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Auditory filter nonlinearity in mild/moderate hearing impairment

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Abstract

Sensorineural hearing loss has frequently been shown to result in a loss of frequency selectivity. Less attention has been paid to the level dependency of selectivity that is so prominent a feature of normal hearing. The aim of the present study is to characterise such changes in nonlinearity as manifested in the auditory filter shapes of listeners with mild/moderate hearing impairment. Notched-noise masked thresholds were measured over a range of stimulus levels at 2kHz 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.

1 Introduction

One of the most fundamental properties of the peripheral auditory system is that it operates as a kind of frequency analyser. It can therefore be thought of in terms of a bank of overlapping band-pass filters. Because of their central and obligatory role in determining the nature of any further auditory processing, much effort has gone into characterising the properties of these *auditory filters.* Apart from numerous investigations of auditory filtering in normal hearing listeners, there have also been many such studies in listeners with hearing loss, primarily because any reductions in selectivity are expected to have wide-ranging implications for understanding the difficulties experienced by hearing-impaired listeners (*e.g.* de Boer and Bouwmeester, 1974; Pick, et al., 1977; Glasberg and Moore, 1986; Florentine, et al., 1980, Faulkner, et al., 1990; Laroche, et al., 1992; Leshowitz and Lindstrom, 1977; Tyler, et al., 1984; Tyler, et al., 1982; Wightman, et al., 1977; Sommers and Humes, 1993a; Sommers and Humes, 1993b; see Moore, 1995 and Moore, 1998 for reviews). The overwhelming consensus from these studies is that sensorineural hearing loss, whether due to noise damage, ototoxic effects or age, results in a loss of frequency selectivity (*i.e.* a broadening of the auditory filter).

Several studies have also tried to correlate the frequency selective ability of the ear with the absolute threshold (Bergman, et al., 1992; Glasberg and Moore, 1986; Laroche, et al., 1992; Lutman, et al., 1991; Tyler, et al., 1982). 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 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 4kHz - listeners with 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 more exposed listeners the

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bandwidths were 10-15% wider than in the least exposed, with little difference in sensitivity between the 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. Specifically, several studies have shown that damage to the outer hair-cells (OHCs) reduces the sensitivity and frequency selectivity of the nerves innervating the organ of Corti in the region of damage (e.g. Dallos and Harris, 1978; see Ruggero, 1991 for review). Measurements of basilar-membrane (BM) movement, which show similar selectivity to that observed in auditory nerve recordings (Narayan, et al., 1998), demonstrate a similar lability of response to that observed in the auditory nerve under various insults such as death, acoustic trauma or application of furosemide (see Patuzzi, 1996 for review of cochlear mechanics). Ruggero and Rich (1991) used the diuretic furosemide to alter hair-cell response in chinchillas and showed a reversible loss of tuning and sensitivity at the centre-frequency (CF) of the BM tuning curve. This behaviour, along with other evidence, points to an important involvement of the OHCs in enhancing sensitivity of the organ of Corti at low stimulus levels, an enhancement that is frequency specific at a given place on the BM, and thus also results in an increase in selectivity. The OHCs appear to be the prime mediators of this cochlear amplifier (e.g. Dallos, 1992; Davis, 1983) and damage to them appears to lead to cochlear mechanics that are more linear than is found in an intact cochlea.

Thus, current evidence suggests that OHC damage is reflected in, and is probably causally related to loss of sensitivity and loss of frequency. At least as important as overall changes in sensitivity due to hearing impairment are the changes in filtering across level. Only recently has this issue been clarified even for normal hearing listeners. While significant effort was initially put into showing whether or not filter shapes change at all with stimulus level in normal hearing listeners, relatively little attention has been paid to the accurate characterisation of this phenomenon. Patterson (1971) measured thresholds for probe tones at three different levels of single noise-band maskers placed either above or below the probe tone frequency and found a filter broadening "entirely due to a decrease in the off-band attenuation rate on the low side of the auditory filter”. In a similar manner, Weber (1977) used a notched-noise masking technique (with symmetrically placed notches only) to measure growth of masking. Fitting rounded-exponential (ROEX) filter shapes to the data, Weber showed a clear broadening of the filter skirts with increasing stimulus level. Pick (1980) used a comb-filtered noise (also known as rippled-noise) as a masker. His derived filter shapes showed a broadening on the low frequency side, as did those of Lutfi and Patterson (1984) who used asymmetric notched-noises.

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. These authors have then used these descriptions of filtering to produce an "excitation pattern" model to attempt to describe the pattern of peripheral auditory excitation from psychoacoustic experiments. Moore and Glasberg (1987) argued, on the basis of their predicted excitation patterns, that the filter-shape is determined primarily by the signal level at the input of the filter, and that filters that depend 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 shape varies with stimulus level in a more robust manner. In particular, we have been able to quantify the behaviour of the skirts of the auditory filter as stimulus level is increased. Using this technique we have shown that when the variations in the low-frequency skirt of the filter are taken into account, an output-controlled nonlinearity produces realistic excitation patterns (Baker, et al., 1998).

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) argued that using a fixed notched-noise masker to measure an auditory filter whose shape depends on its own output (via a feedback mechanism) will result in an artificially narrow imputed filter shape because the filter being measured becomes narrower as the notch is widened. Thus, the correct method is to fix the probe tone level in the experiment.

For listeners with linearised cochleae due to OHC damage the confound between measurement technique and nonlinearity will be less important. Thus notched-noise masking experiments in which the masker level is fixed will result in similar filter shapes to those in which the probe tone level is fixed. While this may be true, making the distinction between fixed masker and fixed probe paradigms less significant for hearing-impaired listeners, the nature of the nonlinearity still has important ramifications for comparisons of filter shapes between hearing impaired and normal listeners. The aim of this present study is to quantify the changes in frequency selectivity in mild/moderate hearing loss, and to relate these to measurements of selectivity in normal hearing listeners.

