Additivity of masking in normally hearing and hearing-impaired subjects

Andrew J. Oxenham and Brian C. J. Moore

Department of Experimental Psychology, University of Cambridge, Downing Street, Cambridge CB2 3EB, England

(Received 25 October 1994; revised 13 April 1995; accepted 28 April 1995)

The effects of combining two equally effective maskers were studied in normally hearing and elderly hearing-impaired subjects. The additivity of nonsimultaneous masking was investigated by measuring thresholds for a brief 4-kHz signal in the presence of a broadband-noise forward masker, a backward masker, and a combination of both. For the normally hearing subjects, combining two equally effective nonsimultaneous maskers resulted in up to a 15-dB greater increase in thresholds than the 3 dB predicted by an energy-summation model ("excess masking"). However, the hearing-impaired subjects showed little or no excess masking. The differences between the two groups is consistent with a theory linking excess masking to the compressive transfer function measured on the basilar membrane (BM). In the hearing-impaired subjects the transfer function is more linear, accounting for the lack of excess masking. The additivity of simultaneous masking was investigated by measuring thresholds for a 100-ms 4-kHz signal in the presence of either a 400-ms broadband-noise masker or a 400 ms sinusoidal masker at the same frequency as the signal, and then combining two equally effective maskers, a noise and a tone. The maximum amount of excess masking (3 to 4 dB) was similar across the two groups of subjects, consistent with an explanation based on the use of different detection cues for the tonal and noise maskers. It is argued that, while peripheral compression may underlie excess masking for pairs of nonsimultaneous maskers, it is unlikely that in simultaneous masking, where the maskers are close in frequency to the signal, the two maskers are compressed individually before their effects are combined. It is further suggested that BM nonlinearity may underlie the effects of the upward spread of masking and the nonlinear growth of forward masking, as well as accounting for the additivity of simultaneous masking when the masker frequencies are well below that of the signal. © 1995 Acoustical Society of America.

PACS numbers: 43.66.Dc, 43.66.Ba, 43.66.Sr, 43.66.Mk

INTRODUCTION

The question of how the effects of two or more maskers combine has been an area of interest for some time. In normally hearing subjects, the additivity of nonsimultaneous masking (Elliott, 1969; Wilson and Carhart, 1971; Robinson and Pollack, 1973; Penner, 1980; Cokely and Humes, 1993; Oxenham and Moore, 1994) and the additivity of simultaneous masking (Green, 1967; Zwicker and Herla, 1975; Lutfi, 1983; Moore, 1985; Humes et al., 1992b) have been systematically investigated. The additivity of masking in hearing-impaired subjects has received less attention. Jesteadt (1983) compared the performance of normally hearing and hearing-impaired subjects using a sinusoidal forward masker, at the same frequency as the signal, and a broadband-noise simultaneous masker. He found that combining these two maskers led to a smaller increase in thresholds for the hearing-impaired than for the normally hearing subjects. Lutfi (1987) compared the performance of normally hearing and hearing-impaired subjects using two simultaneous narrow-band noise maskers centered on either side of the signal frequency. Lutfi also found large differences in performance between the two groups and proposed some possible explanations which are discussed in Sec. II of this paper. We know of no studies which investigate the additivity of purely nonsimultaneous maskers in the hearing-impaired.

Some theories of the additivity of masking have dealt with the effects of nonsimultaneous maskers (Penner, 1980; Penner and Shiffrin, 1980; Oxenham and More, 1994), another has treated simultaneous masking effects (Lutfi, 1983), while another attempts to provide a method of accounting for both situations (Humes and Jesteadt, 1989). All these theories rely on the assumption that the individual stimuli (maskers and signal) are processed independently before their effects are combined. It is this assumption, together with the possible underlying physiological mechanisms, which we investigate in this paper. We first address issues relating to nonsimultaneous masking, and report results from an experiment which investigates whether the additivity of nonsimultaneous masking may be described in terms of peripheral auditory nonlinearities by comparing the performance of three normally hearing subjects with that of three subjects with cochlear hearing impairment.

I. NONSIMULTANEOUS MASKING

The most common way to investigate the additivity of nonsimultaneous masking has been by studying the effects of combining a forward and a backward masker. Consider a forward masker and a backward masker which individually produce an equal amount of masking for a given signal. An energy-summation model of masking predicts a 3-dB in-
crease in signal threshold when the two maskers are combined (Green and Swets, 1966). However, in most situations additional, or "excess," masking of 10–15 dB is observed (Elliott, 1969; Wilson and Carhart, 1971; Robinson and Pollack, 1973; Penner, 1980; Cokely and Humes, 1993; Oxenham and Moore, 1994). Penner (1980) proposed that excess masking could be accounted for by assuming that stimuli are subjected to a compressive nonlinearity before their effects are combined within a linear temporal integrator. Rather than attempting to define the weighting function (or window) associated with the integrator, Penner used the signal threshold in the presence of a single masker as a measure of the masker’s internal effect and derived an expression for the compressive nonlinearity by combining equally effective pairs of maskers over a range of levels. The observation that the amount of excess masking increased with increasing level led Penner to propose a nonlinearity more compressive than a simple power law (where stimulus representations are raised to a power less than unity). Humes and Jesteadt (1989) have since shown that it is possible to account for the level dependency of excess masking with a simple power law, if one includes a constant term to account for absolute threshold. The transform used by Humes and Jesteadt has the general form

$$i_x = (I_{MTX})^p - (I_{QT})^p,$$

where $i_x$ represents the internal effect produced by masker $X$ at the signal frequency, $I_{MTX}$ is the intensity of the signal at masked threshold for masker $X$, $I_{QT}$ is the constant term representing signal intensity at threshold in quiet, and $p$ is the exponent which determines the amount of compression, with a value ranging from 0.0 to 1.0. If two maskers are presented, it is assumed that they are compressed individually before their effects are combined.

