auditory experience because these synapses have less focused spatial distributions in both congenitally deaf cats (Tirko et al., 2009) and normal-hearing gerbils raised in omnidirectional noise (Kapfer et al., 2002). This reduction of inhibition to MSO neurons could contribute to the increase in spontaneous activity in IC units. In both MSO and ventral cochlear nucleus, normal synapses can be partially restored in congenitally deaf animals by chronic electric stimulation through cochlear implants (Ryugo et al., 2005; Tirko et al., 2009).

An imbalance between excitation and inhibition might also contribute to deficits in ITD sensitivity in congenitally deaf cats. Interruption of peripheral auditory inputs causes a downregulation of inhibitory gain in brainstem auditory nuclei, including cochlear nucleus, LSO, and IC (Takesian et al., 2009). A balance between ipsilateral excitation and contralateral inhibition is essential for ITD sensitivity in LSO neurons (Joris and Yin, 1995), and inhibition helps make ITD coding robust to variations in ILD and overall level (Peña et al., 1996; Dasika et al., 2005). Barbiturate anesthesia may also affect the balance between excitation and inhibition and contribute to the strength and prevalence of long-latency responses. However, it is unlikely to account for differences between acutely deafened and congenitally deaf cats because identical methods were used in both groups.

In normal-hearing cats, guinea pigs, and gerbils, firing rates of IC and MSO neurons are most sensitive to changes in ITD within the naturally occurring range of ITD as a result of a correlation between best ITD and the width of ITD tuning curves (McAlpine et al., 2001; Brand et al., 2002; Hancock and Delgutte, 2004). The cochlear traveling wave is likely to contribute to this correlation (Joris et al., 2006), although central neural mechanisms including conduction delays and inhibition may also play a role. Interestingly, we found that the trend also holds in acutely deafened, bilaterally implanted cats (Fig. 9C), although cochlear mechanics are bypassed. The following scenario might account for this finding. Early in development, there might be a broad range of sharpness of ITD tuning and, independently, a broad distribution of best ITDs as observed in adult congenitally deaf cats (Fig. 8) and in animals raised in omnidirectional acoustic environments (Seidl and Grothe, 2005). Auditory experience might provide selective pressure to create a neural network maximally sensitive to changes in ITD about the midline by strengthening synaptic inputs that favor correlation between half-width and best ITD and pruning those that do not, regardless of whether the correlation for each particular input is produced by cochlear or central mechanisms.

Our finding of major deficits in ITD coding in congenitally deaf cats is in harmony with psychophysical and evoked potential studies showing that auditory experience (especially during the neonatal period) impacts ITD sensitivity in human bilateral CI users. The one order of magnitude difference in ITD JNDS predicted by our signal detection model between congenitally deaf animals and acutely deafened animals is in line with the differences in performance among subjects differing in the timing and duration of deafness in psychophysical studies (Poon et al., 2009; Litovsky et al., 2010). Litovsky et al. (2010) found that preliminarily deaf subjects could not lateralize 100pps pulse trains based on ITD, whereas subjects with childhood and adult onset deafness could. Importantly, all subjects could lateralize based on ILD, showing that the effect of deprivation of early auditory experience is specific to ITD processing. Gordon et al. (2008) measured ABRs in bilaterally implanted young children with early onset (probably congenital) deafness 9 months after the beginning of binaural implant use. They find delayed BIC latencies if the interval between the two implantations exceeds 2 years but not in cases of simultaneous or short-delay implantations, pointing to the importance of early binaural experience in the maturation of brainstem binaural circuits.

The present results raise important questions: does the period of auditory deprivation have to encompass the neonatal period to produce abnormalities in neural ITD coding, or do some of these abnormalities also occur in the case of adult-onset deafness? Can chronic stimulation through cochlear implants reverse these abnormalities and improve ITD coding in deaf animals? If so, what stimulation paradigms and training regimens are most effective for this purpose? Answers to such questions will shed additional light on the development and plasticity of neural ITD coding and likely suggest methods for improving the ability of bilateral CI listeners to use ITD information.

References


