Auditory filter nonlinearity in mild/moderate hearing impairment

Richard J. Baker

Human Communication and Deafness, School of Education, University Of Manchester, Oxford Road, Manchester M13 9PL, United Kingdom

Stuart Rosen

Department of Phonetics and Linguistics, University College London, Wolfson House, 4 Stephenson Way, London NW1 2HE, United Kingdom

(Received 8 February 2000; revised 31 May 2001; accepted 10 December 2001)

Sensorineural hearing loss has frequently been shown to result in a loss of frequency selectivity. Less is known about its effects on the level dependence of selectivity that is so prominent a feature of normal hearing. The aim of the present study is to characterize such changes in nonlinearity as manifested in the auditory filter shapes of listeners with mild/moderate hearing impairment. Notched-noise masked thresholds at 2 kHz were measured over a range of stimulus levels in hearing-impaired listeners with losses of 20–50 dB. Growth-of-masking functions for different notch widths are more parallel for hearing-impaired than for normal-hearing listeners, indicating a more linear filter. Level-dependent filter shapes estimated from the data show relatively little change in shape across level. The loss of nonlinearity is also evident in the input/output functions derived from the fitted filter shapes. Reductions in nonlinearity are clearly evident even in a listener with only 20-dB hearing loss. © 2002 Acoustical Society of America. [DOI: 10.1121/1.1448516]

PACS numbers: 43.66.Ba, 43.66.Dc, 43.66.Sr [SPB]

I. INTRODUCTION

A fundamental property of the peripheral auditory system is that it operates as a kind of frequency analyzer. It can therefore be thought of in terms of a bank of overlapping bandpass filters. Because of their central and obligatory role in determining the nature of any further auditory processing, much effort has gone into characterizing the properties of these auditory filters. In hearing-impaired (HI) listeners with sensorineural hearing loss, the overwhelming consensus is that frequency selectivity is reduced, and that such a reduction has important implications for understanding the communication difficulties experienced by such listeners (e.g., de Boer and Bouwmeester, 1974; Leshowitz and Lindstrom, 1977; Pick et al., 1977; Wightman et al., 1977; Florentine et al., 1980; Tyler et al., 1982; Tyler et al., 1984; Glasberg and Moore, 1986; Faulkner et al., 1990; Laroche et al., 1992; Sommers and Humes, 1993a, 1993b; see Moore, 1995, 1998 for reviews).

Several studies have also assessed the relationship between frequency selectivity and absolute threshold (Tyler et al., 1982; Pick and Evans, 1983; Glasberg and Moore, 1986; Lutman et al., 1991; Bergman et al., 1992; Laroche et al., 1992). Generally, selectivity decreases with increasing hearing loss, at least for losses above 30–40 dB HL. Below this level the selectivity remains approximately constant, with little correlation of filter bandwidth (quantified by the equivalent rectangular bandwidth, or ERB) with threshold (e.g., Fig. 9 of Glasberg and Moore, 1986, and Figs. 3 and 4 of Laroche et al., 1992). Indeed, Pick and Evans (1983) noted a marked dissociation between frequency selectivity and threshold at 4 kHz—some listeners had very little hearing loss, but a significant increase in filter bandwidth. Similarly, West and Evans (1990) investigated selectivity and sensitivity in young adults with and without a history of exposure to amplified music. They found that in the subjects more regularly exposed to loud music the filter bandwidths were 10%–15% wider than in the least exposed, while there was little difference in sensitivity between the two groups. Thus, while moderate hearing losses typically lead to a concomitant reduction in selectivity, this may not necessarily be the case for mild hearing losses. Indeed, loss of selectivity may occur without significant loss of sensitivity.

In the physiological domain it has also been shown that cochlear damage affects both sensitivity and selectivity. The outer hair cells (OHCs) appear to be theprime mediators of the cochlear amplifier (e.g., Davis, 1983; Dallos, 1992), and damage to them leads to cochleae that are more linear, and show reduced sensitivity and frequency selectivity, than is found in an intact cochlea (e.g., Dallos and Harris, 1978; see Ruggiero, 1991 and Patuzzi, 1996, for reviews).

Thus, current evidence suggests that OHC damage affects both sensitivity and selectivity. The outer hair cells (OHCs) appear to be the prime mediators of the cochlear amplifier (e.g., Davis, 1983; Dallos, 1992), and damage to them leads to cochleae that are more linear, and show reduced sensitivity and frequency selectivity, than is found in an intact cochlea (e.g., Dallos and Harris, 1978; see Ruggiero, 1991 and Patuzzi, 1996, for reviews).