As Penner and Shiffrin (1980) showed, it is possible to use the signal level at threshold as a measure of the masker’s internal effect as long as the stimuli do not overlap. However, without a derivation of the temporal weighting function it is only possible to predict masked thresholds for combined maskers if the masked thresholds in the presence of the individual maskers are known. In an attempt to derive a suitable weighting function, Oxenham and Moore (1994) presented a model in which the stimuli are first filtered using a (linear) gammatone filterbank (de Boer and Kruidenier, 1990) and, for simplicity, only the output from the filter centered at the signal frequency is processed further. The output of the filter is rectified and subjected to a compressive nonlinearity before being integrated. A weighting function for the integrator, or temporal window, was derived using the results from a number of forward- and backward-masking conditions, with masker–signal intervals ranging from 0 to 100 ms, and from combined-masking (forward and backward) conditions with the signal placed both symmetrically and asymmetrically in time between the forward and backward masker. Four free parameters defined the exact shape of the temporal window, three determining the decay of forward masking (two time constants and a weighting function between them) and one determining the decay of backward masking. This procedure is similar to that used in other temporal window studies (Moore et al., 1988; Plack and Moore, 1990). Absolute threshold was accounted for by assuming a small, constant output from the temporal window. This constant has the same effect as the constant used by Humes and Jesteadt ($I_{QT}$) and so allowed the model to account for the increase in excess masking with level while using a simple power-law nonlinearity. The best fits to the data were achieved by raising the stimulus intensity to the power of approximately 0.3 (i.e., $I_p$, where $p=0.3$).

By combining compressed representations of the stimuli within a temporal window, the model proposed by Oxenham and Moore (1994) was able to account for the additivity of nonsimultaneous maskers, as well as accounting for the decay of forward and backward masking as a function of masker–signal interval. The model presented by Humes et al. (1988) always applies compression to the amount of masking, rather than to representations of the stimuli. This has the advantage that, when the stimuli do not overlap in time, it is not necessary to derive a temporal weighting function in order to predict the amount of excess masking if signal thresholds in the presence of the single maskers are known. The disadvantage of compressing the amount of masking is that excess masking is predicted in all masking situations, including ones where maskers spectrally and temporally overlap, meaning that certain masking conditions must be excluded from the model in order to remain consistent with experimental data. In cases where the stimuli themselves are compressed (e.g., Penner, 1980; Penner and Shiffrin, 1980; Oxenham and Moore, 1994), excess masking is predicted only when the stimuli are compressed independently, i.e., when they do not overlap in time. This is consistent with experimental data using broadband-noise maskers (Cokely and Humes, 1993); other configurations of simultaneous maskers will be discussed in Secs. II and III. Another advantage of applying compression to the stimuli and deriving an appropriate temporal window is that the scheme can be used within a more general model of auditory processing.

Oxenham and Moore (1994) suggested a link between the compression used in their model, and the compressive transfer characteristic observed on the basilar membrane (BM) in physiological studies (Rhode, 1971; Johnstone et al., 1986; Yates, 1990; Ruggiero, 1992). The compressive region of the BM’s transfer function extends from input levels of about 30 to about 90 dB SPL. At very low and very high levels the response is more linear. Also, the compressive function is observed mainly in response to tones around the characteristic frequency (CF) of the measurement point; at frequencies well below the CF, the response is nearly linear for all input levels. Interestingly, the amount of compression observed for on-frequency tones at mid levels is about 5:1 in dB (Yates, 1990). This corresponds to a power-law exponent of 0.2, which is quite close to the exponents derived by Humes and Jesteadt (1989) and Oxenham and Moore (1994), as described above.

Cochlear injury results in reduced sensitivity, broader frequency tuning, and a near-linear transfer function as measured on the BM (Ruggiero and Rich, 1991; Ruggiero et al., 1993). The first two of these three aspects have well-documented psychoacoustic parallels in human cochlear
hearing loss (Glasberg and Moore, 1985). The effects of a more linear BM transfer function are less well understood, although the effect of loudness recruitment (Fowler, 1936) may be mediated by BM linearization (Ytes, 1990; Glasberg and Moore, 1992). If excess masking is indeed mediated by peripheral nonlinearities (in particular a compressive BM function), then one might expect subjects with sensorineural hearing loss to exhibit much less excess masking than normally hearing subjects. The first experiment tests this hypothesis.

A. Conditions and stimuli

Experiment 1 compared the amount of excess masking observed in three normally hearing subjects with that observed in three elderly subjects with cochlear hearing impairment. The signal was a 4-kHz sinusoid (Farnell DSG1) with a 2-ms steady-state portion, gated on and off with 2-ms raised-cosine ramps. The masker (both forward and backward) was a low-pass filtered white noise with a cutoff frequency of 9 kHz (Kemo VBF/3/03 filter, 96-dB/oct slope). The masker had a steady-state duration of 200 ms and was gated with 1-ms raised-cosine ramps.

Initially, the threshold of the signal in quiet was measured for each subject. Then levels of forward and backward maskers were determined which individually produced 5, 10, 15, 20, and 25 dB of masking. At first, forward masker levels were determined for masker-signal intervals of 10 and 25 ms (defined as the interval between the zero-voltage points of the envelope of the electrical signal) and backward masker levels were determined for signal-masker intervals of 1 and 5 ms. After these data had been collected, a further forward-backward condition, using a masker-signal interval of 5 ms, was included. However, in this condition, due to time shortages, masker levels producing 5 dB of masking were not measured, except for subject JC. Once thresholds for the individual forward and backward maskers had been determined, signal thresholds were measured in the presence of pairs of equally effective maskers, on forward and one backward.

A trial consisted of two observation intervals, marked by lights, separated by a silent interval of 500 ms. The signal was presented randomly in either the first or second interval. Stimulus timing was controlled by a Texas Instruments 990/4 computer system. Two analog multipliers (AD 534L) in series were used as gates, giving an on-off ratio exceeding 100 dB. Stimulus levels were varied using Caryaebis model D programmable attenuators. The stimuli were passed through a final manual attenuator to one earphone of a Sennheiser HD414 headset. For the normally hearing subjects, the stimuli were presented to the left ear. For the hearing-impaired subjects the stimuli were presented to the ear with the lower audiometric threshold at 4 kHz. For our three subjects this was also the left ear.

B. Procedure

Thresholds were determined using a two-alternative forced-choice paradigm with a three-down one-up (three-up one-down in the case of the varying masker) adaptive procedure that estimates the 79.4% correct point on the psychoacoustic function (Levitt, 1971). In the initial, single-masker experiments, the masker level was adaptively varied to produce a given amount of masking. In the subsequent combined-masker conditions, the signal level was varied. The initial step size was 5 dB, which was reduced to 2 dB after the first four reversals. A run was terminated after a total of 12 reversals and the threshold was defined as the mean of the levels at the last 8 reversals.

In cases where the masker level was varied, a run was automatically terminated if the masker level reached 55-dB spectrum level (≈75 dB SPL overall) for the normally hearing subjects, or 65-dB spectrum level (105 dB SPL) for the hearing-impaired subjects. In these cases measurement of that data point was abandoned.