2 Methods

2.1 Listeners

Table 1 presents summary data concerning the 5 hearing impaired listeners who participated in the study. The degree of hearing impairment at 2kHz ranged from 20dB HL to 50dB HL. All testing was done monaurally, and in a single ear per listener.

Subject	Age	Threshold (dB HL)										
		125Hz	250Hz	500Hz	750Hz	1kHz	1.5kHz	2kHz	3kHz	4kHz	6kHz	8kHz
BL	64	40	45	50	55	55	55	***50***	55	60	60	55
CH	24	10	5	5	--	5	--	***20***	--	15	--	5
CS	59	15	15	***10***	15	30	35	***35***	40	45	65	80
ET	24	--	15	5	--	10	20	***30***	--	35	--	40
MF	50	20	20	10	10	***15***	20	***50***	60	60	70	60

Table 1. *Absolute threshold of five hearing impaired listeners measured using standard pure tone audiometry. Values in bold italics represent the frequencies at which the listeners' frequency selectivity was measured using notched-noise masking.*

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 sensori-neural hearing loss confirmed in childhood. The remaining three listeners (BL, CS and MF) were recruited from an adult audiology clinic. All five listeners demonstrated no middle ear involvement in their impairment as assessed by bone-conduction thresholds and tympanometry.

2.2 Threshold estimation

Masked thresholds were determined for sinusoidal probe tones of 2 kHz in the presence of notched-noise maskers with variable notch widths. The notches were placed both symmetrically and asymmetrically about the probe and either the probe level or the noise level could be varied to determine the 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 (3 up - 1 down, 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 8 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. 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, two thresholds per listener were typically obtained. Where two measurements of the same condition in the same listener differed by more than 4 dB, a further one or two measurements were taken and the average of all measurements used.

Listener	Absolute Threshold (dB HL)	Frequency (Hz)	Fixed masker spectrum levels (dB SPL)	Fixed probe levels (dB SPL)
BL	50	2000	40, 50, 60	60, 70, 80
CH	20	2000	30, 40, 50, 60	40, 50, 60, 70
CS	35	2000	40, 50, 60	50, 60, 70, 80
ET	30	2000	40, 50, 60	50, 60, 70
MF	50	2000	40, 50, 60	60, 70, 80
CS	10	500	30, 50	30, 50, 70
MF	15	1000	30, 50	30, 50, 70

Table 2. *Stimulus levels used in notched-noise masking experiments at 2kHz.*

2.3 Stimulus configurations

The outside edges of the masker noise were fixed at ± 0.8 x f_0 (400 and 3600 Hz for a probe frequency of 2 kHz). A maximum of sixteen different notch conditions were used, 6 symmetric and 10 asymmetric. The frequencies of the edges of the notch are specified in normalised frequency units (*g*) relative to the probe frequency as given by $g=(|f - f_0|)/f_0$. In the symmetric conditions, both notch edges were placed at normalised 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 normalised value of 0.0, 0.1, 0.2, 0.3 and 0.4, while the other was set to 0.2 normalised 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_0) was chosen, ranging from 20-60 dB SPL/Hz in 10 dB steps and when the probe level was fixed, a subset of probe levels (P_s) was chosen, ranging from 30-80 dB SPL, again in 10 dB steps. The actual values used for each listener depended on the nature of the hearing loss, and are detailed in table 2. In addition to measurements at 2 kHz, CS and MF were also tested at frequencies where their sensitivity was in the normal hearing range (500 Hz for CS and 1kHz for MF).

2.4 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 onsets and offsets. The probe was temporally centred within the masker which consisted of a 460 ms steady-state portion with 20 ms cosine-squared onset and offsets. 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 phases randomised uniformly in the range of 0-2π 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 and 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.

2.5 Analyses

All analyses were performed on an individual basis using the mean threshold for each notch/level combination. A variety of models were fit 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. 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. The value of this parameter is chiefly governed by the low-level wide notch conditions where the probe tone level reaches absolute threshold.