Thus, current evidence suggests that OHC damage affects both sensitivity and selectivity. As well as the loss of sensitivity and selectivity due to OHC damage, there is also the question of how changes in filtering across level are affected by OHC damage. It is only recently, however, that attempts have been made to clarify how filtering changes with level in normal-hearing (NH) listeners. While significant effort was initially put into showing whether or not filter shapes change at all with stimulus level in NH listeners (Patterson, 1971; Weber, 1977; Pick, 1980; Lutfi and Patterson, 1984), relatively little attention has been paid to the accurate characterization of this phenomenon.

*Electronic mail: richard.baker@man.ac.uk
Moore and Glasberg (1987; also Glasberg and Moore, 1990) attempted to tie together the existing notched-noise masking data from various studies to provide a coherent description of how auditory filters change shape as a function of frequency and stimulus level. Moore and Glasberg (1987) argued, on the basis of an excitation pattern model derived from their filter shapes, that the filter shape is determined primarily by the signal level at the input of the filter, and that filters whose shape depends on their own output do not produce realistic excitation patterns. We have shown previously, however, that this is not true. Using a technique of fitting filter shapes to the notched-noise masking data as an explicit function of stimulus level (PolyFit procedure—Rosen and Baker, 1994; Rosen et al., 1998), we have been able to describe how the filter shapes vary with stimulus level much more accurately. Using this technique, we have also shown that when the variations in the low-frequency skirt of the filter are taken into account, an output-controlled nonlinearity can produce realistic excitation patterns (Baker et al., 1998). Recently, Glasberg and Moore (2000) have also found an output-controlled nonlinearity to more adequately account for notched-noise masking results than an input-controlled one.

 Such an issue cannot be treated as a minor detail, as it has ramifications both in the design of experiments, and the interpretation of their results. Rosen et al. (1998) also argued that using a fixed-level probe tone is more appropriate than the fixed-level notched-noise masker typically used to measure auditory filter shapes at a single level in NH listeners. Related to this issue, Leek and Summers (1993) also used the level-dependent fitting procedure of Rosen et al. (1992) to help clarify changes in filtering in the presence of background noise. They showed that the use of probe-level-dependent filter shapes better accounted for their data (measured at a single fixed masker level, with different levels of background noise) than a level-independent filter shape.

 For listeners with OHC damage, the cochlea will behave more linearly, with filter shapes changing less with increasing stimulus level (Stelmachowicz et al., 1987; Murnane and Turner, 1991). The data of Moore et al. (1985; their Table V) also suggest a reduction in the change of filter shape with level in impaired ears, although they did not discuss this aspect of their data. In such cases the confound between measurement technique and nonlinearity will be less important, as the cochlea is likely to behave in a more linear manner. Thus, notched-noise masking experiments in which the masker level is fixed will result in filter shapes similar to those in which the probe tone level is fixed.

The aim of the present study is to use the notched-noise masking technique coupled with the PolyFit analysis to quantify the changes in frequency selectivity in mild/moderate hearing loss over a range of stimulus levels, and to relate these to measurements of selectivity in NH listeners.

II. METHODS

A. Listeners

Table I presents summary data concerning the 5 HI listeners who participated in the study. The degree of hearing impairment at 2 kHz ranged from 20 to 50 dB HL. All testing was done monaurally, and in a single ear per listener.

CH was a 24-year-old male with a reported history of exposure to loud sounds as a musician in a rock band. ET was a 24-year-old female with a mild, bilateral high-frequency sensorineural hearing loss confirmed in childhood. The remaining three listeners (BL, CS, and MF) were recruited from an adult audiology clinic. None of the five listeners demonstrated any middle-ear involvement in their impairment as assessed by bone-conduction thresholds and tympanometry.

B. Threshold estimation

Masked thresholds were determined for sinusoidal probe tones in the presence of notched-noise maskers with variable notch widths. The notches were placed both symmetrically and asymmetrically about the probe frequency, and either the probe level or the noise level was varied to determine thresholds. For listeners CH and ET, a two-interval, two-alternative forced-choice (2I-2AFC) paradigm with feedback was used to estimate the 79% point on the psychometric function (Levitt, 1971). From a starting level at which the probe was clearly audible, the varying sound, either probe or masker, was initially changed in 5-dB steps, with step size decreasing by 1 dB after each turnaround. Once the step size reached 2 dB, it remained constant for a further eight turnarounds, the mean of which was taken as the threshold. For the remaining three listeners a 2I-2AFC maximum-likelihood procedure was used to estimate the 90% point on the psychometric function (see Baker and Rosen, 2001, for a description of the procedure used). In both cases, listeners responded on a button box, with illuminated buttons indicating presentation intervals and providing feedback.
For each particular combination of notch width and fixed probe or fixed masker level, at least two thresholds per listener were obtained. Where two measurements of the same condition in the same listener differed by more than 4 dB, one or two further measurements were taken and the mean of all measurements used.