Each threshold reported is the mean of at least three estimates. If after three estimates the standard deviation across estimates was greater than 2.5 dB, a further three estimates were obtained and the mean of all six was recorded. Subjects were tested individually in a double-walled sound-attenuating chamber.

C. Subjects

The three normally hearing subjects were aged between 24 and 30 yr and had thresholds no greater than 15 dB HL at octave frequencies between 250 Hz and 8 kHz. One was the author AO; the others were paid for their participation. Subjects AO and ST had previous experience in psychoacoustic tasks. The three hearing-impaired subjects were aged between 73 and 77 yr and had bilateral sensorineural hearing loss. Their audiometric thresholds are given in Table I. All three hearing-impaired subjects had previous experience in psychoacoustic tasks, and all were paid for their participation.

All subjects were given at least 8 h of practice, divided between the different conditions, before the data were re-

### Table I. Absolute thresholds, in dB HL, for the hearing-impaired subjects. The left ear was used as the test ear for all subjects.

<table>
<thead>
<tr>
<th>Frequency (kHz)</th>
<th>Subject</th>
<th>0.25</th>
<th>0.5</th>
<th>1</th>
<th>2</th>
<th>4</th>
<th>8</th>
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<tr>
<td></td>
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<td></td>
<td></td>
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<td></td>
</tr>
<tr>
<td>Left</td>
<td>40</td>
<td>40</td>
<td>45</td>
<td>40</td>
<td>65</td>
<td>70</td>
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<tr>
<td>Right</td>
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<td>35</td>
<td>40</td>
<td>55</td>
<td>75</td>
<td>70</td>
<td></td>
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<td>30</td>
<td>45</td>
<td>45</td>
<td>55</td>
<td>90</td>
<td></td>
</tr>
<tr>
<td>Right</td>
<td>10</td>
<td>10</td>
<td>50</td>
<td>55</td>
<td>70</td>
<td>85</td>
<td></td>
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<td>50</td>
<td>65</td>
<td>65</td>
<td>65</td>
<td></td>
</tr>
<tr>
<td>Right</td>
<td>40</td>
<td>55</td>
<td>65</td>
<td>60</td>
<td>70</td>
<td>65</td>
<td></td>
</tr>
</tbody>
</table>

### Table II. Thresholds in quiet for the signals used in (a) experiment 1 and (b) experiment 2 in dB SPL.

<table>
<thead>
<tr>
<th>Subject</th>
<th>MG</th>
<th>AO</th>
<th>ST</th>
<th>JC</th>
<th>AD</th>
<th>GW</th>
</tr>
</thead>
<tbody>
<tr>
<td>(a)</td>
<td>21</td>
<td>19</td>
<td>15</td>
<td>71</td>
<td>67</td>
<td>71</td>
</tr>
<tr>
<td>(b)</td>
<td>7</td>
<td>11</td>
<td>5</td>
<td>62</td>
<td>60</td>
<td>67</td>
</tr>
</tbody>
</table>
corded. No consistent improvements were found during the course of the experiment.

D. Results

Thresholds for the signal in quiet are shown for each subject in Table II. Results from the single-masker conditions are shown in Figs. 1 and 2 for the normally hearing and hearing-impaired subjects, respectively. As mentioned above, the signal level was fixed in each condition to be a given amount above threshold in quiet for each subject individually, and the masker level was varied. Therefore, masker spectrum level is plotted on the ordinate, rather than the more usual signal level. The boxes within each figure show data from the different conditions. For instance, 5F denotes forward masking with a 5-ms masker–signal interval and 1B denotes backward masking with a 1-ms signal–masker interval. The different symbols represent data from different subjects. The short-dashed lines in each box represent linear growth of masking and the long-dashed lines show nonlinear masker-to-signal growth with a ratio (in dB) of 2:1.

FIG. 1. Masked thresholds for the normally hearing subjects in the presence of single forward and backward maskers. The numbers in the panels refer to the masker–signal interval (ms); the letters F and B denote forward and backward masking, respectively. Short-dashed lines represent linear growth of masking and the long-dashed lines show nonlinear masker-to-signal growth with a ratio (in dB) of 2:1.

FIG. 2. As in Fig. 1, but for the hearing-impaired subjects.
growth of masking and the long-dashed lines denote nonlinear growth where masker level grows more rapidly than signal level, with a ratio (in dB) of 2:1.

Consider first the data from the normally hearing subjects (Fig. 1). Consistent with the literature (Jesteadt et al., 1982; Moore and Glasberg, 1983), the forward-masking data show a nonlinear growth of masking which becomes more pronounced with increasing masker–signal interval. The form of the data is very similar across subjects. Data from the two backward-masking conditions are more variable, both within and across subject. Many more thresholds required six, rather than three, runs and the error bars (denoting ± one standard deviation) are generally larger than in the forward-masking conditions. Subject AO (circles) requires markedly higher masker levels for a given signal threshold than subjects MG and ST in the 1B condition and shows no backward masking in the 5B condition, hence the absence of circles in the panel labeled 5B. Subjects MG and ST show increasingly nonlinear growth of backward masking with increasing signal–masker interval. Consistent with the literature, the forward masker was more effective than the backward masker for a given masker–signal interval (compare conditions 5F and 5B) for all normally hearing subjects. This implies an asymmetric temporal window shape, with a steeper slope on the side accounting for backward masking conditions. Subject JC shows the forward masker and backward masker to be approximately equally effective at a 5-ms interval, implying a symmetric temporal window shape, and the data from GW indicate an asymmetry in the opposite direction, as the backward masker is more effective (lower masker level required) than the forward masker. Also, the decay of both forward and backward masking is less steep in the hearing-impaired data. For instance, comparing data points for a given subject and signal level, it can be seen that in many cases the masker level does not increase much and in some cases actually decreases when going from 5F to 10F, although the decreases were not significant (one-tailed t-test: p > 0.05 in each case). A more gradual decay of forward masking in hearing-impaired subjects is consistent with the literature (Glasberg et al., 1987; Nelson and Freyman, 1987; Nelson and Pavlov, 1989). However, to our knowledge, no comparable data exist for backward masking. These and previous results may reflect some abnormalities in the temporal processing of hearing-impaired subjects, and hence an abnormal temporal window shape. Another possible explanation for the shallow slope associated with the decay of forward and backward masking in hearing-impaired subjects is presented in Sec. III B.