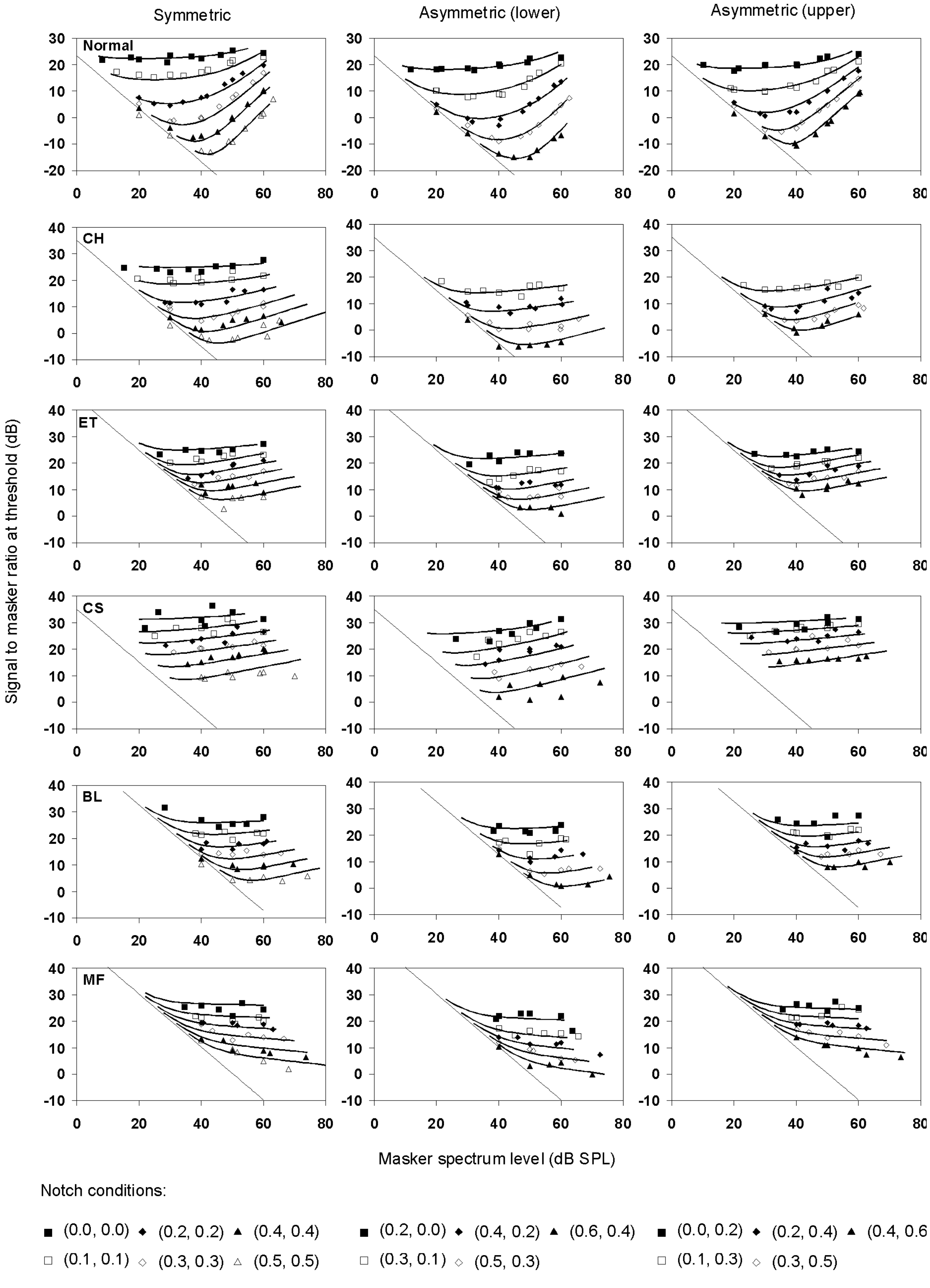

Figure 1. *Growth of masking functions for normal hearing (average of 3 listeners from Rosen, et al., 1998) and hearing impaired listeners. The notch widths are indicated at the bottom of each column. The solid curves show the predictions derived from the filter shapes fitted to the data as described below. 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 all determined by the same fit to the data. The diagonal solid line represents the listener's absolute threshold.*

3 Results

3.1 Growth of Masking functions

3.1.1 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 Lutfi 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. For a filter that broadens with level and a constant efficiency, the no-notch condition will be approximately horizontal, while the growth of masking 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 5 hearing impaired listeners, and also for the average of the three normal listeners described in section IIA 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 section B. For the normal hearing listeners there is a clear convergence of GOM functions as stimulus level is increased – indicating a significant broadening of the underlying filter. For listener CH, with the mildest hearing loss (20dB HL), there is already much less convergence than for the normal 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.

3.1.2 Estimated compression

The growth of masking functions shown in figure 1 give a clear indication of the degree of nonlinearity. 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 growth of masking functions to estimate the degree of compression of the underlying cochlear mechanisms. Oxenham and Plack (1997) have used such a technique in a forward masking paradigm. They measured growth of masking functions for a 6kHz probe tone with a 6kHz masker and with a 3kHz masker. The idea is that with the 6kHz masker, both the tone and the masker will be subject to the same "CF" compression. However, when the masker is at 3kHz it will not be subject to this compression while the probe will. Thus, a 10dB increase in the 3kHz masker will result in a greater than 10dB increase in probe level at masked threshold. Using this technique, Oxenham and Plack (1997) were able to estimate a BM I/O function slope of about 0.25, similar to those 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 figure 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) — figure 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 model. 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 6dB or more above absolute threshold were used. Examples of the fits are shown in figure 2 for two of the hearing-impaired listeners and also from the normal hearing 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 normal hearing listeners.

