

Perceptual Consequences of Cochlear Hearing Loss and their Implications for the Design of Hearing Aids

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This paper provides an overview of changes in the perception of sound that result from cochlear damage. It starts with a brief introduction to the physiology of the cochlea, emphasizing the role of the "active mechanism" and describing how cochlear function is altered by cochlear damage. Then the effects of cochlear damage on various aspects of perception are described, including absolute sensitivity, frequency selectivity, loudness perception and intensity discrimination, temporal resolution, temporal integration, pitch perception and frequency discrimination, and sound localization and other aspects of binaural and spatial hearing. The possible role of each of these aspects of auditory perception in the ability to understand speech in quiet and in noise is discussed and evaluated. It is concluded that, for losses up to about 45 dB, audibility is the single most important factor. However, for greater losses, poor discrimination of suprathreshold (audible) stimuli is also of major importance. The final section of the paper describes applications of the findings to hearing aid design. It is concluded that linear amplification can be of only limited benefit in compensating for the effects of cochlear damage. Hearing aids incorporating compression can help to compensate for the effects of reduced dynamic range. Digital signal processing to enhance spectral contrast may be of some help in compensating for the effects of reduced frequency selectivity.

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The most obvious symptom of cochlear hearing loss is a reduced ability to detect weak sounds. However, cochlear hearing loss is also accompanied by a variety of other changes in the way that sound is perceived. Even if sounds are amplified so that they are well above the threshold for detection, the perception of those sounds is usually abnormal. One aim of this paper is to give an account of the perceptual changes accompanying cochlear damage and to relate them to changes in the physiology of the cochlea. A second aim is to describe the implica-

tions of the perceptual changes for the perception of speech in quiet and in noise and for the design of hearing aids. This paper summarizes the more detailed coverage of these topics in Moore (1995).

PHYSIOLOGICAL CONSEQUENCES OF COCHLEAR DAMAGE

The functioning of the normal cochlea appears to reflect the operation of an active mechanism that is dependent on the integrity of the outer hair cells (OHCs) within the cochlea. This mechanism may involve the application of forces to the basilar membrane (BM) by the OHCs, and it plays an important role in producing the high sensitivity of the BM to weak sounds and the sharp tuning on the BM. The normal BM shows several nonlinearities (Rhode & Robles, 1974), including compressive input-output functions (Robles, Ruggero, & Rich, 1986; Sellick, Patuzzi, & Johnstone, 1982), two-tone suppression (Ruggero, Robles, & Rich, 1992), and combination-tone generation (Robles, Ruggero, & Rich, 1991); these nonlinearities also appear to depend on the operation of the active mechanism.

Cochlear hearing loss often involves damage to the OHCs and inner hair cells (IHCs); the stereocilia may be distorted or destroyed, or entire hair cells may die. The OHCs are generally more vulnerable to damage than the IHCs (Borg, Canlon, & Engström, 1995). When OHCs are damaged, the active mechanism tends to be reduced in effectiveness or lost altogether. As a result, several changes occur: the sensitivity to weak sounds is reduced, so sounds need to be more intense to produce a given magnitude of response on the BM; the tuning curves on the BM become much broader; and all of the frequency-selective nonlinear effects disappear (details are given later).

An example of the effects of cochlear damage on sensitivity and on tuning is given in Figure 1. This shows the input sound level required to produce a constant velocity of motion at a particular point on the BM as a function of stimulus frequency (Sellick, Patuzzi, & Johnstone, 1982). This is some-

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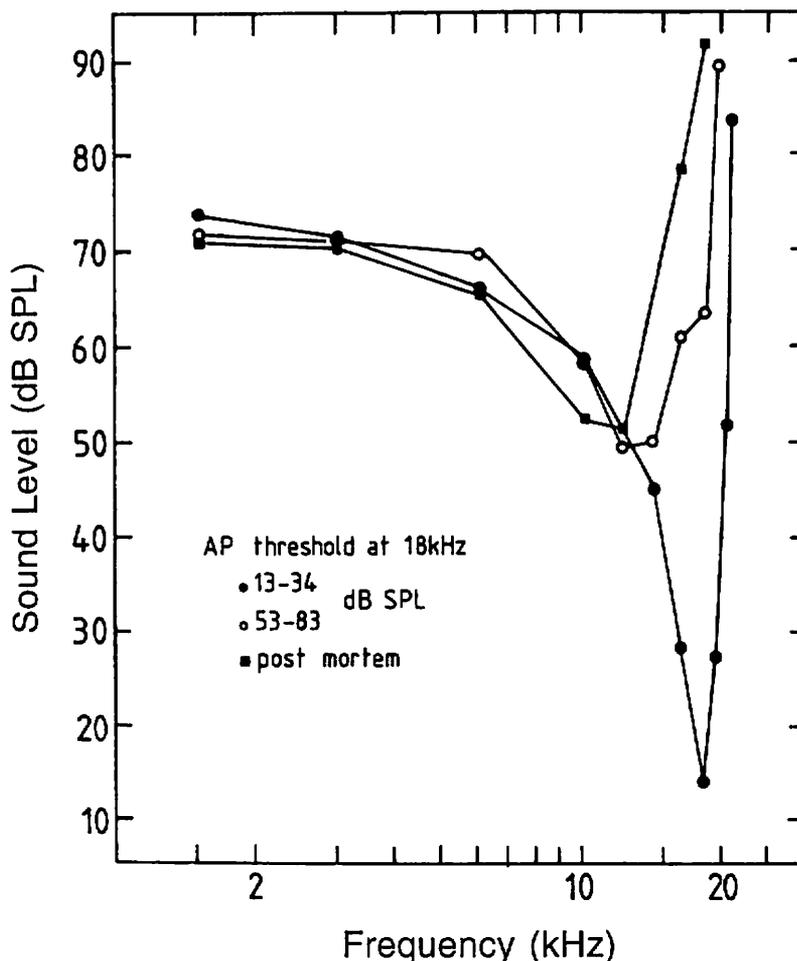


Figure 1. Tuning curves measured at a single point on the basilar membrane. Each curve shows the input sound level required to produce a constant velocity on the basilar membrane, plotted as a function of stimulus frequency. The curve marked by solid circles was obtained at the start of the experiment when the animal was in good physiological condition; the AP threshold was relatively low (13 to 34 dB SPL). The curve shown by open circles was obtained later in the experiment when the AP threshold was higher (53 to 83 dB SPL). Post mortem (squares), no AP threshold was measurable. Redrawn from Sellick et al. (1982).

times called a "constant velocity tuning curve." The condition of the cochlea was monitored by measuring the gross action potential (AP) evoked in the auditory nerve by brief tone pips. At the start of the experiment, when AP thresholds were low, a very sharp tuning curve was obtained (solid circles). During the course of the experiment, the condition of the animal deteriorated. As this happened, the tuning became broader, and the sound level required to produce the criterion response increased markedly around the tip. Such changes appear to be associated primarily with reduced functioning of the OHCs.

Consider next the way that responses on the BM change with sound level. In a normal ear the BM vibration is distinctly nonlinear; the magnitude of the response does not grow directly in proportion with the magnitude of the input (Rhode, 1971; Rhode & Robles, 1974; Ruggero, 1992; Sellick, Patuzzi, & Johnstone, 1982). This is illustrated in

Figure 2, which shows input-output functions of the BM for a place with a characteristic frequency (CF) of 8 kHz (from Robles, Ruggero, & Rich, 1986). A series of curves is shown; each curve represents a particular stimulating frequency, which is indicated by a number (in kHz) close to the curve. The output (velocity of vibration) is plotted on a log scale as a function of the input sound level (in dB SPL). If the responses were linear, the functions would be parallel to the dashed line. Two functions are shown for a CF tone (8 kHz), one (at higher levels) obtained about 1 hr after the other. The slight shift between the two was probably caused by a deterioration in the condition of the animal.