The three forward- and two backward-masking conditions resulted in six combined conditions. Here, pairs of forward and backward maskers, which individually produced the same amount of masking, were combined and the signal level was varied to determine thresholds. Results from the combined conditions are plotted in Figs. 3 and 4 for the normally hearing and hearing-impaired groups, respectively. Signal threshold in the presence of two maskers is plotted against signal threshold in the presence of one masker. The solid lines represent no change in threshold, and the dashed lines show the predictions of an energy-summation model, i.e., a 3-dB increase in threshold. The amount of excess masking is therefore the amount by which the data lie above the dashed line. At lower signal levels, neither group shows much excess masking, and indeed most data points from the hearing-impaired group lie below the 3-dB line. At higher signal levels, the results from the two groups diverge strongly. Figure 3 shows that for the normally hearing group substantial excess masking is seen in all six conditions and for all subjects. These results are consistent with those from previous studies (e.g., Penner, 1980). In contrast, Fig. 4 shows that very little excess masking is observed in the hearing-impaired group, even at the highest levels. The form of the data within each group is similar across the different conditions, although the excess masking for the normally hearing subjects grows more steeply with increasing single-masker threshold in condition 25F, 5B than in the other conditions. The fact that the form generally remains similar across conditions is consistent with other such data (e.g., Penner, 1980; Penner and Shiffrin, 1980).

For comparison, the mean data from each of the two groups, pooled across subject and condition, are plotted in Fig. 5. The long-dashed lines are functions fitted to the data, as explained below in Sec. I E. The mean data of the normally hearing group lie above those of the hearing-impaired group at all levels. At 20 dB SL, the normally hearing group shows 9 dB of excess masking compared with less than 1 dB in the hearing-impaired group. This implies that, in the presence of two equally effective maskers, hearing-impaired subjects perform relatively better than subjects with normal hearing, especially at higher sensation levels.

E. Discussion

The results from experiment I are consistent with the hypothesis, explained above, that excess masking is mediated by the compressive transfer function of the BM and that, in normally hearing subjects, compressed representations of the stimuli are combined linearly at a higher processing stage. Subjects with cochlear hearing loss show little or no excess masking, consistent with physiological findings that damage to the cochlea results in a more linear transfer function, as measured on the BM (Ruggero and Rich, 1991).

Data from the two groups were fitted to the “modified power law” (MPL) function, described by Humes and Jesteadt (1989) and summarized in Eq. (1) above. This func-
tion gives identical predictions to the model of Oxenham and Moore (1994), provided the stimuli do not overlap in the auditory periphery, and has the advantage of not requiring the calculation of a temporal window shape. Simulations of auditory filtering using a gammatone filter (de Boer and Krüdenier, 1990) centered at 4 kHz, with an equivalent rectangular bandwidth (ERB) of 456 Hz (Glasberg and Moore, 1990), showed that even in the 1B condition (backward masking with a 1-ms signal-masker interval) only 0.3% of the signal energy overlapped with the masker, resulting in no significant calculation errors.

The equation used to derive the amount of masking (in dB) in the presence of two equally effective maskers can be simplified to

$$X_2 = 10 \log \left[ 2 \left( 10^{X/n_0} \right)^p - 1 \right]^{1/p},$$

where $X$ is the amount of masking (in dB) in the presence of a single masker and $p$ is the free parameter. Excess masking
is predicted if $p < 1$; if $p = 1$, energy summation is predicted. The constant $(-1)$ accounts for absolute threshold $-(I_{0T})^p$ in Eq. (1)] and allows an initial increase in excess masking with increasing level. This function was used to fit the data in various ways. Functions were fitted to the individual data, to the data pooled across subject within each condition and group, and to the data pooled across condition within subject and group, using a least-squares criterion. The best-fitting values of the exponent $p$ are shown in Table III. Consider first the functions from the data pooled across subject (rows four and eight for the normally hearing and hearing-impaired subjects, respectively). Within each group, the values of $p$ across the different conditions are rather similar except for the 25F, 5B condition in the normally hearing group where $p = 0.06$. This reflects the steep growth of combined masking in that condition, as shown in Fig. 3. Considering the functions for each subject pooled across conditions (final column of Table III), it can be seen that the data from subject AO are best fitted with a more compressive function than those for the two other normally hearing subjects. This indicates generally greater excess masking for this subject, as can be seen from the circles in Fig. 4. The large difference in the value of $p$ also reflects the sensitivity of the function at relatively low sensation levels when $p$ is much less than unity, due to the interaction between the exponent and the threshold constant. For instance, at a single-masker threshold of 10 dB SL, a 3-dB change in the two-masker threshold from 16.8 to 19.8 dB results in a change in the value of $p$ from 0.2 to 0.01, a factor of 20. Similarly, for $p > 1$, $p$ takes on values from 1 to infinity within a 3-dB range, between the two lines shown in Figs. 3 and 4.

The best-fitting functions to the data pooled across subject and condition, are shown in Fig. 5. The best-fitting value of $p$ for the hearing-impaired group was 1.4. Although this value is somewhat greater than unity, the fit to the data was not significantly better than when $p = 1$ [$F(47,47) = 1.15$, $p > 0.25$]. Thus the data from the hearing-impaired group can be well described using a linear function. The data from the normally hearing group were best fitted by a compressive nonlinearity of $p = 0.2$. The fit here is significantly better than when $p = 1$ [$F(62,62) = 3.25$, $p < 0.001$]. This value of $p$ also corresponds well with previous estimates of the nonlinearity in nonsimultaneous-masking experiments (Humes and Jesteadt, 1989; Cokely and Humes, 1993; Oxenham and Moore, 1994). Given the sensitivity of the function and the variability of thresholds, especially in backward-masking conditions, it would appear wise not to place undue weight on the exact value of $p$. However, as mentioned above, 0.2 is also the exponent found by Yates (1990) in his study of BM nonlinearity.

The finding that the hearing-impaired data are well modeled with a linear function, while the data from normally hearing subjects were best fitted with a strongly compressive function, has some implications for theories describing data from hearing-impaired subjects. Some theories of masking and loudness coding assume the same compressive nonlinearity when accounting for data from both normally hearing and hearing-impaired subjects (Humes et al., 1988, 1992a; Humes and Jesteadt, 1991). Such models account for threshold elevation and loudness recruitment in the hearing impaired solely by assuming an increase in the internal noise, rather than a decrease in sensitivity and a reduction in nonlinearity. Given that the data from the hearing-impaired group could only be modeled by an effective elimination of compression, this assumption may require some reconsideration. Also, it has been recently shown that a model which assumes a more linear transfer function when modeling loudness scaling data from hearing-impaired subjects is able to account for loudness recruitment and loudness summation effects (Launer, 1995).