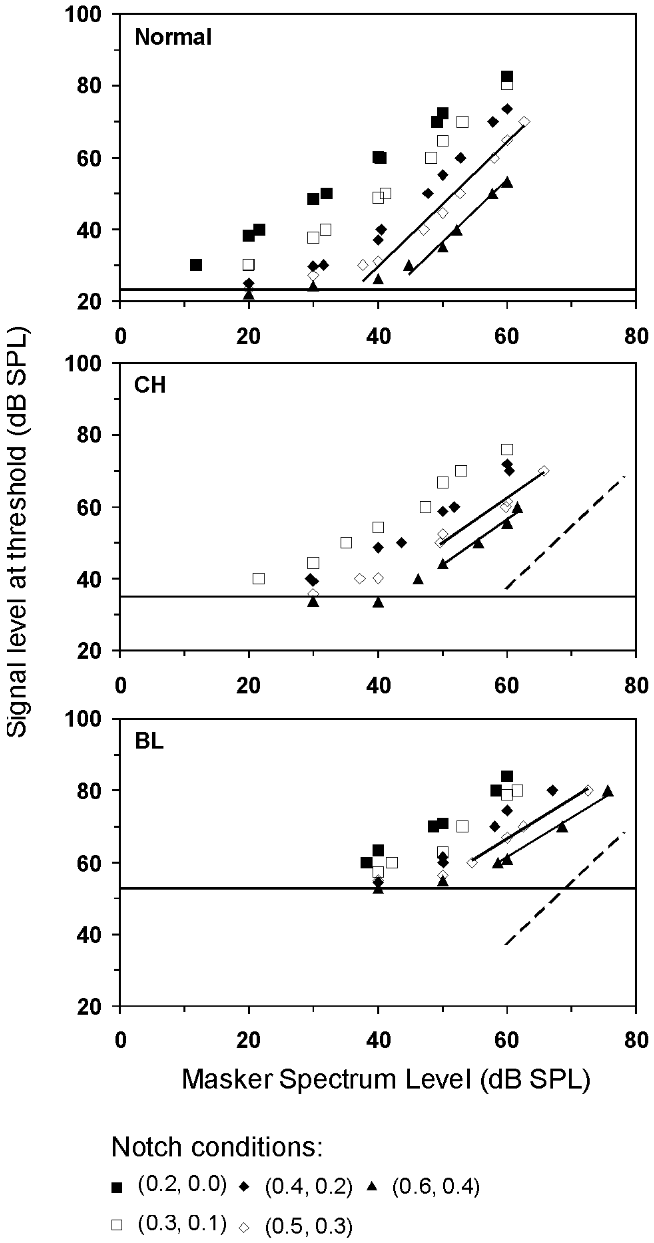

Figure 2. *Examples of growth of masking functions with data taken from the middle column of figure 1 and replotted as signal level against masker level. The solid diagonal lines show two of the three linear fits to the wide notch data [the third being for the (0.5, 0.5) notch condition not shown here]. The dashed diagonal lines in the lower two plots serve to indicate the slope of the normal hearing functions. The horizontal solid line represents the listeners' absolute threshold.*

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), the reciprocal of the slope of the GOM functions obtained as above gives an estimate of the slope of the input output (I/O) function of the underlying filtering mechanism. Table 3 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 2kHz, estimates are given at 500Hz and 1kHz from listeners CS and MF respectively who also completed the notched-noise masking tests at these frequencies where their hearing loss fell within the normal range. For comparison with these two estimates, table 3 also gives estimates at these two frequencies from the mean of two listeners data given in Baker, et al. (1998).

Subject	Frequency	Threshold, dB HL	GOM slope	I/O slope (GOM)	I/O slope (filter)
Normal Rosen et al. (1998)	2kHz	8	1.731	0.578	0.47
CH	2kHz	20	1.257	0.796	0.72
ET	2kHz	30	1.028	0.973	0.81
CS	2kHz	35	1.105	0.905	0.77
BL	2kHz	50	1.108	0.903	0.80
MF	2kHz	50	0.652	1.534	1.18
Normal Baker et al. (1998)	500Hz	9	1.284	0.779	0.55
CS	500Hz	10	1.371	0.729	0.54
Normal Baker et al. (1998)	1kHz	5	1.776	0.563	0.52
MF	1kHz	15	1.709	0.585	0.34

Table 3. *Growth of masking for the low frequency skirt of the filter derived using linear fits as described in the text. The reciprocal of the GOM function slope gives an estimate of the slope of the underlying input-output function. The final column gives the slope of the I/O function estimated from the fitted filter shapes as described later*

The estimated I/O function slope for the normal listener at 2kHz is 0.578 indicating compression approaching 2:1. However, listener CH, with as little as 20dB hearing loss, the compression is much reduced.For listeners ET, CS and BL the derived I/O functions approach linearity. The derived slopes for the listeners CS and MF at frequencies of normal hearing are very similar to those obtained from the normal hearing listeners' data from Baker, et al. (1998).

3.2 Level dependent filter shapes

Using the PolyFit procedure described by Rosen *et al.* (Rosen and Baker, 1994; Rosen, et al., 1998) the notched noise masking data were fitted with level dependent filter shapes based on the rounded exponential (ROEX) family described by Patterson (1976). Any of the fitted parameters could be allowed to be constant, a linear, or a quadratic function of stimulus level (either masker or probe level). Because we have consistently found probe-dependent models to be vastly superior to masker-dependent ones, we focus our attention on them. Each data set was fit with a set of filter shapes of varying complexity with the fitted model being simplified until the goodness of fit worsened significantly. Typically a complex model was fit to the data. One of the parameters was then removed from the model and the data re-fit. If the error in the fit worsened significantly then that parameter was deemed necessary to the model. Thus a hierarchy of models was obtained for each listeners data from which it was possible to see which parameters were important to accurately describe the data and which were relatively superfluous. The chosen "best" filter shape for each listener is shown in figure 3.