The function for the CF tone approaches linearity at low input levels (less than 20 dB SPL) and at high levels (above 90 dB) but has a very shallow slope at midrange levels. This indicates a compressive nonlinearity; a large range of input sound levels is compressed into a smaller range of responses on the

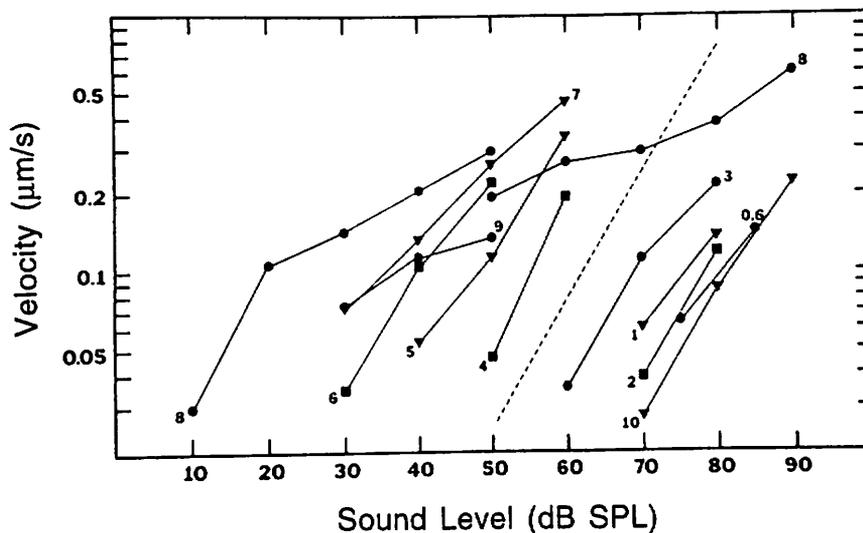


Figure 2. Input-output functions for a place on the BM with CF = 8 kHz. The stimulating frequency, in kHz, is indicated by a number close to each curve. The dashed line indicates the slope that would be obtained if the responses were linear (velocity directly proportional to sound pressure). Redrawn from Robles et al. (1986).

BM. The form of this function can be explained, at least crudely, in the following way. At low and medium sound levels, the active mechanism amplifies the response on the BM. The amplification may be as much as 55 dB. As the sound level increases, the amplification progressively reduces. Thus, the response grows more slowly than it would in a linear system. When the sound level is sufficiently high, around 90 dB SPL, the active mechanism is unable to contribute any amplification and the response becomes linear.

The nonlinearity mainly occurs when the stimulating frequency is close to the CF of the point on the BM that is being monitored. For stimuli with frequencies well away from the CF, the responses are more linear. Hence, the curves for frequencies of 7 and 9 kHz (close to CF) show shallow slopes, whereas the curves for frequencies below 7 kHz and above 9 kHz show steeper (linear) slopes. Effectively, the compression occurs only around the peak of the response pattern on the BM. As a result, the peak in the pattern flattens out at high sound levels.

The effects of cochlear damage on the input-output functions of the BM of a chinchilla are illustrated in Figure 3 (Ruggero & Rich, 1991). The solid curve with black squares, labeled "Before," shows the input-output function obtained when the cochlea was in good condition; the stimulus was a CF tone at 9000 Hz. The curve shows a compressive nonlinearity for input sound levels between about 30 and 90 dB SPL. In contrast, the response to a tone with a frequency of 1000 Hz, well below the CF, is steeper and is almost linear (solid curve with open circles).

To manipulate the functioning of the cochlea, the

animal was injected with furosemide, a diuretic that is known to disrupt hair cell potentials. The dashed curves in Figure 3 were obtained at various times after injection of the drug; the time is indicated by a range in minutes next to each curve. Shortly after the injection (11 to 19 minutes), the input-output function for the CF tone was markedly altered. The biggest alteration was at low sound levels. To produce a given response on the BM (say, 40 µm/s), the input level had to be increased by about 25 dB relative to the level measured before the injection. However, the response to a CF tone at a high level (80 dB SPL) was almost normal. This is consistent with the idea that the contribution of the active mechanism reduces progressively as the sound level is increased above about 40 dB. After a sufficiently long time (112 to 118 minutes), the input-output

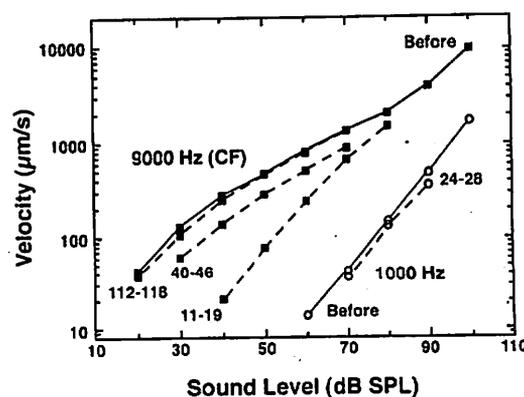


Figure 3. Input-output functions on the BM immediately before (solid lines) and after (broken lines) an intravenous furosemide injection. See text for details. Redrawn from Ruggero and Rich (1991).

function returned to normal. Thus, in this case the cochlear damage was reversible. Larger doses of the drug, or treatment with other drugs, can result in permanent cochlear damage.

It is noteworthy that injection of the drug did not change the input-output function for the 1000 Hz tone (see the curve with open symbols labeled 24 to 28). This is consistent with the idea that the active mechanism mainly influences responses around the peak of the response pattern evoked by a tone. Responses to tones with frequencies well away from the CF are linear and remain so when the active mechanism is damaged.

Information about the properties of sounds is carried in the auditory nerve not only in terms of the amounts of activity in different neurons (rate-place code), but also in terms of the time patterns of neural impulses (phase locking). The effect of cochlear damage on phase locking is not clear. Harrison and Evans (1979) used the drug kanamycin to produce hair cell damage (mainly to OHCs) in the guinea pig and found that phase locking was not affected. However, Woolf, Ryan, and Bone (1981) carried out a similar experiment using the chinchilla and found that phase locking was adversely affected by damage to the OHCs. For neurons with CFs corresponding to frequencies where the behavioral thresholds were elevated by 40 dB or more compared to normal, phase locking was significantly reduced.

EFFECTS OF COCHLEAR DAMAGE ON FREQUENCY SELECTIVITY

Frequency selectivity refers to the ability of the auditory system to separate or resolve (to a limited extent) the components in a complex sound. It is often quantified by using masking experiments to measure psychophysical tuning curves (PTCs) (Chistovich, 1957; Small, 1959) or to estimate auditory filter shapes using rippled noise or notched noise (Glasberg & Moore, 1990; Glasberg, Moore, & Nimmo-Smith, 1984; Houtgast, 1977; Moore & Glasberg, 1983; Moore & Glasberg, 1987; Patterson, 1976; Patterson & Moore, 1986; Patterson & Nimmo-Smith, 1980; Pick, Evans, & Wilson, 1977). It seems likely that frequency selectivity depends to a large extent on the filtering that takes place in the cochlea (Evans, Pratt, & Cooper, 1989). Hence, it would be expected that frequency selectivity as measured behaviorally would be poorer than normal in people with cochlear hearing loss. However, comparisons of frequency selectivity in normal-hearing and hearing-impaired subjects are complicated by several factors. One factor is the sound level of the stimuli used. The auditory filters of subjects with normal hearing sharpen on the low-frequency side

with decreasing level (Moore & Glasberg, 1987). This effect probably depends on the active mechanism in the cochlea.

The active mechanism is usually damaged or completely nonfunctioning in ears with cochlear damage. Hence, changes in frequency selectivity with level are absent or much less pronounced (Moore, Laurence, & Wright, 1985; Stelmachowicz, Lewis, Larson, & Jesteadt, 1987). As a result, the differences between normal-hearing and hearing-impaired subjects tend to decrease at high sound levels.

A second complicating factor is off-frequency listening; the signal may be detected using an auditory filter that is not centered at the signal frequency (Johnson-Davies & Patterson, 1979). Some measures of frequency selectivity, especially PTCs, can be strongly influenced by off-frequency listening. More importantly, the role of off-frequency listening may vary markedly depending on the sensation level (SL) of the stimuli and the frequency selectivity of the subject.

There have been several studies comparing PTCs in normal subjects and subjects with cochlear hearing loss (Bonding, 1979; Carney & Nelson, 1983; Festen & Plomp, 1983; Florentine, Buus, Scharf, & Zwicker, 1980; Hoekstra & Ritsma, 1977; Leshowitz, Linstrom, & Zurek, 1975; Nelson, 1991; Stelmachowicz, Jesteadt, Gorga, & Mott, 1985; Tyler, Wood, & Fernandes, 1982; Zwicker & Schorn, 1978). Although the studies differ in detail, their results are in general agreement that PTCs are broader than normal in the hearing-impaired subjects. However, it is difficult to quantify the differences from normal, owing to the problems discussed above. Most studies have found that the sharpness of tuning of the PTCs decreases with increasing absolute threshold, although the correlation between the two varies markedly across studies. No systematic differences in PTCs have been reported between cochlear losses of different origin such as noise-induced, Ménière's, aging, and hereditary losses.