As mentioned above, measurements of BM response show an initial linear growth, followed by a mid level compressive region and, at very high levels, a more linear region again. We have modeled our results using a simple power law, which could be thought of as representing the mid level
region of the BM transfer function (ca. 30–90 dB SPL). Given that we cannot obtain results for very high signal levels, due to the hazardous masker levels needed, and that results using very low signal levels seem to reflect the approach to absolute threshold in both groups of subjects, the approximation of a simple power law seems reasonable. Physiological investigations demonstrating BM compression have mainly been carried out using pure-tone stimuli (e.g., Yates, 1990; Ruggero, 1992). To our knowledge there are no published data on the growth of overall response at a single point on the BM using broadband noise. Fortunately, the method used above to derive the nonlinearity relies solely on the response to the tonal signal, and not on the response to the noise masker, as shown in the Appendix.

II. SIMULTANEOUS MASKING

While the effects of combining nonsimultaneous maskers have often been investigated, a number of workers have also reported examples of excess masking in simultaneous masking conditions. For example, Green (1967) measured the threshold of a sinusoidal signal in the presence of either a broadband noise or a sinusoidal masker at the same frequency as the signal. The maskers were either gated with the signal or presented continuously, and the sinusoidal masker was added either in random phase or in phase with the signal. Green then measured the combined effect of the two types of masker by presenting the signal in the presence of two equally effective maskers. Excess masking of between 0 and 11 dB was found, depending on the exact conditions. Green suggested that in cases where excess masking was found, different detection mechanisms might be employed in the presence of different maskers and that presenting them simultaneously might result in nonoptimal conditions for both types of detection mechanism. Lufti (1983), however, suggested that the effects of the maskers might in fact be processed independently and be compressed before their effects are combined, much in the same way as suggested by Penner (1980) for nonsimultaneous maskers. This approach has often been adopted since (Lufti, 1983; Humes et al., 1988; Humes and Jesteadt, 1989; Humes et al., 1992a; Humes and Jesteadt, 1991; Humes et al., 1992b).

While a simple power law, as suggested by Lufti (1983), provides a good fit to data from a number of different simultaneous-masking situations, it seems unlikely that excess masking in many of these situations can be accounted for in terms of the peripheral physiological mechanisms that we have suggested for nonsimultaneous masking. It is, for instance, not clear how the effects of two simultaneous maskers and a signal within one critical band could be separated in order for them to be compressed independently before being recombined. To examine the theoretical difficulties involved, consider the case of two simultaneous narrowband noise maskers with center frequencies just above and just below that of the signal (e.g., Lufti, 1983, Fig. 3). Ignoring issues of "off-frequency listening" (Johnson-Davies and Patterson, 1979; O'Loughlin and Moore, 1981), let us assume that detection of the signal is achieved by attending to the output of an auditory filter, or a place along the BM, with a best frequency very close to that of the signal. Measurements of BM responses to tones suggest that the function remains compressive for stimuli within at least ±10% of the CF (Ruggero, 1992, Fig. 1). Thus, we assume that the two maskers and the signal undergo the same compressive nonlinearity at the place of detection. At a given time, the activity due to the combined stimuli may be expressed as

\[ y(t) = (S(t) + N_1(t) + N_2(t))^p, \]

where \( S(t), N_1(t), \) and \( N_2(t) \) are the filtered representations of the signal and the two noise maskers, respectively, in intensity units, and \( p \) is the compressive index, as explained above. Note that schemes based on Eq. (3) do not predict excess masking, as the effects of the stimuli are combined before being compressed. This results in linear addition of effects, independent of whether the processing following filtering is linear or not. In the case of nonsimultaneous masking, where \( S(t), N_1(t), \) and \( N_2(t) \) do not overlap, Eq. (3) is equivalent to

\[ y(t) = S(t)^p + N_1(t)^p + N_2(t)^p. \]

Using Eq. (4), if \( y(t) \) is integrated over time, it is possible to predict excess masking, as the effects due to each of the stimuli are compressed before being combined. Clearly, Eq. (3) can not be rewritten as Eq. (4) in a simultaneous-masking situation, where the stimuli overlap, and yet it is Eq. (4) which is assumed for simultaneous masking in the schemes proposed by Lufti (1983) and Humes and Jesteadt (1989). As Lufti (1985) has noted, it is difficult to conceive of a physiological basis for such schemes which imply the ability of the auditory system to separate the effects of stimuli which are unresolved in the auditory periphery, compress them individually, and then recombine them. Excess masking in such cases is probably due to a disruption of detection cues (Moore, 1985), a concept already suggested by Green (1967). Note also that if the loss of a particular cue results in a threshold increase which is approximately constant with masker level, a power-law model will provide a good fit to combined-masker data, regardless of whether the effects of the stimuli are in fact processed separately. The situation is, however, different when the masker components are remote in frequency from the signal; such conditions are considered in Sec. III A.

In experiment 1, using nonsimultaneous maskers, we found a large difference in the best-fitting nonlinearity between the normally hearing and the hearing-impaired group. As we have argued above, simultaneous excess masking, where maskers and signal are close in frequency, is probably not mediated by the same mechanisms and can probably be accounted for by considering the available detection cues in each situation. Thus, if the same cues are available to both normally hearing and hearing-impaired subjects in the presence of a single masker, we would predict similar amounts of excess masking for both groups in the presence of two maskers. However, this prediction seems not to be consistent with data provided by Lufti (1987). He compared the effects of combining two equally effective simultaneous narrowband maskers in normally hearing and hearing-impaired subjects and found evidence for excess masking only in the normally hearing subjects; effectively no excess masking was
found in the hearing-impaired subjects. Lufti considered the possibility that the results may reflect more linear processing with the hearing-impaired subjects but, as explained above, it seems unlikely that separate compression of the maskers and signal underlies this phenomenon. As Lufti stated, a number of alternative explanations could account for his results. In his experiments, Lufti used 50-Hz-wide noise bands as maskers, one above and one below the signal frequency. Lufti lists a number of possible cues available to normally hearing subjects in the presence of one of the maskers which may not have been available to the hearing-impaired subjects. For instance, it is known that combination tones (e.g., Greenwood, 1971) are generally absent in hearing-impaired subjects (Leshowitz and Lindstrom, 1977); impaired frequency selectivity would reduce the effectiveness of off-frequency listening, and hearing-impaired subjects may not be able to make such effective use of "lip listening" (Hall and Grose, 1989).