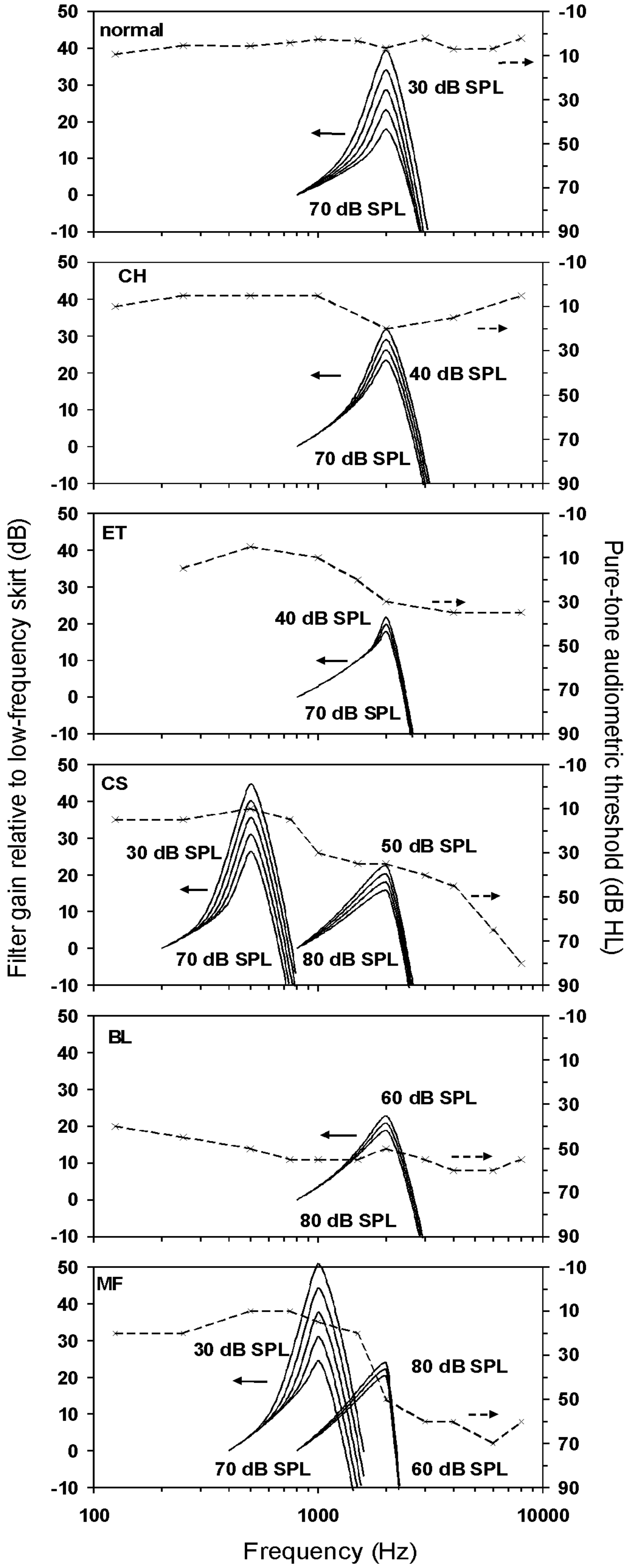

Figure 3. Filter shapes fitted to the data using the PolyFit procedure. The filter shapes are all normalised to 0dB at the lowest frequency point (0.4 x CF) and are plotted at probe-level intervals of 10dB within the ranges indicated. The dashed lines, associated with the right-hand ordinate, indicate the listeners' audiograms.

The filter shapes obtained by Rosen, et al. (1998) for averaged data from 3 normal hearing listeners are presented at the top of figure 3. For this data the high frequency side of the filter could be described by a 1 parameter, roex(p), shape in which the slope parameter (p_u) was allowed to be a cubic function of probe-tone level. The lower frequency side of the filter was better described by a 3 parameter, roex(p, w, t), shape in which the slope of the filter in the pass-band (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 pass-band and tail of the filter (w_l) was required to be a linear function of probe level, with the tip-tail ratio decreasing as probe level increased. Following the nomenclature of Rosen, et al. (1998), this model is referred to as *p1312x2x* 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 BM measurements. We thus normalise all filter shapes to have the same gain (an arbitrary 0 dB) at a frequency that is 0.4 times the filter CF.

The lower 5 plots in figure 3 show the fitted filter shapes for each of the 5 hearing-impaired (HI) listeners, along with their absolute thresholds. The filter models plotted are given in the 6th column of table 4. Clearly all of the HI listeners show a reduced change in gain at CF reflecting a more linear filter. 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. All the fitted filter shapes apart from the 2kHz filter of MF show filters that broaden and reduce in gain with increasing stimulus level. The 2kHz measurement of MF shows slight changes in the reverse direction with the filter at 60dB SPL being slightly broader than that at 80dB SPL. It must be emphasised however, that the lack of nonlinearity in the HI listeners means that the filter shapes can, in some cases be fit almost as well with filter shapes that don't change with stimulus level.

Table 4 gives the goodness of fit (as root-mean-square deviations of the data from the predictions) for 3 different filter shape models to the data of each listener. Firstly, the most complex model uses a roex(p, w, t) shape in which the p, w and t on both sides of the filter are allowed to be linear functions of probe level. Such a model has a total of 14 parameters including a constant k and absolute threshold, and of course fits the data better then the more simplified models. Table 4 also shows the rms error for the "best" models chosen, as shown in figure 3, and also for a level independent (linear) model. The last column in table 4 gives the percentage increase in rms error that occurs when fitting the data with a linear shape rather than on in which all the parameters apart from k can vary with stimulus level. As the fitted shapes are describing the same underlying sets of data, it is to be expected that the more linear the filter, the less the difference would be in the goodness of fit of the two models. For the normal hearing listeners the linear model shows a 515% increase in rms error relative to the most complex model. For the hearing-impaired 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.

Subject	Threshold (dB HL)	Number of thresholds	p2212222	m2212222	Best model	RMS error for best model	p1111111 (linear)
NH	5	158	0.81	1.47 (81%)	p1312x2x	0.86 (6%)	4.98 (515%)
CH	20	110	1.19	1.47 (24%)	p1112x1x	1.20 (1%)	2.08 (75%)
ET	30	94	0.99	1.13 (14%)	p1112x1x	1.07 (8%)	1.69 (71%)
CS	35	111	1.68	2.14 (27%)	p211x2x1	1.76 (5%)	2.45 (46%)
BL	50	96	1.16	1.24 (7%)	p1112x1x	1.25 (8%)	1.48 (28%)
MF	50	93	1.45	1.31 (-10%)	p211xxxx	1.60 (10%)	1.63 (12%)

Table 4. *Details of goodness of fit of various filter shapes to the notched-noise masking data at 2kHz. Complexity of fitted filter shapes are denoted in the form p2212222 or p1111111 for example, where p indicates that 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 cubic function (3 – 3 fitted coefficients). 'x' indicates that a parameter is not required. The order of the parameters are pl, pu, k, wl, wu, tl, tu, where 'l' indicates the lower, and 'u' the upper side of the filter. The percentages in bracket show the percentage increase in rms error relative to the p2212222 model.*