**C. Stimulus configurations**

The outside edges of the masker noise were fixed at \( \pm 0.8 \times f_p \) [400 and 3600 Hz for a probe frequency \( f_p \)] of 2 kHz. A maximum of 16 different notch conditions was used, 6 symmetric and 10 asymmetric. The frequencies \( f \) of the edges of the notch (the inner edges of the two noise bands) are specified in normalized frequency units \( g \) relative to the probe frequency as given by \( g = (|f - f_p|)/f_p \). In the symmetric conditions, both notch edges were placed at normalized values of 0.0, 0.1, 0.2, 0.3, 0.4, and 0.5. In the asymmetric condition, one of the notch edges was set at a normalized value of 0.0, 0.1, 0.2, 0.3, and 0.4, while the other was set to 0.2 normalized units further away (0.2, 0.3, 0.4, 0.5, and 0.6). When the masker level was fixed, a subset of noise spectrum levels \( N_g \) was chosen, ranging from 20–60 dB SPL in 10-dB steps, and when the probe level was fixed, a subset of probe levels \( P_p \) was chosen, ranging from 30–80 dB SPL in 10-dB steps. The specific values used for each listener depended on the nature of the hearing loss, and are detailed in Table II. In addition to measurements at 2 kHz, CS and MF were also tested at frequencies where their sensitivity was in the normal range (0.5 kHz for CS and 1 kHz for MF).

**D. Stimulus generation**

All the stimuli were computer generated at a sampling frequency of 20 kHz. The time waveform of the probe consisted of a steady-state portion of 360 ms plus 20-ms raised-cosine onset and offset. The probe was temporally centered within the masker, which consisted of a 460-ms steady-state portion with 20-ms cosine-squared onset and offset. To generate the masker, the desired frequency spectrum was defined by setting all the spectral components (spaced at intervals of 0.61 Hz) within the appropriate frequency limits to have equal amplitudes while those outside were set to zero. Non-zero components had their phase randomized uniformly in the range of 0–2\( \pi \) radians. An inverse FFT was then applied to generate the time waveform. At the start of each threshold determination, a 3.2768-s buffer of noise was generated for use during that test. On each trial, a 500-ms portion of the buffer was chosen randomly for each of the two masker intervals within each trial.

The probe and masker were played out through separate channels of a stereo 16-bit D/A converter (12-bit D/A for listener CH) and attenuated independently under computer control before being electrically mixed. The signal was then presented monaurally via Etymotic ER2 insert earphones to listeners CH and ET via Beyer DT48 circumaural headphones to the remaining listeners. For listeners using the Beyer DT48 headphones, the noise spectrum was shaped at the synthesis stage to give a flat spectrum of appropriate spectrum level as measured in a B&K 4157 ear simulator.

**E. Analyses**

All analyses were performed on an individual basis using the mean threshold for each notch/level combination. A variety of models was fitted to each data set, using the PolyFit technique described in detail by Rosen et al. (1998). All of the models were variants of the asymmetric roex\((p,w,t)\) model (Patterson et al., 1982). These included simplified models in which, for example, the upper half of the filter was described with a roex\((p)\) shape whereas the lower half was a complete roex\((p,w,t)\) shape. It is also necessary to estimate \( k \), the signal-to-noise ratio necessary for detection at the output of the filter. All of these parameters can be arbitrary polynomial functions of the level of the masker or the probe, but we have never investigated models with more than a quadratic dependence on level (that is, three coefficients per parameter to be estimated). Finally, we also estimate an absolute threshold by allowing a further parameter to be added to the predicted masked threshold in power terms (see Rosen et al., 1998 for details). This parameter was added to take into account the fact that for low-level wide notch conditions the probe-tone level is governed by the listener’s absolute threshold. In the present study this estimated threshold parameter was always within 5 dB of the listener’s pure-tone threshold at that frequency.

**III. RESULTS**

**A. Growth-of-masking functions**

1. **Data and predictions**

The notched-noise masking experiments carried out across a range of stimulus levels allow the data to be plotted...
as growth-of-masking (GOM) functions for each notch configuration. In producing these GOM functions we follow the convention of Lutti and Patterson (1984) in plotting signal-to-noise ratio at threshold versus masker spectrum level. This has the advantage over plotting signal level versus masker level in that a simple linear filter with constant detector efficiency will result in a set of horizontal parallel lines, one for each of the different notch conditions. For a filter that broadens with level and a constant detector efficiency, the GOM functions for the no-notch condition will be approximately horizontal, while the GOM functions for the wider notch conditions will show a positive slope (for data above absolute threshold).