Pick, Evans, and Wilson (1977) estimated auditory filter bandwidths using a rippled-noise masker for both normal-hearing subjects and subjects with cochlear hearing loss. The overall noise level was either 90 or 100 dB SPL. Test frequencies of 0.5, 1, 2, and 4 kHz were used. When the absolute thresholds of the hearing-impaired subjects were less than about 20 dB HL at the test frequency, the filter bandwidths were generally within the normal range. With increasing hearing loss above 20 dB, the filter bandwidths generally increased. On average, bandwidths reached about twice the normal values for absolute thresholds in the range 40 to 50 dB HL, although considerable scatter was evident in the

data. For example, some subjects with thresholds of about 50 dB HL had almost normal bandwidths, whereas others had bandwidths four to five times greater than normal. Broadly comparable results were obtained by Hoekstra (Reference Note 4) and by Festen and Plomp (1983).

Auditory filter shapes of subjects with cochlear impairments have been estimated in several studies using notched-noise maskers (Dubno & Dirks, 1989; Glasberg & Moore, 1986; Laroche, Hétu, Quoc, Josseland, & Glasberg, 1992; Leek & Summers, 1993; Leeuw & Dreschler, 1994; Peters & Moore, 1992; Sommers & Humes, 1993; Stone, Glasberg, & Moore, 1992; Tyler, Hall, Glasberg, Moore, & Patterson, 1984). The results generally agree in showing that auditory filters are broader than normal in hearing-impaired subjects and that, on average, the degree of broadening increases with increasing hearing loss.

Figure 4 summarizes values of the equivalent rectangular bandwidth (ERB) of the auditory filter obtained from subjects with cochlear hearing loss in my laboratory and in the laboratory of Robert Peters. The summary was prepared by Michael Stone (Reference Note 7). The upper panel shows ERB values plotted relative to the value for young, normally hearing subjects at moderate sound levels. The lower panel shows ERB values plotted relative to the value for young, normally hearing subjects tested at the same sound pressure level as the impaired subjects, assuming that the ERB for normal subjects varies with level as described by Glasberg and Moore (1990). A value of 1 would indicate normal auditory filters. In both cases, the ERB values are plotted as a function of the absolute threshold (dB HL) at the test frequency. There is a clear trend for the ERB to increase with increasing absolute threshold. The increase is less in the lower panel because the auditory filters for normal subjects broaden with increasing level of the test stimuli. However, the trend is still quite clear. There is also considerable scatter in the data, indicating that the ERB of the auditory filter cannot be predicted reliably from the absolute threshold.

It is obvious from the studies discussed above that masking effects are often more pronounced in hearing-impaired than in normal-hearing subjects. However, the size of the difference depends greatly on the spectral characteristics of the signal and masker. When the signal and masker overlap spectrally (for example, when the signal is a sinusoid and the masker is a broadband noise without distinct spectral peaks), masked thresholds are usually only slightly greater for hearing-impaired than for normal listeners. However, when the signal and masker differ in spectrum, masking may be consid-

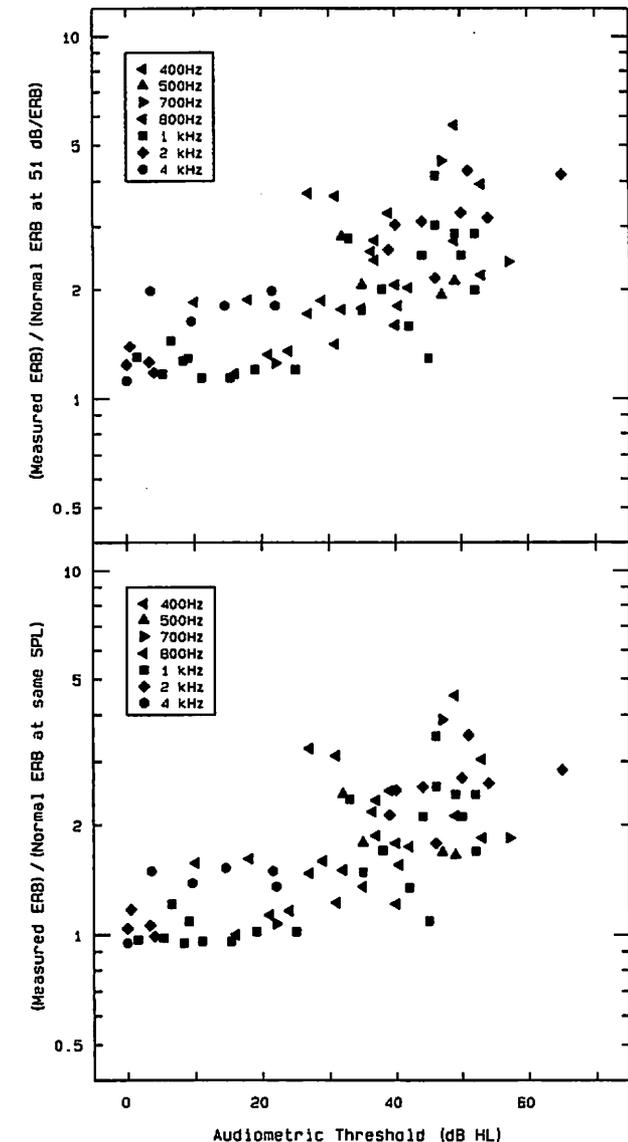


Figure 4. Values of the ERB of the auditory filter for subjects with cochlear hearing loss, plotted as a function of absolute threshold (dB HL) at the test frequency. The ERB values are plotted either relative to the values for young normally hearing subjects at a moderate sound level or relative to the ERBs for young normally hearing subjects tested at the same sound level.

erably greater in the hearing impaired. There are two obvious situations where this might apply. One is when the average spectrum of a masking sound differs from that of a signal. For example, the signal may be a high-frequency warning siren and the masker may be primarily low-frequency noise from air-conditioning or machinery. The second is when the signal and the masker differ in their short-term spectra. For example, the signal might be a vowel sound from an attended talker and the masker might be a different vowel sound from an interfering talker. A hearing-impaired person with reduced fre-

quency selectivity will be less able than normal to take advantage of the spectral differences between the two vowels. This issue is examined in more detail later.

EFFECTS OF COCHLEAR DAMAGE ON LOUDNESS PERCEPTION

Most, if not all, people suffering from cochlear damage show loudness recruitment (Fowler, 1936; Steinberg & Gardner, 1937). The absolute threshold is higher than normal. However, when a sound is increased in level above the absolute threshold, the rate of growth of loudness level with increasing sound level is greater than normal. When the level is sufficiently high, usually around 90 to 100 dB SPL, the loudness reaches its "normal" value; the sound appears as loud to the person with impaired hearing as it would to a person with normal hearing. With further increases in sound level above 90 to 100 dB SPL, the loudness grows in an almost normal manner.

A complementary way of describing this effect is in terms of dynamic range. This refers to the range of sound levels over which sounds are both audible and comfortable. The lower end of the dynamic range is determined by the absolute threshold for detecting sounds. The upper end is determined by the level at which sounds start to become uncomfortably loud. Typically, in people with cochlear hearing loss, the absolute threshold is elevated, but the level at which sounds become uncomfortably loud is about the same as normal. Hence, the dynamic range is reduced compared with normal.

On average, the rate at which loudness grows with increasing intensity goes up with increasing absolute threshold at the test frequency (Glasberg & Moore, 1989; Hellman & Meiselman, 1990 and 1993; Kiessling, Steffens, & Wagner, 1993; Miskolczy-Fodor, 1960). This is consistent with the idea that threshold elevation and loudness recruitment are both linked to the loss of the active mechanism in the cochlea. When the absolute threshold is high, the dynamic range can be very small indeed.

A plausible explanation for loudness recruitment is that it arises from a reduction in or loss of the compressive nonlinearity in the input-output function of the BM. If the input-output function on the BM is steeper (less compressive) than normal in an ear with cochlear damage, it would be expected to lead to an increased rate of growth of loudness with increasing sound level. However, at high sound levels, around 90 to 100 dB SPL, the input-output function becomes almost linear in both normal and impaired ears. The magnitude of the BM response at high sound levels is roughly the same in a normal

and an impaired ear (see Figure 3). This can explain why the loudness in an impaired ear usually "catches up" with that in a normal ear at sound levels around 90 to 100 dB SPL.

Kiang, Moxon, and Levine (1970) and Evans (1975) suggest that reduced frequency selectivity might be the main factor contributing to loudness recruitment. They suggest that, once the level of a sound exceeds threshold, the excitation in an ear with cochlear damage spreads more rapidly than normal across the array of neurons, and this leads to the abnormally rapid growth of loudness with increasing level. However, both experimental studies (Hellman, 1978; Hellman & Meiselman, Reference Note 3; Moore, Glasberg, Hess, & Birchall, 1985; Zeng & Turner, 1991) and theoretical analyses (Moore, 1995) suggest that reduced frequency selectivity plays only a minor role.