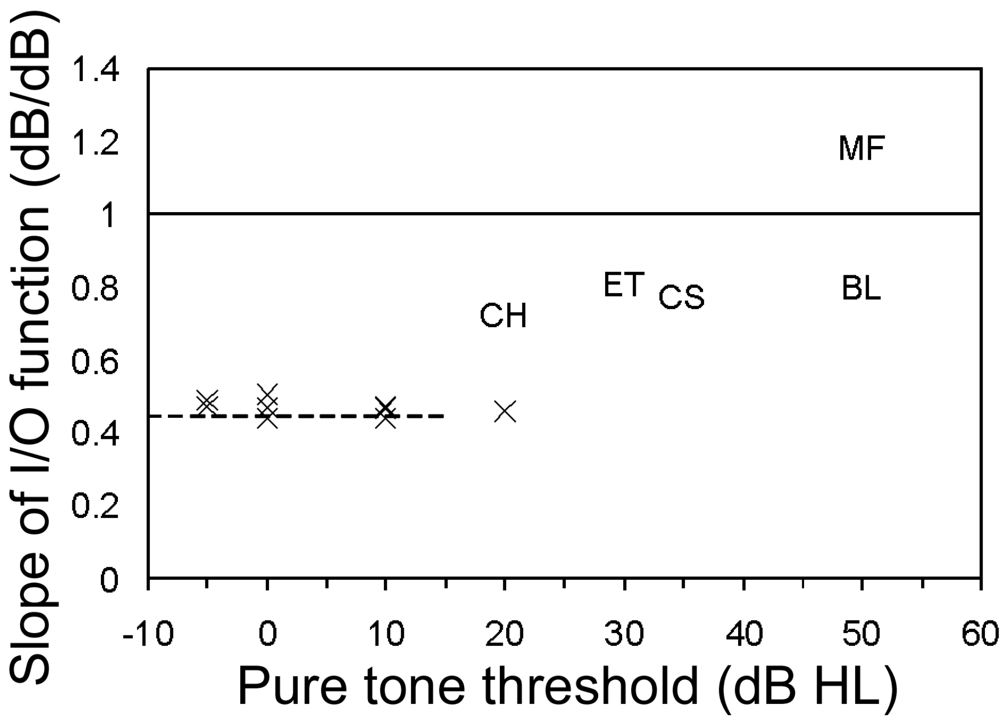

Figure 4. *Input/output function slopes derived from the filter shapes shown in figure 3 for the HI listeners. The 'x's indicate measurements from the 9 normal hearing listeners described by Rosen, et al. (1998) and the dashed line represents the "mean of three" normal hearing listeners described in the same study. A slope of value 1.0 indicates a linear filter, and a value of 0.5 indicates a compression of 2:1.*

3.3 Estimated Compression

We have already estimated the underlying basilar membrane I/O function directly from the growth of masking function. Our technique of normalising filter shapes at a frequency significantly below CF (as illustrated in figure 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 gives the estimated I/O function slope for the 5 hearing-impaired (HI) listeners, and also for the 9 normal hearing listeners measured at 2kHz by Rosen, et al. (1998). For the normal hearing listeners the estimated slope is about 0.47 giving a compression ratio just over 2:1. For the HI listeners the range is from 0.72 to 1.18 (see last column of table 3). Note that listener MF shows a value of greater than 1 indicating an *expansive* nonlinearity and reflecting the fact that the fitted filter shape narrows slightly with increasing stimulus level. It is also important to remember that the same listener shows the most "linear" filter in table 4. A final point from figure 4 is that listener CH evidences a markedly less compressive cochlea than one of the normal hearing listeners despite the fact that his absolute threshold is lower than that listener.

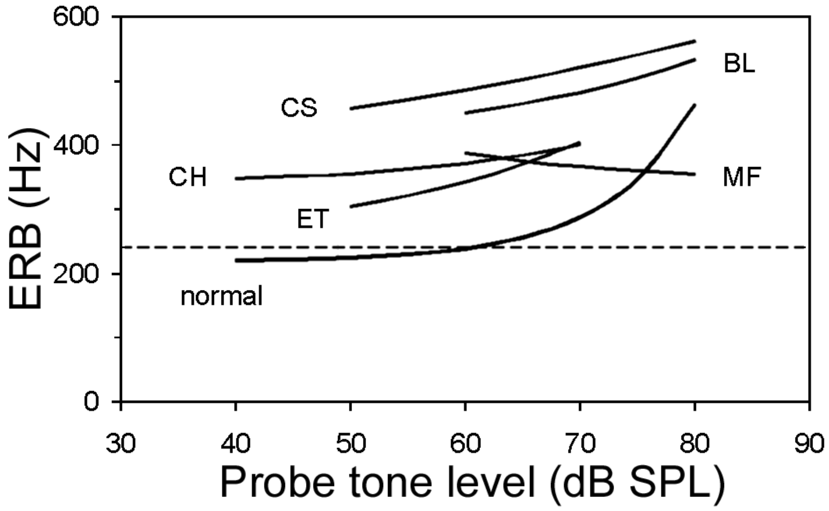

Figure 5. *Equivalent rectangular bandwidths derived from the fitted filter shapes given in figure 3. The dashed line represents the value of 241Hz from the equation given by Glasberg and Moore (1990) for a frequency of 2kHz.*

3.4 Equivalent Rectangular Bandwidth

From the fitted filter shapes it is possible to quantify selectivity in terms of the filters' bandwidths. The measure that has typically been used is the equivalent rectangular bandwidth or ERB. Figure 5 shows how the ERB, calculated from the filter models shown in figure 3, changes as a function of probe-tone level. For the normal hearing 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 to 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 hearing-impaired listeners there are three major points worth noting. Firstly, at low probe levels the ERB is always broader than the normal hearing listeners. Secondly, the ERBs measured in the impaired ears change by smaller amounts than in the normal hearing ears – the filter is more linear. Thirdly, at high levels the ERBs measured in the impaired ears are much more similar to those of the normal ears.

3.5 Filter shapes at a fixed masker level

Given the nonlinear nature of the cochlea and the manifestations of this nonlinearity in psychoacoustics, it is clear that careful consideration needs to be paid to the way in which notched-noise masking experiments are carried out. Rosen, et al. (1998) argued that fixing the probe level is more appropriate than fixing the masker spectrum level as has typically been done in the past. While such considerations are necessary for normal-hearing listeners, for listeners with hearing losses where the OHCs are virtually non-functional then it may be expected that both methods would result in the same filter shapes. Rosen, et al. (1998) argued that fixing the masker spectrum level in a notched-noise masking experiment in normal hearing listeners results in a filter which narrows as the notch widens because the filter output decreases resulting in artificially narrow filter shapes. For hearing-impaired listeners, however, such an effect will be much reduced, and for total OHC loss (resulting in linear cochlear mechanics) the same filter shapes should result from both fixed-probe and fixed-masker experiments.