Figure 1 shows such GOM functions for each of the five HI listeners, and also for the average of the three NH listeners described in Sec. II A of Rosen et al. (1998). The curves through the data points show the fits to the data obtained using the PolyFit procedure described later in Sec. B. For the NH listeners there is a clear convergence of GOM functions as stimulus level is increased—indicating a broadening of the underlying filter. For listener CH, with the mildest hearing loss (20 dB HL), there appears to be less convergence than for the NH listeners. This is even clearer for listeners ET, CS, and BL, where the GOM functions (and the predictions obtained from the PolyFit procedure) are approximately parallel. For listener MF there appears to be a slight divergence of the GOM functions, indicating a slight narrowing of the filter with increasing stimulus level. These changes in the slopes of the GOM functions relative to the NH listeners are quantified in the following section.

2. Estimates of amplitude compression

If it is assumed that maskers near the tone frequency undergo the same compression as the tone, whereas maskers well below the tone frequency are subject to little compression, it is possible to use the GOM functions to estimate the degree of amplitude compression arising from the underlying cochlear mechanisms. Oxenham and Plack (1997) have used such a technique in forward masking. They measured GOM functions for a 6-kHz probe tone with a 6-kHz masker and with a 3-kHz masker. The idea is that with the 6-kHz masker, both the tone and the masker will be subject to the same “CF” compression. However, when the masker is at 3 kHz it will not be subject to compression at the 6-kHz place while the probe will. Thus, a 10-dB increase in the 3-kHz masker will result in a greater than 10-dB increase in probe level at masked threshold. Using this technique, Oxenham and Plack (1997) estimated a basilar membrane input/output (I/O) function slope of about 0.25, similar to the slope measured directly in physiological experiments.

Making similar assumptions about the nature of the underlying filtering mechanism (i.e., linearity at frequencies well below CF and also that the high-frequency side of the filter is approximately level independent), similar estimates can be made from the data given in Fig. 1. Taking the thresholds for the conditions where the lower masker band is at least 1 octave below the probe tone frequency—notch conditions (0.6,0.4), (0.5,0.3), and (0.5,0.5)—Fig. 1 shows that the GOM functions are approximately parallel when points near absolute threshold are excluded. Using these three notch conditions, estimates of the slopes of the GOM function for the low-frequency skirt of the filter were derived from a simple linear regression. Here, it was assumed that the GOM functions from the three conditions had identical slopes, but were permitted to have different intercepts. Only data points 6 dB or more above absolute threshold were used. Examples of the fits are shown in Fig. 2 for two of the HI listeners and also from the NH listeners of Rosen et al. (1998) (plotted as signal level versus masker spectrum level).

The solid horizontal line shows the absolute threshold, while the solid diagonal lines show the linear fits to the data for the (0.5,0.3) and (0.6,0.4) notch conditions. The dashed diagonal lines in the lower two plots indicate the slope of GOM functions for the NH listeners.

Since the slope of the GOM functions for the no-notch condition is approximately one (zero when signal to masker ratio is plotted versus masker spectrum level), the reciprocal of the slope of the GOM functions obtained as above gives an estimate of the slope of the I/O function of the underlying cochlear mechanism (although this may be somewhat confounded by the presence of suppression in simultaneous masking). Table III shows the estimated slopes of the GOM functions using the linear fits described above and also the derived I/O function slopes. In addition to the data for the normal and impaired listeners at 2 kHz, estimates are given at 0.5 and 1 kHz from listeners CS and MF, respectively, i.e., frequencies where absolute thresholds for these listeners were within the normal range. For comparison, Table III also includes estimates at these two frequencies for the two listeners of Baker et al. (1998).

The estimated I/O function slope for the NH listeners at 2 kHz is 0.578, indicating compression approaching 2:1. However, compression was reduced for listener CH (compression ratio approximately 1.25:1), whose absolute threshold at 2 kHz was 20 dB HL. For listeners ET, CS, and BL the derived I/O functions approach linearity (1.03:1, 1.1:1, and 1.1:1, respectively). For listener MF, the GOM functions show a degree of divergence; this is reflected in an I/O function slope of greater than unity indicating an expansive nonlinearity. The derived slopes for the listeners CS and MF at 0.5 and 1 kHz were similar to those obtained previously from NH listeners.