Most studies of loudness recruitment have used steady sounds of relatively long duration, such as tone bursts or bursts of noise (Allen & Jeng, 1990; Kiessling, Steffens, & Wagner, 1993; Miskolczy-Fodor, 1960; Moore, Glasberg, Hess, & Birchall, 1985; Moore, Johnson, Clark, & Pluvinaige, 1992; Pluvinaige, 1989). Moore, Wojtczak, and Vickers (1996) examined how loudness recruitment affects the perception of amplitude modulation depth to see whether the results could be predicted on the basis of measures of recruitment obtained with steady tones. Three subjects with unilateral cochlear hearing loss were used. Both loudness-matching functions and modulation-matching functions between the two ears were obtained. In the latter case, modulated tones were presented alternately to the two ears, and subjects had to adjust the modulation depth in one ear until the depth of modulation appeared the same in the two ears. The modulation was sinusoidal on a decibel scale. The modulation rates used (4, 8, 16, and 32 Hz) were chosen to span the range of the most prominent modulations present in the envelope of speech.

An example of the results is given in Figure 5. A given modulation depth in the impaired ear was matched by a greater modulation depth in the normal ear, consistent with the idea that recruitment affects perception of the dynamic aspects of sounds. To a first approximation, the modulation-matching functions were independent of modulation rate. The modulation-matching functions could be predicted reasonably well from the loudness-matching results obtained with steady tones; these predictions are shown by the dashed lines in the figure. Hence, the results were consistent with the idea that loudness recruitment results from the loss of a fast-acting compressive nonlinearity that operates in the normal peripheral auditory system.

For sounds with inherent amplitude fluctuations,

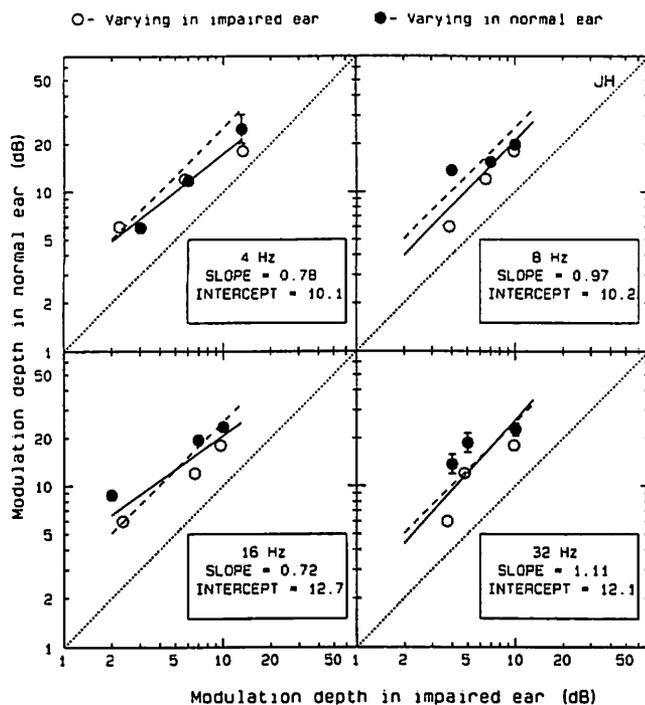


Figure 5. Modulation-matching functions for a subject with unilateral cochlear hearing loss, shown separately for each modulation rate. Solid symbols indicate matches with the variable (adjusted) tone in the normal ear and open symbols indicate matches with the variable tone in the impaired ear. Error bars show ± 1 SE. They are omitted where they would be smaller than the symbol used to represent a given point. The solid lines are best-fitting lines. The insets show the slopes of these lines and the value of the ordinate (modulation depth in the normal ear) for an abscissa (modulation depth in the impaired ear) value of 5 dB; this latter quantity is referred to as the "intercept." The dashed lines are predictions from the loudness-matching data for steady tones. Redrawn from Moore, Wojtczak, and Vickers (1996).

such as speech or music, recruitment results in an exaggeration of the perceived dynamic qualities. The sound appears to fluctuate more in loudness than it would for a normally hearing person. When listening to music, the forte passages may be perceived at almost normal loudness, but the piano passages may be inaudible. This is illustrated by some simulations on a compact disc available from the author (Reference Note 5).

EFFECTS OF COCHLEAR DAMAGE ON INTENSITY RESOLUTION

The ability to detect changes in intensity, or to compare the intensity of two separate sounds, is usually assumed to be based on the loudness sensations evoked by the sounds. In people with cochlear damage, a given change in intensity usually results in a larger-than-normal change in loudness. Hence, it might be expected that intensity discrimination

would be better than normal. However, this expectation is based on the assumption that the just-detectable change in loudness is unaffected by the cochlear damage, and this assumption may not be valid. For several different measures of intensity discrimination, including the detection of amplitude modulation and the detection of differences in intensity of separate pulsed tones, people with cochlear hearing loss perform as well as or better than normal when the comparison is made at equal sensation level (SL). However, when compared at equal sound pressure level (SPL), the performance of subjects with cochlear damage is not better than normal, and may be worse than normal (Buus, Florentine, & Redden, 1982; Glasberg & Moore, 1989; Schroder, Viemeister, & Nelson, 1994; Turner, Zwillocki, & Fillion, 1989).

In everyday life, hearing-impaired people often listen at lower SLs than normally hearing people, so their intensity discrimination can be worse than normal. However, this does not appear to lead to marked problems, because it is rare in everyday life for critical information to be carried by small changes in intensity. Although intensity contrasts can convey information in speech, the contrasts involve rather large changes in intensity, changes that are usually well above the threshold of detection for both normally hearing and hearing-impaired persons.

EFFECTS OF COCHLEAR DAMAGE ON TEMPORAL RESOLUTION

Although temporal resolution can be adversely affected by cochlear damage, this is not the case for all measures of temporal resolution. To understand the reasons for this, it is helpful to use a model of temporal processing in the normal auditory system and to consider how the different stages of the model may be altered by cochlear pathology. Temporal processing depends on both analysis of the time pattern occurring within each frequency "channel" and comparison of the time patterns across channels. This paper concentrates mainly on within-channel processes.

Several models of temporal resolution are similar in general form. There is an initial stage of bandpass filtering, reflecting the action of the auditory filters. Each filter is followed by a nonlinear device. This device may be thought of as crudely representing some aspects of the process of transduction from excitation at a particular point on the BM to activity in the auditory nerve. The output of the nonlinear device is fed through a "smoothing" device, which can be implemented either as a lowpass filter (Viemeister, 1979) or a sliding temporal integrator

(Moore, Glasberg, Plack, & Biswas, 1988). Usually this smoothing device is thought of as occurring after the auditory nerve; it is assumed to reflect a relatively central process. The output of the smoothing device is fed to a decision device.

Several factors can influence measures of temporal resolution in people with cochlear hearing loss. Not all of these factors are directly connected with the model outlined above, although some are. One important factor is the sound level used. Many measures of temporal resolution show that performance in normally hearing subjects worsens at low SLs (Buus & Florentine, 1985; Fitzgibbons & Gordon-Salant, 1987; Peters, Moore, & Glasberg, 1995; Plomp, 1964; Shailer & Moore, 1983). The changes in performance with sound level may reflect mainly changes in the efficiency of central decision processes rather than changes in the temporal integrator or temporal window (Peters et al., 1995). It is not generally possible to test hearing-impaired subjects at high SLs because they have loudness recruitment. On some measures of temporal resolution, such as the detection of gaps in bands of noise or the rate of recovery from forward masking, hearing-impaired subjects appear markedly worse than normal-hearing subjects when tested at the same SPLs but only slightly worse at equal SLs (Fitzgibbons & Wightman, 1982; Glasberg, Moore, & Bacon, 1987; Tyler, Summerfield, Wood, & Fernandes, 1982).

For deterministic stimuli that have no inherent random fluctuations, hearing-impaired subjects can actually perform a little better than normally hearing subjects when tested at equal SLs. This applies to the detection of gaps in sinusoids (Moore & Glasberg, 1988b; Moore, Glasberg, Donaldson, McPherson, & Plack, 1989) and to the discrimination of Huffman sequences (Jesteadt, Bilger, Green, & Patterson, 1976); the latter are click-like sounds with identical power spectra but different phase spectra.