In order to test our hypothesis that similar amounts of simultaneous excess masking should be seen in normally hearing and hearing-impaired subjects, a paradigm is required in which both groups are able to utilize the same detection cues. For this we return to the experiments of Green (1967) which combined sinusoidal and broadband noise maskers. A possible explanation for the excess masking observed in his experiments is as follows. Consider first the cues available with each masker individually. In the case of a sinusoidal masker, subjects are able to use the off-frequency spread of excitation as a cue and, as stated by Green (1967), can utilize any envelope fluctuations in the otherwise stationary stimulus. The importance of off-frequency listening in pure-tone intensity discrimination has been demonstrated by Viemeister (1972) and Moore and Raab (1974), and the fact that no excess masking was observed by Green (1967) when the maskers were gated with the signal suggests that the onset and offset ramps were utilized in the detection process. These two factors lead to a relatively low threshold signal-to-masker ratio at the output of the auditory filter centered at the signal frequency. For a broadband noise masker, the main cue may simply be the change in energy at the output of the auditory filter centered at the signal frequency (Green and Swets, 1966; Green, 1967). Hence, the signal-to-masker ratio at the output of the auditory filter centered at the signal frequency is higher than for the sinusoidal masker. When the two maskers are combined, the broadband noise disrupts the off-frequency excitation and the envelope cues that were available for the sinusoidal masker. Conversely, the sinusoidal masker adds considerably to the output of the auditory filter centered at the signal frequency. This means that the change in level produced by adding the signal is markedly reduced. Therefore, when equally effective tonal and noise maskers are combined, the cues which allowed reasonable performance for each masker separately are no longer available and excess masking occurs.

It would be expected that hearing-impaired subjects would be able to utilize a cue based on a tone's spread of excitation. Indeed, at low sensation levels, reduced frequency selectivity may increase the initial spread of excitation due to a tonal stimulus (Evans, 1975) which in turn may lead to a greater loss of information in the presence of a noise masker and so to somewhat greater excess masking than for normally hearing subjects at equal sensation levels. Experiment 2 tests this hypothesis using the same subjects as in experiment 1, and the same signal frequency of 4 kHz.

A. Conditions, stimuli, and procedure

Thresholds of a 4-kHz sinusoidal signal were measured in the presence of either a broadband white-noise masker (9-kHz low pass) or a sinusoidal masker of the same frequency as the signal. The signal had a steady-state duration of 100 ms and was presented in the temporal center of a 400-ms masker. All stimuli were gated using 5-ms raised-cosine ramps. In the case of the sinusoidal masker, the signal was added in phase with the masker. Initially, the threshold of the signal was measured in quiet for each subject. Then, for the normally hearing subjects, the level of each masker required to produce 10, 20, 30, 40, and 50 dB of masking was determined. For the hearing-impaired group, the reduced available dynamic range limited the signal level to about 20 dB SPL, and so masker levels were determined which individually produced 5, 10, 15, and 20 dB of masking. Once thresholds for the sinusoidal and the noise maskers had been individually determined, signal thresholds were measured in the presence of pairs of equally effective maskers, the sinusoidal and the noise. A trial consisted of two observation intervals, marked by lights, separated by a silent interval of 500 ms. The signal was presented randomly in either the first or second interval. The equipment, the method of presentation, and the procedure for measuring thresholds were the same as those used in experiment 1. In cases where the tonal masker level was varied, a run was automatically terminated if the masker level reached 105 dB SPL for the normally hearing subjects or 110 dB SPL for the hearing-impaired subjects. In these cases measurement of that data point was abandoned.

B. Subjects

The subjects were the same as those used in experiment 1. All subjects were given at least 4 h of practice, divided between the different conditions, before the data were recorded. No consistent improvements were found during the course of the experiment.

C. Results

Thresholds in quiet are given in Table II. Results from the single-masker conditions are shown in Fig. 6. Again, as the signal level was fixed and the masker level varied, masker level is plotted on the ordinate. The dashed lines in all four panels indicate linear growth of masking. Consider first the data from the normally hearing subjects (left panels). As can be seen in the top left panel, consistent with Weber's law, the growth of masking with the noise masker is linear. For the tonal masker (bottom left panel) the well-known "near miss" to Weber’s law can be seen (Viemeister, 1972); the Weber fraction decreases with increasing signal level.
FIG. 6. Simultaneous-masking thresholds in the presence of a noise masker (top panels) and a tonal masker (bottom panels). Data from the normally hearing and hearing-impaired subjects are presented in the left and right panels, respectively. Dashed lines denote linear growth of masking.

The data from the hearing-impaired subjects are somewhat more variable across subjects, although the general pattern of results is similar across the two groups.

The results from combining equally effective pairs of maskers are plotted in Fig. 7. The solid lines indicate no threshold increase in the presence of two maskers, the short-dashed lines indicate a 3-dB increase (energy summation), and the long-dashed lines show some predictions discussed below. At the highest levels, both groups show some excess masking, although the amount is much less than that seen in nonsimultaneous-masking conditions for normally hearing subjects. The amount of excess masking at the highest levels (3–4 dB) is similar to that found by Green (1967) in a comparable condition. For the normally hearing group there is no excess masking for single-masker thresholds below 40 dB SL; for the hearing-impaired group, no excess masking is observed at levels below 15 dB SL. In the hearing-impaired group, subject JC shows no excess masking at all. The fact that excess masking is only observed at the highest levels is consistent with the explanation involving the spread of excitation in the presence of a tonal masker; at low masker levels, the spread of excitation would be minimal and so would not provide an additional cue.

Interestingly, at the lowest levels a few data points lie below the solid line, indicating that the signal threshold decreased in the presence of two maskers. The single-masker conditions for these points were repeated and found to be consistent. The largest effect is seen at 5 dB SL with the hearing-impaired subject JC (right panel, open circle). From Fig. 6 (bottom right panel) it can be seen that at the lowest signal level, the tonal masker level for subject JC was approximately equal to the signal level. Thus, when the noise and tonal maskers were combined, the tonal masker was itself presented at threshold. As the tonal masker was barely audible in the combined situation, the task effectively became detection of a tone in noise. When the tone was presented, its effective level was 6 dB higher, due to its in-phase addition with the tonal masker. This may have enhanced detection. At higher levels, and for other subjects at all levels, the tonal masker was well above threshold and so the task in the combined condition was not simply that of detecting a tone in noise, but of detecting an increment in the tonal masker in the presence of the noise.