At relatively high stimulus levels in normal hearing listeners (where the influence of OHCs is small) it may be expected that selectivity will be similar to that observed in impaired ears. A model of selectivity that includes an accurate depiction of the nature of the nonlinearity should allow realistic comparisons of filtering in NH and HI listeners. A “high-level” filter shape from such a model should be similar to the actual filter shapes obtained from HI listeners. Furthermore, such a model should also be able to predict the pattern of masked thresholds in NH listeners regardless of whether a fixed-probe or a fixed-masker paradigm was used. If the model is accurate, it should also be able to predict the “artificially-narrow” filter shapes measured using a fixed-masker paradigm in normal hearing listeners.

Qualitative support for both these arguments can be obtained by comparing filter shapes from the model to high-level fixed-masker experiments in both NH and HI listeners. Figure 6 shows such a comparison for a fixed masker spectrum level of 50 dB SPL (the highest used for both NH and HI listeners). For such a masker level, the probe level in the no-notch condition will be approximately 70 dB SPL. Making the assumptions that (a) the hearing-impaired (HI) listeners have little residual nonlinearity and (b) the ears of normal hearing (NH) listeners are approaching a region of linear operation for a probe level of 70 dB SPL, then filter shapes derived from a fixed-probe level of 70 dB SPL in NH listeners should be similar to filter shapes derived from fixed-masker spectrum levels of 50 dB SPL in HI listeners. Such a comparison is shown in the lower plot of figure 6. The shapes from the HI listeners are derived from individual sets of data at the single spectrum level of 50 dB SPL fit with a roex(p) shape (a more complex filter shape did not improve the goodness of fit). The filter shape for the NH listeners is obtained from the PolyFit derived model shown at the top of figure 3 with a probe level of 70 dB SPL

If this probe-dependent description of NH filtering is correct then the model should also be able describe the hypothesised narrowing of NH filters in a fixed-masker experiment. The symbols in the upper plot of figure 6 show the fitted filter shapes for a fixed masker spectrum level of 50 dB SPL in four normal hearing listeners from Rosen, et al. (1998) (those measured with the largest set of notch-widths). For model comparisons with this data, the filter shape model shown in the top of figure 3 was used to predict masked thresholds for the complete set of notch widths at this fixed masker level. These predicted threshold were then fitted with a roex(p) filter shape for comparison with the empirical data (a shape more complex than a roex(p) was not required to describe the data). As expected, the model predictions closely match the filter shapes from the empirical data. In summary, the same "output" dependent filter model derived from NH listeners is, at least qualitatively, able to predict the filtering in both NH and HI listeners measured using a fixed-masker level paradigm.

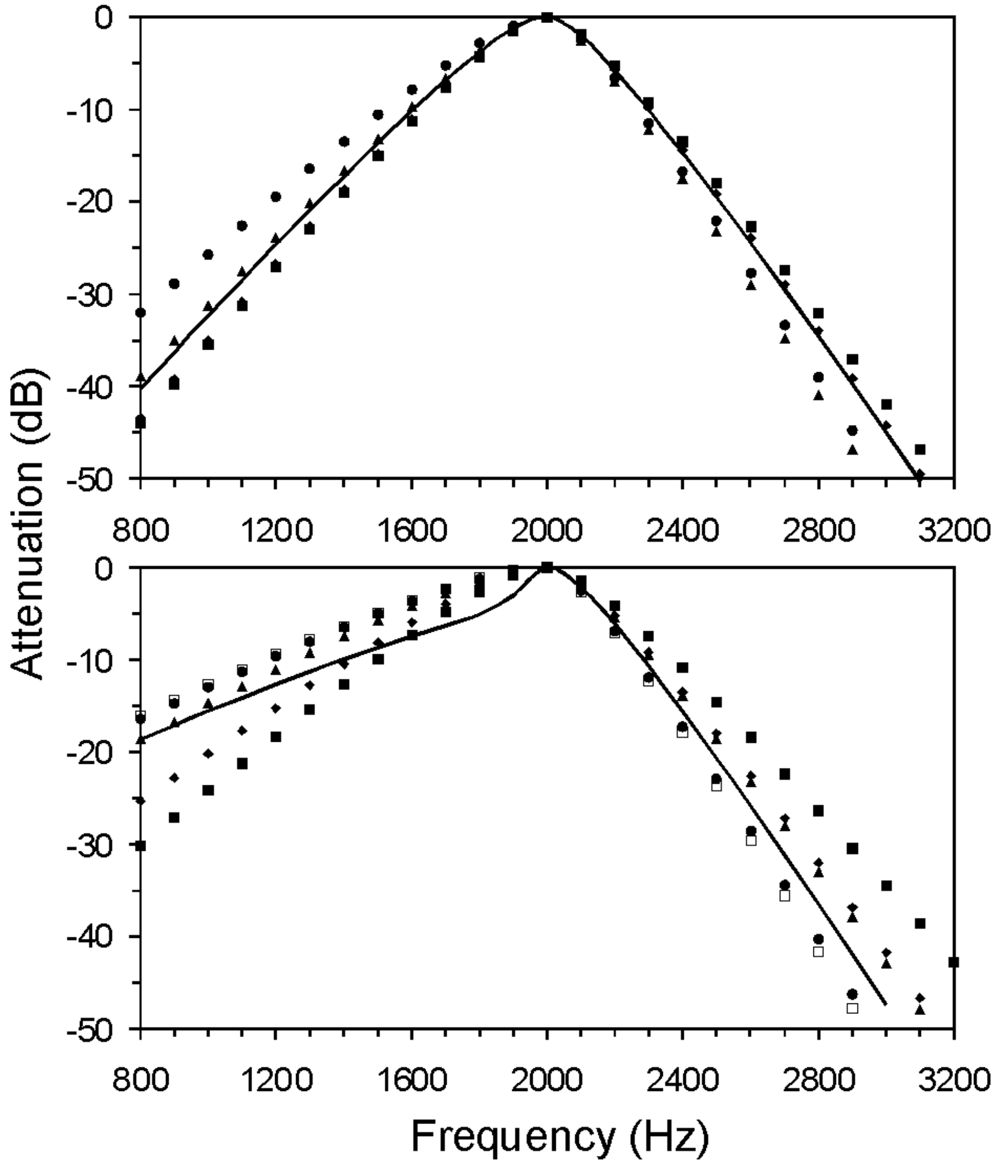

Figure 6. *Filter shapes fitted to notched-noise masking data obtained at a fixed masker spectrum level of 50dB SPL. Filters from 4 normal hearing listeners (symbols) are given in the upper plot and from the 5 impaired listeners given in the lower plot. The solid lines show predictions from a model of filtering in normal hearing listeners (see text for details).*