Another important consideration is the bandwidth available to the listeners. This can be seen clearly by consideration of studies measuring the temporal modulation transfer function (TMTF). The TMTF shows the amount of amplitude modulation required for detection of the modulation, plotted as a function of modulation rate (Viemeister, 1979). It is generally assumed that the ability to detect modulation at high rates is limited by the temporal resolution of the ear. Several studies measuring TMTFs for broadband noise carriers show that hearing-impaired listeners are generally less sensitive to high rates of modulation than are normal-hearing listeners (Bacon & Viemeister, 1985; Lamore, Verweij, & Brocaar, 1984; Formby, Reference Note 2). However, this may be largely a consequence of the

fact that high frequencies were inaudible to the impaired listeners (Bacon & Viemeister, 1985); most of the subjects had greater hearing losses at high frequencies than at low. When the broadband noise is lowpass filtered, as a crude simulation of the effects of threshold elevation at high frequencies, normally hearing subjects also show a reduced ability to detect modulation at high rates (Bacon & Viemeister, 1985).

Bacon and Gleitman (1992) measured TMTFs for broadband noise using subjects with relatively flat hearing losses. They found that at equal (high) sound pressure levels (SPLs) performance was similar for hearing-impaired and normally hearing subjects. At equal (low) SLs, the hearing-impaired subjects tended to perform better than the normally hearing subjects. Moore, Shailer, and Schooneveldt (1992) controlled for the effects of listening bandwidth by measuring TMTFs for an octave-wide noise band centered at 2 kHz, using subjects with unilateral and bilateral cochlear hearing loss. Over the frequency range covered by the noise, the subjects had reasonably constant thresholds as a function of frequency in both their normal and their impaired ears. This ensured that there were no differences between subjects or ears in terms of the range of audible frequencies in the noise. To ensure that subjects were not making use of information from frequencies outside the nominal passband of the noise, the modulated carrier was presented in an unmodulated broadband noise background. The results for the subjects with unilateral impairments are shown in Figure 6. The modulation detection thresholds are expressed as $20\log m$, where m is the modulation index. Thresholds are plotted with more negative numbers (corresponding to better performance) at the top. It can be seen that performance is similar for the normal and the impaired ears at both equal SPL and equal SL, although there is a slight trend for the impaired ears to perform better at equal SL.

Studies of gap detection also show clear effects of the audible frequency range of the stimuli. When a broadband noise is used to "mark" the gap, gap thresholds become progressively larger as the audible frequency range of the stimuli is reduced by increasing high-frequency hearing loss (Buus & Florentine, 1985; Florentine & Buus, 1984; Salvi & Arehole, 1985).

As described earlier, subjects with cochlear hearing loss usually have auditory filters that are broader than normal. One might expect that this would lead to improved temporal resolution because broad filters have a faster temporal response than narrow filters. However, the auditory filters in normal ears appear to play little role in limiting tem-

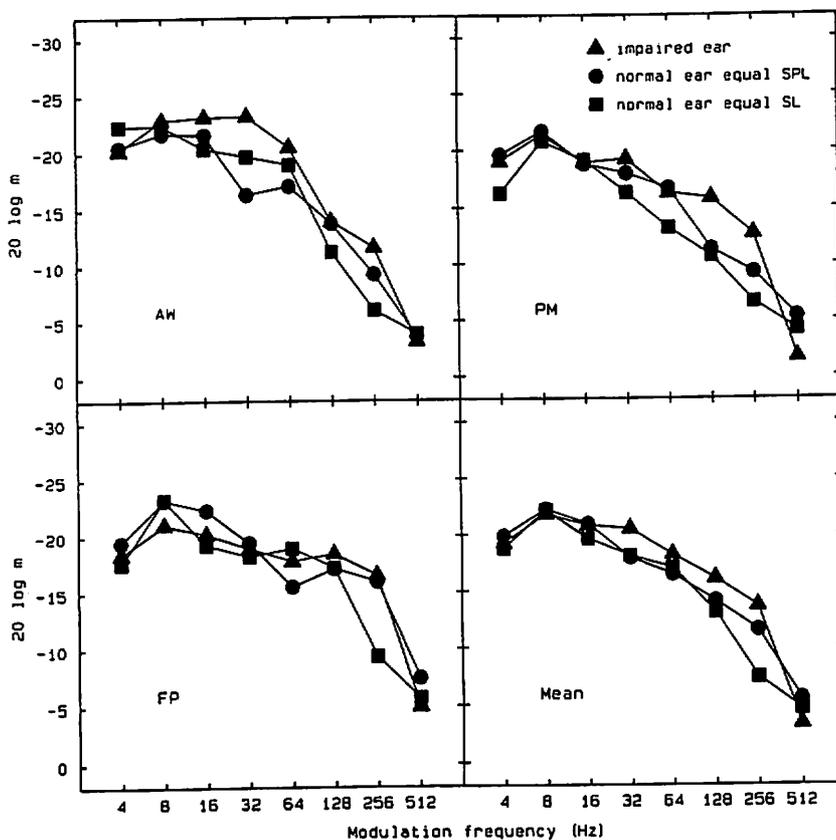


Figure 6. Temporal modulation transfer functions obtained using a bandpass noise carrier for the normal and impaired ears of three subjects with unilateral cochlear damage.

poral resolution except at very low frequencies (Plack & Moore, 1990). Hence, it has proved difficult to demonstrate improvements in temporal resolution resulting from broadened auditory filters.

For stimuli that contain slow random fluctuations in amplitude, such as narrow bands of noise, subjects with cochlear damage often perform more poorly than normal in tasks such as gap detection even when the stimuli are well above threshold and when all of the components of the stimuli fall within the audible range (Buus & Florentine, 1985; Fitzgibbons & Wightman, 1982; Florentine & Buus, 1984; Glasberg, Moore, & Bacon, 1987). However, gap detection is not usually worse than normal when the stimuli are sinusoids, which do not have inherent amplitude fluctuations (Moore & Glasberg, 1988b; Moore, Glasberg, Donaldson, McPherson, & Plack, 1989). Glasberg et al. (1987) and Moore and Glasberg (1988b) suggest that the poor gap detection for narrowband noise stimuli might be a consequence of loudness recruitment. For a person with recruitment, the inherent amplitude fluctuations in a narrowband noise would result in larger-than-normal loudness fluctuations from moment to moment, so that inherent dips in the noise might be more confusable with the gap to be detected. This idea can also be expressed in terms of the model of temporal resolution described earlier. When cochlear damage

occurs, the input-output function of the BM becomes less compressive, having a slope closer to unity; this can be incorporated in the model by making the nonlinearity less compressive. Experiments in which the amplitude fluctuations in bands of noise are either expanded or compressed support the idea that increased fluctuations result in impaired gap detection (Glasberg & Moore, 1992).

For most subjects with cochlear damage, recruitment or, equivalently, a reduction in the peripheral compressive nonlinearity may provide a sufficient explanation for increased gap thresholds. Thus, it is not usually necessary to assume any abnormality in temporal processing occurring after the cochlea. However, a few subjects show impairments in temporal resolution even using nonfluctuating stimuli (Jesteadt, Bilger, Green, & Patterson, 1976; Moore & Glasberg, 1988b; Moore, Glasberg, Donaldson, McPherson, & Plack, 1989; Plack & Moore, 1991). It is possible that the subjects showing this impaired resolution had damage to both the OHCs (affecting the active process and the compressive nonlinearity) and IHCs (affecting the transduction process), or that they had a retrocochlear component to their hearing loss.

Unfortunately, most sounds in everyday life are characterized by unpredictable fluctuations in amplitude from moment to moment. For such sounds,

people with cochlear damage will have more difficulty than normal in following the temporal structure of the sounds. In addition, temporal resolution may be poor because the sounds are at low SPLs and/or because the audible bandwidth of the stimuli is restricted. All of these factors can lead to problems in understanding speech and in discriminating and identifying music and environmental sounds.

EFFECTS OF COCHLEAR DAMAGE ON TEMPORAL INTEGRATION

For people with cochlear damage, the change in absolute threshold with signal duration is often smaller than it is for normally hearing people. If the thresholds are plotted on dB versus log-duration coordinates, the slopes are usually much less in absolute value than the typical value of -3 dB/doubling found for normally hearing people. This is often described as reduced temporal integration (Carlyon, Buus, & Florentine, 1990; Chung, 1981; Elliott, 1975; Gengel & Watson, 1971; Hall & Fernandes, 1983; Pedersen & Eberling, 1973). There is a trend for higher absolute thresholds to be associated with flatter slopes. In other words, the greater the hearing loss, the more reduced is the temporal integration.