The data from the two groups were fitted using the MPL method (Humes and Jesteadt, 1989), as explained above. The best-fitting functions are shown in Fig. 7. Both groups were best fitted by weakly compressive functions, the normally hearing group with $p=0.7$ and the hearing-impaired group with $p=0.9$. Using a comparison of regression test (Snedecor
and Cochran, 1967), it was found that the difference in the value of \( p \) across the two groups was not significant [\( F(1,22) = 0.1, p > 0.25 \)]. If the data across groups are compared at equal, low sensation levels, the hearing-impaired group generally shows a greater increase in threshold in the presence of two maskers than the normally hearing group. For instance, at 20 dB SL the mean increase in masking for the hearing impaired is 5.4 dB (2.4-dB excess masking), while that for the normally hearing group is only 2 dB (less than energy summation). This is consistent with the prediction, outlined above, that reduced frequency selectivity leads to greater spread of excitation for the hearing impaired at low sensation levels.

In summary, both hearing-impaired and normally hearing groups showed some excess masking at the highest levels tested, although the amount was smaller for both groups. No excess masking was observed at lower levels. In contrast to the data for nonsimultaneous masking, the pattern of results was similar across the two groups and the form of the data is consistent with an explanation based on the spread of excitation in the presence of the tonal masker. This suggests that excess masking in this situation is not mediated by peripheral nonlinearities.

### III. General Discussion

The results of experiment 1, showing that hearing-impaired subjects exhibit much less excess masking than normally hearing subjects, are consistent with a theory linking the compressive transfer function \( \alpha \) of the BM to excess masking in nonsimultaneous masking conditions. A number of models of peripheral auditory mechanics have appeared in recent years (e.g., Zwicker, 1986; Lumer, 1987a.b; Goldstein, 1990; Yates, 1990), all of which share properties which are thought to reflect the active mechanisms of the healthy cochlea, namely, a compressive response to on-frequency stimuli and a linear response to frequencies much lower than CF, as well as a complex interaction of components resulting in suppression effects. In the following sections we discuss how such models may help in accounting for both simultaneous- and nonsimultaneous-masking effects, and how the absence or reduction of this nonlinearity before addition whereas the LRLF model, argues that by using as maskers a low-frequency tone and a relatively wideband noise, with its passband well below the signal frequency, any excess masking would not be due to loss of cues such as off-frequency listening or combination tones. In this case both the MPL and LRLF models predict excess masking, but for different reasons. The MPL model assumes that all stimuli are separately compressed using the same nonlinearity before addition whereas the LRLF model, consistent with the physiology, assumes that on-frequency stimuli are compressed and low-frequency stimuli are passed linearly. Seen in this context, the conditions of Zwicker and Herla (1975) and Humes et al. (1992b) could be considered as further cases of upward spread of masking. Note that the predictions of the MPL and the LRLF models are identical when the same exponent is used because, with the LRLF model, an increase of X dB in low-frequency masking energy (whether due to the addition of a second masker or an increase in the level of a single masker) requires an increase of \( X/p \) dB in signal level in order to maintain the threshold signal-to-noise ratio, where \( p \) is the compressive exponent, as used above. A comparison with Eq. (2) reveals that this increase in masker level of 1 dB will result in a 1-dB increase in activity at the BM place corresponding to the signal frequency, while a 1-dB increase in signal level will result in a less than 1-dB increase, due to the compression. Thus, in order to maintain the threshold signal-to-masker ratio, the signal level must be raised by more than 1 dB. This implies that upward spread of masking is a natural consequence of BM nonlinearity. Both Stelmachowicz et al. (1987) and Murnane and Turner (1991) have suggested that the upward spread of masking is related to the nonlinear behavior observed in the healthy cochlea and both have shown that, consistent with this suggestion, the growth of masking in hearing-impaired subjects is more linear. Also, a nonlinear model of cochlear mechanics (Lumer, 1987a.b) has been shown to be able to account for the upward spread of masking (Lumer, 1985).

The model of Oxenham and Moore (1994) uses a linear filter followed by a compressive nonlinearity which is not frequency dependent. In order to predict simultaneous-masking effects such as the upward spread of masking this scheme would need to be replaced by a more physiologically based nonlinear filter. However, for the purposes of illustrating, we simply postulate a peripheral model which exhibits a compressive power-law response to stimuli close to CF and a linear response to low frequencies (LRLF model). Clearly, an explanation simply in terms of compression at CF and linear response for stimuli well below CF is not fully satisfactory as it does not take into account the interaction of simultaneous stimuli within a nonlinear system, leading to effects such as two-tone suppression and combination tones. Nevertheless, upward spread of masking has also been observed in a forward-masking situation, where presumably no peripheral interaction occurs (Kidd and Feth, 1981).

The LRLF model can account for the simultaneous excess masking observed when both maskers are well below the signal frequency. This was the condition used by Humes et al. (1992b) in their second experiment and is similar to the conditions used by Zwicker and Herla (1975). Humes et al. argued that by using as maskers a low-frequency tone and a relatively wideband noise, with its passband well below the signal frequency, any excess masking would not be due to loss of cues such as off-frequency listening or combination tones. In this case both the MPL and LRLF models predict excess masking, but for different reasons. The MPL model assumes that all stimuli are separately compressed using the same nonlinearity before addition whereas the LRLF model, consistent with the physiology, assumes that on-frequency stimuli are compressed and low-frequency stimuli are passed linearly. Seen in this context, the conditions of Zwicker and Herla (1975) and Humes et al. (1992b) could be considered as further cases of upward spread of masking. Note that the predictions of the MPL and the LRLF models are identical when the same exponent is used because, with the LRLF model, an increase of X dB in low-frequency masking energy (whether due to the addition of a second masker or an increase in the level of a single masker) requires an increase of \( X/p \) dB in signal level in order to maintain the threshold signal-to-noise ratio, where \( p \) is the compressive exponent, as used above. A comparison with Eq. (2) reveals that this


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increase is identical to that predicted by the MPL model for the addition of two maskers. For example, the addition of two equally effective maskers gives a value $X=3$ and so with $p=0.2$ the signal threshold increases by 15 dB, as it does with the MPL. (This does not take into account the “internal noise” constant as it also has an identical effect in both schemes.)