B. Level-dependent filter shapes

Using the PolyFit procedure described by Rosen and colleagues (Rosen and Baker, 1994; Rosen et al., 1998) the notch-delay masking data were fitted with level-dependent filter shapes based on the rounded exponential (roex) family described by Patterson et al. (1982). Any of the fitted parameters could be allowed to be constant or a linear or quadratic function of stimulus level (either masker or probe level). Results are presented for probe-dependent models because they have been found to give a better fit to notch-delay data than masker-dependent models (see also Rosen et al., 1998). Each data set was fitted with a set of filter shapes of varying complexity. Typically a complex model was first fitted to the data. One of the parameters was then removed from the model and the data refitted. If the error in the fit...
FIG. 1. Growth-of-masking functions for normal-hearing listeners (average of three listeners from Rosen et al., 1998) and hearing-impaired listeners. The notch widths are indicated at the bottom of each column. The solid curves show the predictions derived from the filter shapes fitted to the data as described in Sec. III B. Note that for any particular listener a single model describes all the data. Thus, for one row of plots the solid lines are not independent of each other, but are all determined by the same fit to the data. The diagonal solid line represents the listener’s absolute threshold.
worsened then that parameter was deemed necessary to the model. Thus, it was possible to see which parameters were important to accurately describe the data and which were relatively superfluous. The models that fitted each listener’s data best vary in structure since the degree of nonlinearity evident in the filtering varies. The listeners with a greater degree of hearing loss show filtering which can be described by models with fewer free parameters than those where a greater degree of nonlinearity is present. The filter shapes obtained using the “best” model for each listener are shown in Fig. 3.

The filter shapes obtained by Rosen et al. (1998) for averaged data from 3 NH listeners are presented at the top of Fig. 3. For these data the high-frequency side of the filter could be described by a one-parameter, \( \text{roex}(p) \), shape in which the slope parameter \( (p_u) \) was allowed to be a quadratic function of probe-tone level. The lower frequency side of the filter was better described by a three-parameter, \( \text{roex}(p,w,t) \), shape in which the slope of the filter in the passband \( (p_l) \) was constant while the slope of the skirt \( (t_l) \) was a linear function of probe level. Also, the point of intersection between the passband and tail of the filter \( (w_l) \) was required to be a linear function of probe level, with the tip-to-tail ratio decreasing as probe level increased. Following the nomenclature of Rosen et al. (1998), this model is referred to as \( p_{1312}x^2x \), where \( p \) specifies that the parameters are dependent on the probe-tone level, and the digits refer to the number of fitted coefficients in the polynomial that describes how that parameter varies with tone level. The parameters are specified in the order \( p_l, p_u, k, w_l, w_u, t_l, t_u \) (\( x \) indicates that a parameter is not required in the fit). In order to estimate the changes in gain at CF as stimulus level is increased, we used the approach of Rosen et al. (1998). This technique relies on the assumption that the gain of the filter is constant well below the filter’s CF, as evidenced in basilar membrane measurements. We thus normalize all filter shapes to have the same gain (an arbitrary 0 dB) at a frequency that is 0.4 times the filter CF.

The rest of the panels in Fig. 3 show the filter shapes for the five HI listeners, along with their absolute thresholds. The number of parameters required to adequately describe the filter shapes varied between 4 and 7 for the HI listeners.

### TABLE III

<table>
<thead>
<tr>
<th>Subject</th>
<th>Frequency (kHz)</th>
<th>Threshold, dB HL</th>
<th>GOM slope</th>
<th>I/O slope (GOM)</th>
<th>I/O slope (filter)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Normal—Rosen et al. (1998)</td>
<td>2</td>
<td>8</td>
<td>1.731</td>
<td>0.578</td>
<td>0.47</td>
</tr>
<tr>
<td>CH</td>
<td>2</td>
<td>20</td>
<td>1.257</td>
<td>0.796</td>
<td>0.72</td>
</tr>
<tr>
<td>ET</td>
<td>2</td>
<td>30</td>
<td>1.028</td>
<td>0.973</td>
<td>0.81</td>
</tr>
<tr>
<td>CS</td>
<td>2</td>
<td>35</td>
<td>1.105</td>
<td>0.905</td>
<td>0.77</td>
</tr>
<tr>
<td>BL</td>
<td>2</td>
<td>50</td>
<td>1.108</td>
<td>0.903</td>
<td>0.80</td>
</tr>
<tr>
<td>MF</td>
<td>2</td>
<td>50</td>
<td>0.652</td>
<td>1.534</td>
<td>1.18</td>
</tr>
<tr>
<td>Normal—Baker et al. (1998)</td>
<td>0.5</td>
<td>9</td>
<td>1.284</td>
<td>0.779</td>
<td>0.55</td>
</tr>
<tr>
<td>CS</td>
<td>0.5</td>
<td>10</td>
<td>1.371</td>
<td>0.729</td>
<td>0.54</td>
</tr>
<tr>
<td>Normal—Baker et al. (1998)</td>
<td>1</td>
<td>5</td>
<td>1.776</td>
<td>0.563</td>
<td>0.52</td>
</tr>
<tr>
<td>MF</td>
<td>1</td>
<td>15</td>
<td>1.709</td>
<td>0.585</td>
<td>0.34</td>
</tr>
</tbody>
</table>