It seems likely that the main cause of reduced temporal integration in people with cochlear damage is a reduction or complete loss of the compressive nonlinearity on the BM. This leads to steeper input-output functions on the BM (see Figure 3) and to steeper rate-versus-level functions in the auditory nerve. According to the models of temporal integration proposed by Zwislocki (1960) and by Penner (1972), this will lead to reduced temporal integration. Figure 7, adapted from Moore (1991), illustrates schematically two rate-versus-level functions: the left-hand curve shows a typical function for a low-threshold neuron in a normal auditory system; the right-hand curve shows a typical function for a neuron in an auditory system with cochlear damage. The curve is shifted to the right, reflecting a loss of sensitivity, and is steeper, reflecting loss of the compressive nonlinearity on the BM.

Consider now how steeper rate-versus-level functions can lead to reduced temporal integration. Assume that, to a first approximation, the threshold for detecting a sound requires a fixed number of neural spikes to be evoked by that sound. Assume also, for the sake of simplicity, that the neurons involved in detection at absolute threshold are relatively homogeneous in terms of their input-output functions. The lower dashed horizontal line in Figure 7 indicates the number of neural spikes per second, N_1 , needed to achieve absolute threshold for

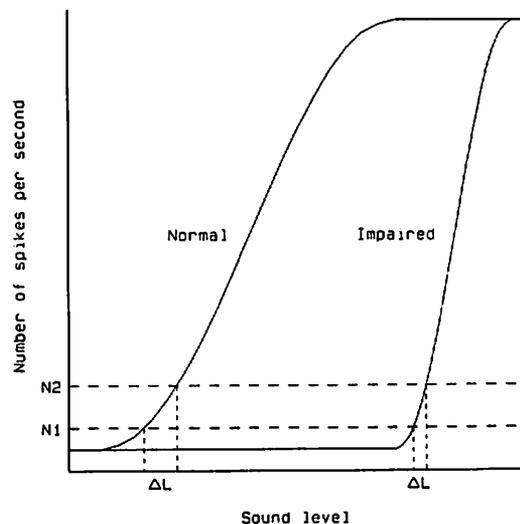


Figure 7. Schematic illustration of rate-versus-level functions in single neurons of the auditory nerve for a normal ear (left curve) and an impaired ear (right curve). The horizontal dashed lines indicate the mean firing rate needed for threshold for a long-duration sound (lower line, rate N_1) and a short-duration sound (upper line, rate N_2) on the assumption that threshold corresponds to a fixed total number of spikes.

a long-duration sound. If the duration of the sound is decreased by a factor R , then the level has to be increased to restore the total number of spikes evoked by the sound. Assume that the higher spike rate needed for the shorter-duration sound is N_2 , where $N_2 = R \times N_1$. For example, if the duration is halved, the spike rate has to be increased by a factor of two to achieve the same total spike count. The increase in level, ΔL , needed to achieve this increased rate is greater for the normal than for the impaired ear because the rate-versus-level function is steeper for the impaired ear, and this explains the reduced temporal integration.

One consequence of reduced temporal integration is that the loss of sensitivity to weak sounds experienced by people with cochlear damage is less severe for short sounds than for long sounds. Consider, as an example, two sounds with durations 400 msec and 10 msec. For a normally hearing person, the level required for detection of these two sounds might be, for example, 4 dB SPL and 20 dB SPL, respectively; the shorter sound has to be about 16 dB higher in level to reach the absolute threshold. For a person with moderate cochlear damage, the threshold for detecting the longer sound might be 54 dB SPL, i.e., 50 dB higher than normal. However, the threshold for detecting the shorter sound might be 60 dB SPL, which is only 40 dB higher than normal. Thus the "loss" relative to normal hearing is 10 dB less for the shorter sound than for the longer sound. This could be of relevance to speech perception

because the loss of audibility for brief speech sounds (such as plosives), relative to normal, would be less than the loss for longer-duration sounds (such as vowels).

EFFECTS OF COCHLEAR DAMAGE ON PITCH PERCEPTION AND FREQUENCY DISCRIMINATION

Traditionally there have been two main theories of pitch perception. One, the place theory, suggests that the pitch of a sound is related to the distribution of activity evoked by that sound in the peripheral auditory system, i.e., the excitation pattern. Usually it is assumed that pitch is determined by the place of maximum vibration on the BM or by the CFs of the neurons that are excited most. Frequency discrimination is assumed to be related to the sharpness of the excitation pattern (Zwicker, 1956 and 1970). Hence, cochlear damage, which results in a broadening of the excitation pattern, should result in poor frequency discrimination. The alternative theory, called the temporal theory, assumes that pitch is determined from the exact timing of neural impulses, i.e., from phase locking. According to this theory, cochlear damage may affect frequency discrimination if it results in abnormal phase locking.

Several studies have measured thresholds for detecting differences in frequency between successive tone pulses in subjects with cochlear damage (Freyman & Nelson, 1986, 1987, and 1991; Gengel, 1973; Hall & Wood, 1984; Moore & Glasberg, 1986b; Moore & Peters, 1992; Simon & Yund, 1993; Tyler, Wood, & Fernandes, 1983). I will refer to thresholds measured in this way as difference limens for frequency (DLFs). The results have generally shown that DLFs are adversely affected by cochlear damage. However, there is considerable variability across subjects, and the size of the DLF has not been found to be strongly correlated with the absolute threshold at the test frequency, nor is it strongly correlated with measures of frequency selectivity (Moore & Peters, 1992; Tyler et al., 1983). Several other lines of experimental evidence, reviewed in Moore (1995), suggest that place models of pitch perception are not adequate to account for the increased DLFs associated with cochlear hearing loss, although place models seem to account reasonably well for changes in the ability to detect frequency modulation.

An alternative way of accounting for the fact that cochlear damage results in larger-than-normal DLFs is in terms of loss of neural synchrony (phase locking) in the auditory nerve. Another possibility is that the central mechanisms involved in the analysis of phase-locking information make use of differences in the preferred time of firing of neurons with

different CFs; these time differences arise from the propagation time of the travelling wave on the BM (Loeb, White, & Merzenich, 1983; Shamma, 1985). The propagation time along the BM can be affected by cochlear damage, and this could disrupt the processing of the temporal information by central mechanisms.

Cochlear damage sometimes leads to changes in perceived pitch. For people with unilateral cochlear damage, or asymmetrical hearing losses, the same tone presented alternately to the two ears may be perceived as having different pitches in the two ears. This effect is given the name diplacusis. Sometimes different pitches are perceived even when the hearing loss is the same in the two ears. The origin of these shifts is unclear; neither place theory nor temporal theory appear to provide an adequate account (Burns & Turner, 1986).

The pitch discrimination of complex tones by hearing-impaired people has been the subject of several studies (Arehart, 1994; Hoekstra & Ritsma, 1977; Moore & Glasberg, 1988c and 1990; Moore & Peters, 1992; Rosen, 1987). Most studies have required subjects to identify which of two successive harmonic complex tones had the higher fundamental frequency (F_0). The threshold determined in such a task will be described as the difference limen for a complex (DLC). Although considerable individual variability is apparent in the results, some general points can be made:

1. For some subjects, when F_0 was low, DLCs for complex tones containing only low harmonics were markedly higher than for complex tones containing only high harmonics, suggesting that pitch was conveyed largely by the higher, unresolved harmonics.
2. For some subjects, DLCs were larger for complex tones with lower harmonics (1 to 12) than for tones without lower harmonics (4 to 12 and 6 to 12) for F_0 s up to 200 Hz. In other words, adding lower harmonics made performance worse. This may happen because, when auditory filters are broader than normal, adding lower harmonics can create more complex waveforms at the outputs of the auditory filters, making temporal analysis more difficult.
3. The DLCs were for the most part correlated only weakly with measures of frequency selectivity. There was a slight trend for large DLCs to be associated with poor frequency selectivity, but the relationship was not a close one. Some subjects with very poor frequency selectivity had reasonably small DLCs.
4. There can be significant effects of the component phase. In several studies, DLCs have been measured with the components of the

harmonic complexes added in one of two phase relationships, all cosine phase or alternating cosine and sine phase. The former results in a waveform with prominent peaks and low amplitudes between the peaks. The latter results in a waveform with a much flatter envelope. The DLCs tended to be larger for complexes with components added in alternating sine/cosine phase than for complexes with components added in cosine phase. However, the opposite effect was sometimes found. The direction of the phase effect varied in an unpredictable way across subjects and across type of harmonic complex.