4 Discussion

The emphasis of this study was to characterise the way in which auditory filters in hearing impaired 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 (see Moore, 1998) for review), such studies have nearly all used a fixed-masker level paradigm. In characterising 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 normal hearing listeners the choice of paradigm in the normal hearing 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 that the filter shape appears to be controlled by its own output. The likely mediators of such a feedback mechanism are the OHCs which provide a “fast mechanical positive feedback” to the basilar membrane (Patuzzi, 1996). Given such a feedback mechanism, Rosen, et al. (1998) argued that in a fixed-masker experiment the output of the auditory filter would decrease as the notch in the masker was widened and thus the filter would become narrower. To overcome this problem Rosen, et al. (1998) used both fixed maskers and fixed probes and described all the data in terms of a single filter whose shape broadened as a function of probe-tone level. Here we have used the same procedure to describe filter shapes from listeners with mild to moderate sensorineural hearing loss in a way that allows direct comparison with the normal hearing data from Rosen, et al. (1998).

We have been able to estimate the degree of compression from normal and impaired ears, firstly using the growth of masking functions for wide notches in a way similar to Oxenham and Plack (1997), and secondly indirectly from the change in gain at the CF of the fitted filter shapes. As expected, both these produce similar estimates of the degree of BM compression. The compression ratio derived from the fitted filter shapes for the normal hearing listeners is approximately 2:1. Such a value is typical for estimates using simultaneous masking (eg Stelmachowicz, et al., 1987), while in forward masking experiments compression ratios are more typically of the order of 5:1 (Oxenham and Plack, 1997). This latter figure is more comparable to the compression ratios measured directly from the basilar membrane. It is likely that suppression plays an important role in “linearising” the ear in simultaneous masking experiments but not in forward masking. Such suppression was demonstrated physiologically by Ruggero, et al. (1992) where the response to a CF tone was reduced and its growth linearised by the presence of a low-frequency suppressor tone. Clearly, repeating our experiments in normal hearing listeners using forward masking may lead to measurements of compression that are more directly comparable to those of Oxenham and Plack (1997). 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 simultaneous. Moore and Glasberg (1986) showed that the difference in selectivity between normal and impaired ears in forward masking is greater than in simultaneous, suggesting that suppression reduces the enhancement of selectivity in simultaneous masking in the NH listeners. Using the PolyFit procedure to characterise changes in selectivity across level using forward masking in NH listeners should reveal a greater degree of nonlinearity than is observed in simultaneous masking.

For the hearing impaired listeners the growth of masking functions are much more parallel across the different notch widths indicating a smaller change in filter shape across level. The estimates of compression from these GOM functions, and also from the fitted filter shapes indicate a much-reduced degree of compression in these impaired ears. One thing that is particularly remarkable is the loss of nonlinearity present in listener CH. His threshold at 2kHz is 20dB HL, yet the degree of compression is only about 1.26:1. Listener MB from Rosen, et al. (1998) had a similar threshold, but showed a compression ratio of just over 2:1 (right most ‘x’ in figure 6). It is tempting to speculate that these differences reflect a mainly OHC loss in listener CH and a loss not involving OHCs in MB.

This reduced nonlinearity is also evident in the filter shapes fitted to the HI data. Comparison of the filter shapes from figure 3 show that for the impaired listeners the tip-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. Also while the range of gains at CF is much less in the HI than in the NH listeners, the order is the same in all but MF. That is, the low-level filters are the most sharply tuned, with highest gain at the CF.

Subject MF shows the reverse of this, the higher level filter shapes being slightly sharper than the lower level shapes. However this in itself may be of little significance as that same data can be fit almost as well by a linear filter shape. However, there are two other points worth noting from MF’s filter shapes in figure 3. Firstly, the high-frequency side of the filter at 2kHz appears to be abnormally steep.Secondly, the degree of compression in the filter measured at 1kHz in this listener seems abnormally large (26.4dB change in gain for 40dB change in probe level giving a compression ratio of 2.9:1). One possibility is that the filtering of this listener is unduly influenced by the steepness of his audiogram (20dB HL at 1.5kHz and 50dB HL at 2kHz).

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/tail ratio becomes small). A similar effect is observed with PTCs. Nelson, et al. (1990) showed that for forward masked PTCs the Q_{10dB} remains relatively constant at low levels and only begins to decrease at high stimulus levels where the tip/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 comparable.

In support of our previous claims that using a fixed masker in a notched-noise masking experiment overestimates the filter sharpness (since the filter’s output decreases as the notch widens, Rosen, et al., 1998) we have also compared filters obtained with a fixed masker level in both NH and HI listeners. As has been shown previously, the NH filters are significantly sharper than HI ones. However, if our assumptions about the filter shape being determined by its own output are correct, then it may be expected that a probe level dependent filter shape measured in normal hearing listeners should be able to describe fixed masker data from both NH and HI listeners. Given the assumptions described in section E, this appears to be the case, thus adding further evidence to the claim that auditory filters in normal hearing listeners are more likely to be controlled by their own output rather than by some aspect of the stimulus at filter input.

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