In summary, the LRLF model predicts no excess masking in the presence of temporally overlapping stimuli falling within a narrow frequency range, but does predict excess masking in the presence of nonsimultaneous maskers, and simultaneous maskers whose frequencies are well below that of the signal. Upward spread of masking is also predicted by the LRLF model. All these attributes are consistent with the experimental data. Also, an increase in the value of $p$ (a reduction in compression), designed to simulate cochlear hearing loss, results in the prediction of reduced excess masking (cf. experiment 1) and more linear growth of masking (Stielmachowicz et al., 1987; Murnane and Turner, 1991). However, the model described here is primarily intended as a conceptual approach to highlight the possible importance of peripheral nonlinearities in both simultaneous- and nonsimultaneous-masking situations. It seems that a more complete approach to modeling psychoacoustic data may require the use of nonlinear cochlear models, such as those described by Zwicker (1986), Lumer (1987a,b), Yates (1990), and Goldstein (1990). In fact, Lumer (1985) has modeled psychoacoustic data (including those of Zwicker and Herla, 1975) encompassing the upward spread of masking and the combination of two simultaneous maskers.

**B. The growth and decay of forward masking**

As with the upward spread of masking, it has long been known that the growth of forward masking is generally nonlinear; the function can be fitted with a straight line on dB vs ms coordinates, but the slope is less than unity (Widin and Viemeister, 1979; Jesteadt et al., 1982; Moore and Glasberg, 1983). Our findings from experiment 1 also show this for normally hearing subjects at masker-signal intervals of 10 and 25 ms, while the growth at a masker-signal interval of 5 ms is more linear. The data from the hearing-impaired subjects indicate a more linear growth of masking at all masker-signal intervals. The nonlinear growth of forward masking in the normally hearing might, at least qualitatively, be accounted for by the change in BM nonlinearity as a function of level. At medium masker levels, where signal levels are within the more linear region of the BM transfer function (<30–40 dB SPL) and masker levels are mostly within the more compressive region, a given increase in the (compressed) masker level would require a smaller increase in signal level to maintain a given signal-to-noise ratio, giving nonlinear growth. The nonlinear growth would be expected to increase with increasing masker–signal time interval, as the difference in level between masker and signal would be greater, and so the signal at threshold would more likely be in the linear region of the BM transfer function. At masker levels where the signal at threshold would also fall within the compressive region this hypothesis would predict more linear growth of masking. As mentioned above, a number of studies have investigated the growth of forward masking, but in most of those the signal levels have been fairly low. Those points which do lie above about 40 dB SPL show a tendency toward more linear growth. For instance, the data of Moore and Glasberg (1983, Fig. 1) show that between the penultimate and highest masker level, at a signal frequency of 4 kHz, the growth function undergoes a dramatic steepening, becoming nearly linear. As mentioned, the data from the hearing-impaired subjects show a much more linear growth of forward masking than those from the normally hearing group at all masker–signal intervals. This would be expected with a linearization of the BM transfer function. However, further study is required to determine whether or not the BM nonlinearity can provide a full account of the growth of forward masking.

As shown here and in previous studies, the decay of forward masking in the hearing impaired is less rapid than for normally hearing subjects. This may also be accounted for by the decrease in compression in hearing-impaired subjects. For a fixed temporal window shape, a decrease in compression leads to a more gradual decay in predicted forward masking (Glasberg et al., 1987). Thus the observed reduction in the decay of both forward and backward masking may reflect only a change in nonlinearity, rather than an alteration of the underlying temporal window shape. This does not, however, account for the abnormal temporal asymmetries observed with the hearing-impaired subjects.

**IV. SUMMARY**

The elderly hearing-impaired subjects showed much less excess masking than normally hearing subjects for combined nonsimultaneous maskers (experiment 1). The lack of excess masking in the hearing-impaired subjects can be well described using a linear function (simple energy summation). The data from the normally hearing subjects were best accounted for by assuming that the stimuli were compressed before their effects were combined, with the best fit achieved by raising stimulus intensity to the power of 0.2. This is consistent with the theory linking excess masking in nonsimultaneous conditions to the nonlinear transfer characteristic of the basilar membrane in a healthy cochlea (Oxenham and Moore, 1994). However, these nonsimultaneous-masking results do not support models which assume the same compressive nonlinearity for both normally hearing and hearing-impaired subjects (Humes et al., 1988).

In a simultaneous-masking condition (experiment 2), the same pattern of results was found for both normally hearing and hearing-impaired subjects. This is consistent with an explanation based on the use of different cues in the different simultaneous-masking situations. The BM’s nonlinear transfer characteristic may also be important in accounting for the upward spread of masking, the nonlinear growth of forward masking, and the more gradual decay of forward masking observed in the hearing impaired.

**ACKNOWLEDGMENTS**

This research was supported by the Medical Research Council (U.K.), by a research studentship to the first author.
from the Engineering and Physical Sciences Research Council (U.K.), and by Meridian Audio. We would like to thank Tom Baer for helpful comments on an earlier version of this paper. Constructive criticism during the review process from Larry Humes, Walt Jesteadt, Armin Kolb Kraus, and Robert Lutfi also improved the paper.

**APPENDIX**

In fitting the data from nonsimultaneous-masking experiments, we assume that the stimuli are compressed before their effects are combined and we further assume that the model's compression is related to the compressive transfer function as measured on the BM. As mentioned in the discussion of experiment 1, the derived compressive exponent for normally hearing subjects (0.2) is close to that found experimentally for BM responses to pure tones at CF. We know of no published data on the growth of overall BM response in the presence of white noise, such as that used for the maskers in experiment 1. Given the nonlinear nature of the system, it may be premature to assume that the same nonlinearity applies to both broadband and tonal stimuli. Fortunately, the method of measuring the threshold of a tonal signal in the presence of two equally effective noise maskers is only sensitive to the nonlinear transformation of the signal. Thus our comparison of the model's exponent and the physiologically derived exponent remains valid, as shown below:

Assume that the noise maskers are in fact compressed differently from the tonal signal, the compression being characterized by exponents \( q \) and \( p \), respectively. In the presence of a single masker and signal at threshold, the output \( X_1 \) of the linear temporal integrator is

\[
X_1 = M_1^q + S_1^p
\]

where \( M_1^q \) is the output due to the masker and \( S_1^p \) is the output due to the signal at threshold. In the presence of two equally effective maskers, the output due to the maskers will be \( 2 M_1^q \). This doubles the contribution due to the maskers regardless of the value of \( q \). In order for the signal to remain at threshold, the value of \( S_1^p \) must also double. This implies that the signal level \( S_1 \) must increase by a factor of \( 2^q \). Thus the amount of excess masking predicted in the presence of two equally effective maskers is dependent only on the value of \( p \).