Overall, these results suggest that people with cochlear damage depend relatively more on temporal information and less on spectral information than normally hearing people. The variability in the results across subjects, even in cases where the audiometric thresholds are similar, may occur partly because of individual differences in the auditory filters and partly because loss of neural synchrony is greater in some subjects than in others. Subjects in whom neural synchrony is well preserved may have good pitch discrimination despite having broader-than-normal auditory filters. Subjects in whom neural synchrony is adversely affected may have poor pitch discrimination regardless of the degree of broadening of their auditory filters.

The perception of pitch plays an important role in the ability to understand speech. The pitch patterns of speech indicate which are the most important words in an utterance, they distinguish a question from a statement, and they indicate the structure of sentences in terms of phrases. However, these aspects of pitch perception are not usually revealed in laboratory tests of speech intelligibility because such tests typically use either isolated words or nonsense syllables or sentences with a uniform structure. Also, the existence of pitch anomalies such as diplacusis may affect the enjoyment of music. There have been few, if any, studies on diplacusis for complex sounds, but it is likely to occur to some extent.

EFFECTS OF COCHLEAR DAMAGE ON SOUND LOCALIZATION AND BINAURAL HEARING

There are several advantages associated with having two ears. Firstly, differences in the amplitude or time of arrival of sounds at the two ears provide cues that contribute greatly to the ability to localize sound sources. Secondly, the ability to detect signals in noise can be improved by comparison of the stimuli reaching the two ears. Thirdly, when listening to a target sound such as speech in the

presence of background noise, the target-to-noise ratio may be much higher at one ear than at the other ear. Under these circumstances, people are able to make use of the ear receiving the higher target-to-noise ratio. Finally, even when the signals reaching the two ears are identical (diotic stimuli), the ability to discriminate or identify the signals is often slightly better than when the signals are delivered to one ear only (monaural stimuli). These advantages can be reduced by cochlear damage, but this does not always happen.

Sound Localization and Lateralization

Durlach, Thompson, and Colburn (1981) surveyed studies of localization and lateralization in hearing-impaired people. The majority of studies used either wideband noise or filtered noise as stimuli. Durlach et al. conclude that many of the studies were hard to interpret because they did not distinguish between systematic errors in localization and poor resolution. Nevertheless, there was a clear trend for poor localization and lateralization in people with unilateral or asymmetrical cochlear damage. Subjects with symmetrical cochlear losses often showed near-normal performance, especially when tested at reasonably high sound levels.

More recent studies (Gabriel, Koehnke, & Colburn, 1992; Häusler, Colburn, & Marr, 1983; Kinkel, Kollmeier, & Holube, 1991; Smoski & Trahiotis, 1986) show that binaural performance can vary markedly across subjects. Subjects with unilateral or asymmetric losses tend to show larger than normal thresholds for detecting interaural time differences (ITDs) and interaural intensity differences (IIDs). Subjects with symmetrical losses sometimes show normal or near-normal localization for broadband noise stimuli. However, they often show impaired performance for narrowband stimuli. It is possible, as pointed out by Colburn and Trahiotis (1992), that good performance for a restricted frequency range may be enough to ensure good performance for broadband stimuli.

The poor discrimination of ITDs, when it occurs, may be the result of several factors. Firstly, it may be partly related to the relatively low SL of the stimuli; ITD discrimination in normally hearing subjects worsens markedly below about 20 dB SL (Häusler, Colburn, & Marr, 1983). Secondly, it may result from abnormalities in the travel time of the traveling wave along the BM or in the phase of spike initiation and from differences in travel time or phase of spike initiation between the two ears (Ruggero & Rich, 1987; Ruggero, Rich, & Recio, 1993). Thirdly, it may be the result of abnormalities in phase locking.

Abnormalities in IID discrimination may have multiple causes also. Firstly, as with ITD discrimination, they may result from the relatively low SL of the stimuli. Secondly, they may result from abnormal intensity coding and from differences in intensity coding at the two ears. In this context it is noteworthy that IID discrimination in normally hearing subjects can be markedly impaired by putting an earplug in one ear (Häusler, Colburn, & Marr, 1983).

Some people with cochlear damage have essentially no ability to use spectral cues provided by pinna transformations (Häusler, Colburn, & Marr, 1983). This may happen either because the cues are inaudible or because the patterns of spectral peaks and dips cannot be resolved. The lack of pinna cues creates difficulty in deciding whether a sound comes from in front or behind and from above or below. It should be noted that pinna cues are drastically altered or removed altogether by hearing aids; hearing aids alter the spectral patterns at the eardrum and usually do not amplify sounds in the frequency range above 6 kHz, where pinna cues are most effective.

Binaural Masking Level Differences (MLDs) for People with Cochlear Damage

The binaural MLD refers to the improvement in detection of a signal in noise that occurs when the signal and noise differ in ITD or IID (Moore, 1989). Durlach, Thompson, and Colburn (1981) surveyed studies of the MLD using hearing-impaired subjects. Although there was considerable variability in the results across studies, generally it was found that cochlear hearing loss leads to reduced MLDs, even when the hearing loss is reasonably symmetrical. For example, Quaranta and Cervellera (1974) report abnormally small MLDs in 86% of cases. Several more recent studies confirm that MLDs are typically smaller than normal in subjects with cochlear hearing loss (Hall, Tyler, & Fernandes, 1984; Jerger, Brown, & Smith, 1984; Kinkel, Kollmeier, & Holube, 1991; Staffel, Hall, Grose, & Pillsbury, 1990). These studies show a trend for smaller MLDs in subjects with higher absolute thresholds, although the association is not generally strong, and subjects with similar absolute thresholds can have very different MLDs. MLDs also tend to decrease with increasing asymmetry of the loss (Jerger et al., 1984).

Head Shadow Effects—Selecting the Better Ear

When listening for a signal in background noise in everyday situations, it is often the case that the signal to noise ratio is much better at one ear than at

the other. An advantage of having two ears is that the listener can effectively "select" the ear giving the higher signal to masker ratio.

Bronkhorst and Plomp (1988) studied the effects of ITDs and head shadow on the intelligibility of speech in noise under binaural listening conditions. The sounds were recorded using a realistic model of a human head and torso (KEMAR; see Burkhard, 1975). The speech was presented from directly in front of KEMAR, while the noise was presented at seven azimuths ranging from 0° (frontal) to 180°. The noise had the same long-term average spectrum as the speech. The recorded sounds were digitally processed to derive two signals, one containing only ITDs (identical intensity at the two ears at all frequencies) and the other containing only IIDs due to head shadow. These stimuli were presented via earphones. The speech to noise ratio required for 50% of sentences to be understood (the speech reception threshold, SRT) was determined for each noise azimuth. The decrease in SRT, relative to the case where both speech and noise came from 0° azimuth, is called the binaural intelligibility level difference (BILD). The BILD due to ITDs was between 3.9 and 5.1 dB (for noise azimuths between 30° and 150°), whereas the BILD due to IIDs was 3.5 to 7.8 dB. When both ITDs and IIDs were present (i.e., when the unprocessed stimuli were used), the improvements were larger still, ranging from 5.8 to 10.1 dB. The presence of IIDs decreased the effectiveness of the masking release due to ITDs. In additional experiments where the stimulus to one ear was turned off, Bronkhorst and Plomp showed that the advantage gained from IIDs mainly depends on the ear receiving the highest speech to noise ratio. However, this advantage decreases when the noise in the other ear is fairly loud.

In summary, spatial separation of speech and background noise can lead to a BILD of up to 10 dB. Most of this effect (7 to 8 dB) is due to the fact that the speech to noise ratio is improved at one ear by head shadow effects. A small part (2 to 3 dB) is due to binaural processing of ITDs.

Bronkhorst and Plomp (1989) carried out similar experiments to those described above using 17 subjects with symmetrical hearing losses and 17 subjects with asymmetrical losses (differences in threshold between the two ears, averaged over 500, 1000, and 2000 Hz, ranging from 5 to 31 dB). Most subjects were diagnosed as having mild to moderate cochlear hearing losses. The noise level was adjusted for each subject so as to be as far as possible above absolute threshold without being uncomfortably loud. When the speech and noise both came from 0° azimuth, the SRTs were, on average, 2.5 dB higher than found for normally hearing subjects. The hear-

ing-impaired subjects showed 2.6 to 5.1 dB less binaural gain than normal when the noise azimuth was changed to 90°. In other words, in this condition SRTs were 5.1 to 7.6 dB higher than normal, a considerable difference.