FIG. 2. Examples of growth-of-masking functions with data taken from the middle column of Fig. 1 and replotted as signal level against masker level. The solid diagonal lines show two of the three linear fits to the wide notch data [the third being for the (0.5,0.5) notch condition not shown here]. The dashed diagonal lines in the lower two plots indicate the slope of the normal-hearing listeners’ functions. The horizontal solid line represents the listeners’ absolute threshold.
compared with 9 for the NH listeners, and the parameter structures of those plotted in Fig. 3 are given in the sixth column of Table IV. These “best-fitting” models were selected according to the strategy described above such that the simplest shape is chosen that does not result in a large increase in the rms error of the fit. Clearly all of the HI listeners show a reduced change in gain at CF, reflecting a more linear filter compared with those of NH listeners. It is also apparent that the two listeners who were tested at frequencies of normal sensitivity (CF and MF) display an unimpaired degree of nonlinearity at these frequencies. All the listeners, apart from MF, show filters that broaden and decrease in gain with increasing stimulus level. The 2-kHz measurement of MF shows slight changes in the reverse direction, with the filter at 60 dB SPL being slightly broader than that at 80 dB SPL. It must be emphasized, however, that the reduced degree of nonlinearity in the HI listeners means that the filter shapes can, in some cases, be fitted almost as well with filter shapes that don’t change at all with stimulus level.

Table IV gives the goodness of fit (rms error) for three different filter shape models to the data of each listener. The model with the greatest number of parameters obviously fits the data best. Removing parameters from this model to find the best fit gives a marginally worse fit to the data (columns 6 and 7 of Table IV). Removing all the level dependency from the fitted model further increases the rms error. As the fitted shapes are describing the same underlying sets of data, it is to be expected that the more linear the underlying filter, the less the difference would be in the goodness of fit between the linear and nonlinear models. For the NH listeners the linear model shows a 515% increase in rms error relative to the most complex model. For the HI listeners this difference is much reduced, ranging from 75% in the least impaired to 12% in the most impaired, again providing evidence for a more linear filter.

C. Estimated compression

We have already used GOM functions to estimate the slope of the underlying basilar-membrane I/O function. Our technique of normalizing filter shapes at a frequency significantly below CF (as illustrated in Fig. 3) also allows an estimate of this I/O function by measuring the change in gain at CF across the range of stimulus levels used. Figure 4 plots the estimated I/O function slope against the absolute threshold for the five HI listeners, and also for the nine NH listeners measured at 2 kHz by Rosen et al. (1998). For the NH listeners the estimated slope is about 0.47, giving a compression ratio just over 2:1. This estimated slope shows little variation over a 25-dB range of absolute thresholds (~25 to 20 dB HL). For the HI listeners (with absolute thresholds ranging from 20 to 50 dB HL) the range is from 0.72 to 1.18 (see the last column of Table III).

While the actual values of the I/O slopes derived from the filter shapes differ from those estimated from the GOM-functions, the general pattern is the same. Each of the HI
listeners shows values less than those of NH listeners and MF shows a slightly expansive nonlinearity. The discrepancies that do exist between these two measures may be due to the fact that the values estimated from the GOM functions use only a limited subset of the data, whereas those estimated from the fitted filter shapes use the entire data set. A final point from Fig. 4 is that results from listener CH suggest a markedly less compressive cochlea than in one of the NH listeners (listener MB from Rosen et al., 1998) despite the fact that their absolute thresholds are the same. Clearly the labels NH and HI for these two listeners are misleading, since they have the same absolute threshold. However, the impaired filtering of CH suggests that such a classification based on absolute threshold alone does not tell the whole story.

D. Equivalent rectangular bandwidth

From the fitted filter shapes it is possible to provide a simple quantitative measure of selectivity in terms of filter bandwidth. The measure that has typically been used is the ERB. For a probe-level-dependent filter shape the ERB can be calculated over a range of individual probe levels. Figure 5 shows how the ERB, calculated from the best filter models shown in Fig. 3 (and given in Table IV), changes as a function of probe-tone level. For the NH listeners the ERB remains approximately constant at low levels where the low-frequency skirt of the filter has little influence on the amount of energy the filter would pass. It is not until the probe level is increased above 70 dB SPL that the low-frequency skirt makes a significant contribution to the amount of energy passed by such a filter. For the HI listeners there are three major points worth noting. First, at low probe levels the ERB is always broader than the NH listeners. Second, the ERBs measured in the impaired ears change by smaller amounts than in the NH ears—the filter is more linear. Third, at high levels the ERBs measured in the impaired ears are more similar to those of the normal ears.

IV. DISCUSSION

The emphasis of this study was to characterize the way in which auditory filters in HI listeners change in shape as stimulus level increases. While previous studies have demonstrated that cochlear hearing loss results in broadened auditory filters when using the notched-noise masking technique, such studies have nearly all used a fixed masker-level paradigm. In characterizing filter shapes for ears where OHC loss is severe the choice of fixed masker-level or fixed probe-level paradigm will have little consequence, as the cochlea in question will be approximately linear. However, when comparing these “linear” filters with those of NH listeners, the choice of paradigm in the NH listeners is crucial. Rosen et al. (1998) have shown that it is more appropriate to use a fixed probe level than a fixed masker level and argued that

<table>
<thead>
<tr>
<th>Subject</th>
<th>Threshold (dB HL)</th>
<th>Number of thresholds</th>
<th>p2212222</th>
<th>m2212222</th>
<th>Best model</th>
<th>rms error for best model</th>
<th>p1111111 (linear)</th>
</tr>
</thead>
<tbody>
<tr>
<td>NH</td>
<td>5</td>
<td>158</td>
<td>0.81</td>
<td>1.47</td>
<td>p1312x2x</td>
<td>0.86 (6%)</td>
<td>4.98 (515%)</td>
</tr>
<tr>
<td>CH</td>
<td>20</td>
<td>110</td>
<td>1.19</td>
<td>1.47</td>
<td>p1112x1x</td>
<td>1.20 (1%)</td>
<td>2.08 (75%)</td>
</tr>
<tr>
<td>ET</td>
<td>30</td>
<td>94</td>
<td>0.99</td>
<td>1.13</td>
<td>p1112x1x</td>
<td>1.07 (8%)</td>
<td>1.69 (71%)</td>
</tr>
<tr>
<td>CS</td>
<td>35</td>
<td>111</td>
<td>1.68</td>
<td>2.14</td>
<td>p2112x1x</td>
<td>1.76 (5%)</td>
<td>2.45 (46%)</td>
</tr>
<tr>
<td>BL</td>
<td>50</td>
<td>96</td>
<td>1.16</td>
<td>1.24</td>
<td>p1112x1x</td>
<td>1.25 (8%)</td>
<td>1.48 (28%)</td>
</tr>
<tr>
<td>MF</td>
<td>50</td>
<td>93</td>
<td>1.45</td>
<td>1.31</td>
<td>p211xxxx</td>
<td>1.60 (10%)</td>
<td>1.63 (12%)</td>
</tr>
</tbody>
</table>

TABLE IV. Details of goodness of fit of various filter shapes to the notched-noise masking data at 2 kHz. Complexity of fitted filter shapes is denoted, for example, in the form p2212222 or p1111111, where p indicates the parameters change as a function of probe-tone level (m—masker spectrum level). The digit denotes that a parameter is constant (1), a linear function (2—2 fitted coefficients) or a quadratic function (3—3 fitted coefficients). “x” indicates that a parameter is not required. The order of the parameters is pl, pu, k, wl, wu, tl, tu, where “l” indicates the lower, and “u” the upper side of the filter. The percentages in brackets show the percentage increase in rms error relative to the p2212222 model.
the filter shape appears to be controlled by its own output, rather than by its input, via some sort of feedback mechanism on the basilar membrane. The same argument was followed by Baker et al. (1998) to produce a realistic output-controlled excitation pattern model. More recently, Glasberg and Moore (2000) have suggested that the filter shape “depends on the power at its output divided by the gain” and that “this dependence may arise because filtering on the BM involves a feedback mechanism.” The likely mediators of such a feedback mechanism are the OHCs, which provide a “fast mechanical positive feedback” to the basilar membrane (Patuzzi, 1996). Here, we have used the same procedure as Rosen et al. (1998) to describe filter shapes from listeners with mild to moderate sensorineural hearing loss in a way that allows direct comparison with the NH data.

We have been able to estimate the degree of compression from normal and impaired ears, first using an approach somewhat similar to that of Oxenham and Plack (1997), and second indirectly from the change in gain at the CF of the fitted filter shapes. As expected, both these methods show a reduced degree of compression in HI listeners compared to NH listeners. For the HI listeners the GOM functions are more parallel across the different notch widths than in the NH listeners, indicating a smaller change in filter shape across level.

This reduced nonlinearity is evident in the filter shapes fitted to the HI data (Fig. 3). For the impaired listeners, the “tip-to-tail” gain is about 20 dB (slightly more for CH, the least-impaired listener), which is similar to the higher level filters from the NH listeners. However, the change in tip-to-tail gain increases with decreasing stimulus level much more slowly in HI listeners than in NH listeners.

One thing that is somewhat unusual is the degree of loss of nonlinearity present in listener CH. His threshold at 2 kHz is 20 dB HL, yet the degree of compression is only about 1.26:1. Listener MB from Rosen et al. (1998) had the same absolute threshold, but showed a compression ratio of just over 2:1 (right-most “X” in Fig. 4). It is tempting to speculate that these differences reflect mainly an OHC loss in listener CH and a loss not involving OHCs in MB, although other factors such as differences in middle-ear efficiency cannot be ruled out.

Subject MF shows evidence of a slightly expansive, rather than compressive, nonlinearity. The higher level filter shapes are slightly sharper than the lower level shapes. This in itself may be of little significance as a linear filter shape can fit the same data almost as well. However, there are two other points worth noting from MF’s filter shapes in Fig. 3. First, the high-frequency side of the filter at 2 kHz appears to be abnormally steep. Second, the degree of compression in the filter measured at 1 kHz in this listener seems abnormally large (26.4-dB change in gain for a 40-dB change in probe level giving a compression ratio of 2:9:1). The reason for these differences is not clear.

Quantitative comparison of the selectivity of these listeners using the ERB clearly shows that at low stimulus levels the HI ears show less selectivity than NH ears. However, at higher levels the NH filters broaden and become more similar to those measured in impaired ears. The impaired ears themselves show a much smaller change in ERB with increasing stimulus level. One problem with using the ERB as a measure of selectivity is that it is dominated by the filter tip and is little affected by the skirt. Rosen et al. (1998) showed that the low-frequency skirt of the filter is an important part of the filter shape, as is also the case in psychoacoustic tuning curves (PTCs). At low stimulus levels the low-frequency skirt has little effect on the ERB. It is not until the stimulus reaches higher levels that the ERB increases significantly (as the tip-to-tail ratio becomes small). A similar effect is observed with PTCs. Nelson et al. (1990) showed that for forward-masked PTCs the Q10 dB remains relatively constant at low levels, and only begins to decrease at high stimulus levels where the tip-to-tail ratio decreases. For the HI listeners in the present study, the effect of the impairment is to reduce the sharpness (and gain) of the tip at lower stimulus levels—resulting in a reduced tip—tail ratio, and hence a larger ERB than for NH listeners at the same level. At higher stimulus levels, the ERBs of the NH and HI are more similar.

The compression ratios derived from the fitted filter shapes and from the GOM functions in NH listeners are approximately 2:1. Such values are typical for estimates using simultaneous masking (e.g., Stelmachowicz et al., 1987). In forward masking compression ratios are more typically of the order of 4 or 5:1 (Oxenham and Plack, 1997). This latter figure is more comparable to the compression ratios measured directly from the basilar membrane. Previous comparisons of selectivity measured using both forward and simultaneous masking at a single level suggest that a greater degree of nonlinearity may be revealed by forward masking than by simultaneous masking, and that using forward masking to compare NH and HI listeners will result in a larger difference in the derived compression ratios (e.g., Moore and Glasberg, 1986). Using the PolyFit procedure to characterize changes in selectivity across level using forward masking in NH listeners should reveal a greater degree of nonlinearity than is observed in simultaneous masking, and should lead to measurements of compression that are more directly comparable to those of Oxenham and Plack (1997) and to the physiological measurements.

V. CONCLUSIONS

In summary, we have used notched-noise masking and the PolyFit procedure to characterize auditory filter shapes as a function of stimulus level in mild/moderate hearing impairment. Comparison of the fitted filter shapes to previous results from NH listeners shows that the filter shapes in HI listeners are broader and change less with increasing stimulus level. Estimates of basilar-membrane input–output function slopes from masking functions and from the fitted filter shapes are in broad agreement with each other, suggesting compression ratios of approximately 2:1 in NH listeners and a reduced degree of compression in HI listeners.

ACKNOWLEDGMENTS

We are grateful to Dr. B. Cadge and the staff at the Adult Audiology Clinic, Royal National Throat Nose and Ear Hos-