The BILDs due to IIDs alone ranged from 0 dB to normal values of 7 dB or more. The size of BILD depended on the high-frequency loss in the ear receiving the higher speech to noise ratio; greater high-frequency losses were associated with a reduced advantage. This makes sense because head shadow effects are greatest at high frequencies, and if those high frequencies are inaudible, little advantage can be gained. The average BILD due to ITDs alone was nearly normal (4.2 dB as compared with 4.7 dB for normally hearing subjects) for subjects with symmetrical hearing losses. However, subjects with asymmetrical losses showed smaller BILDs, averaging 2.5 dB. When ITDs were introduced in stimuli already containing IIDs, the gain was 2 to 2.5 dB for both groups, comparable with what was obtained for normally hearing subjects.

In summary, subjects with cochlear hearing loss are generally less able than normal to take advantage of spatial separation of speech and interfering noise. When tested under conditions where speech and noise are spatially separated, they perform more poorly, relative to normal, than when the speech and noise come from the same position in space. The disadvantage appears to arise mainly from the inaudibility of high frequencies in the ear at which the speech to noise ratio is highest.

EFFECTS OF COCHLEAR DAMAGE ON SPEECH PERCEPTION

One of the most common complaints of people with cochlear hearing loss is difficulty understanding speech. There has been considerable controversy in the literature about the reasons for this difficulty. Some researchers suggest that the difficulty arises primarily from reduced audibility; for a given speech level, the proportion of the speech spectrum which is above threshold is less than for normal listeners (Humes, Dirks, & Kincaid, 1987; Lee & Humes, 1993; Zurek & Delhorne, 1987). Other researchers (Dreschler & Plomp, 1980 and 1985; Glasberg & Moore, 1989; Plomp, 1978 and 1986) argue that the difficulty arises, at least partly, from changes in the perception of sounds which are well above the absolute threshold. Some of these changes have been reviewed in this paper. I will argue that, for losses up to about 45 dB, audibility is the single most important factor. However, for greater losses, poor discrimination of suprathreshold (audible) stimuli is also of major importance.

The Magnitude of the Noise Problem

Plomp (1994) reviewed several studies which measured the SRT for sentences presented in a continuous speech-shaped noise. For high noise levels, people with cochlear damage had higher SRTs than did normally hearing people. The increase in SRT varied from about 2.5 dB for people with mild hearing losses caused by noise exposure or associated with aging, to about 7 dB for people with moderate to severe losses caused by Ménière's syndrome or by unknown pathologies. An elevation in SRT of 2.5 dB is sufficient to create a substantial loss of intelligibility in difficult listening situations.

The elevation in SRT can be much greater when a fluctuating background noise or a single competing talker is used instead of a steady noise. Normally hearing subjects are able to take advantage of temporal and spectral "dips" in the interfering sound to achieve a much lower SRT than when steady background noise is used (Baer & Moore, 1994; Duquesnoy, 1983; Eisenberg, Dirks, & Bell, 1995; Festen & Plomp, 1990; Hygge, Rönnerberg, Larsby, & Arlinger, 1992; Moore, Glasberg, & Vickers, 1995). For normal-hearing subjects, the SRT when the background is a single talker is 7 to 18 dB lower than when the background is speech-shaped noise. People with cochlear damage appear to be less able than normally hearing people to take advantage of the temporal and spectral dips. For hearing-impaired subjects, SRTs are not greatly different for a steady noise background than for a single talker background (Duquesnoy, 1983; Eisenberg et al., 1995; Festen & Plomp, 1990; Hygge et al., 1992). Hence, when the background is a single talker, the SRT is 9 to 25 dB higher for people with cochlear damage than for normally hearing people. This represents a very large deficit.

Finally, as described earlier, people with cochlear damage are less able than normally hearing people to take advantage of spatial separation of the target speech and the interfering sound(s). This can lead to a further elevation in SRT, relative to that found for normally hearing subjects, of about 7 dB.

In summary, in some listening situations common in everyday life, such as trying to listen to one person when another person is talking, people with cochlear damage may require speech to background ratios 16 dB or more higher than normal (Duquesnoy, 1983). This represents a very substantial problem. However, the majority of laboratory experiments show a less severe problem, as they have used steady speech-shaped noise coming from the same direction as the target speech as a background sound.

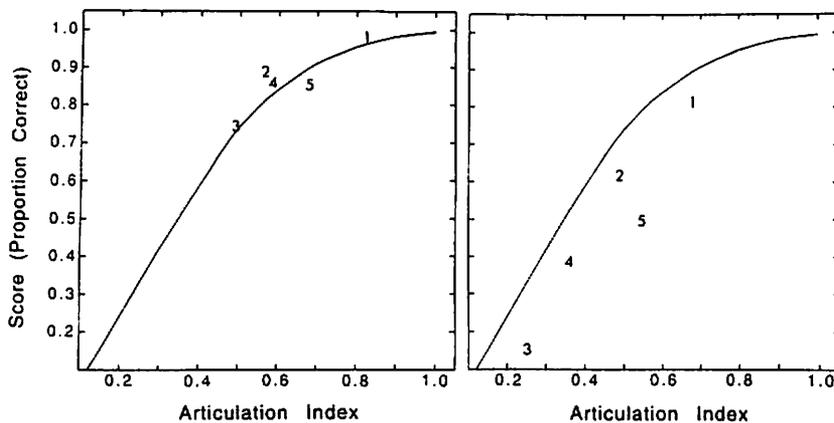


Figure 8. Comparison of speech recognition scores of hearing-impaired subjects with predictions based on the AI. Each number represents the mean score across subjects for a specific condition of filtering/background noise. For subjects with mild losses, the predictions are accurate (left panel); for subjects with more severe losses, the obtained scores fall below the predicted values (right panel). Redrawn from Pavlovic (1984).

The Role of Audibility

There is no doubt that audibility is crucial for speech intelligibility; if part of the speech spectrum is below the absolute threshold or is masked by background sound, then information is lost and intelligibility will suffer to some extent. The Articulation Index (AI) provides a way of quantifying the role of audibility (Fletcher, 1952; French & Steinberg, 1947; Kryter, 1962; Pavlovic, 1984). The AI is based on the assumption that speech intelligibility is uniquely related to a quantity that, for a normally hearing person, can be calculated from the long-term average spectra of the speech and background sound reaching the ear of the listener. Several researchers have examined the question of whether the AI can be used to predict speech intelligibility for hearing-impaired listeners. Although a few researchers have reported accurate predictions using the AI (Aniansson, 1974; Lee & Humes, 1993), most studies have shown that speech intelligibility is worse than would be predicted by the AI (Dugal, Braida, & Durlach, 1978; Fletcher, 1952; Pavlovic, 1984; Pavlovic, Studebaker, & Sherbecoe, 1986; Smoorenburg, 1992), especially for listeners with moderate or severe losses. The predictions are often quite accurate for those with mild losses.

An example of the results of this type of experiment is given in Figure 8; the data are from Pavlovic (1984). The speech materials were word lists presented under various conditions of filtering (broadband, lowpass, and highpass), either with or without background white noise, at a speech to noise ratio of 10 dB. Sixteen subjects with noise-induced hearing loss were used. For the eight subjects with the mildest losses (thresholds better than 50 dB HL at 4000 Hz), the mean scores across subjects for the five different conditions, indicated by the numbers in the left panel of the figure, are close to the predictions based on the AI (solid curve). For the eight subjects with more severe losses (thresholds 55 dB HL or worse at 4000 Hz), the mean scores, shown

in the right panel, fall consistently below the predicted values.

Overall, the results from studies using the AI suggest that, although audibility is of major importance, it is not the only factor involved, at least for people with moderate to severe cochlear losses. Additionally, it should be noted that the AI does not give accurate predictions of speech intelligibility under conditions where the background noise is fluctuating (e.g., when there is a single background talker) because it is based on the long-term average spectrum of the background noise.

Another way to evaluate the importance of audibility is to measure the SRT in noise as a function of overall noise level. If the performance of hearing-impaired subjects is limited by part of the speech spectrum being below absolute threshold, then the SRT, expressed as a speech to noise ratio, should decrease progressively with increasing noise level; as the noise level is increased, the speech level has to be increased also, and so more and more of the speech spectrum should lie above absolute threshold. Furthermore, the SRT for hearing-impaired subjects should approach that for normally hearing subjects at sufficiently high noise levels.

One study, using 20 elderly subjects with mild hearing losses (mean threshold about 20 dB HL at medium frequencies, increasing to 40 dB HL at 4000 Hz), reports results consistent with these predictions (Lee & Humes, 1993). However, the results of most other studies do not show the predicted pattern: the SRT for hearing-impaired subjects remains approximately constant for noise levels above a certain value, and the SRT at high noise levels remains greater for hearing-impaired than for normally hearing subjects (Plomp, 1978, 1986, 1994; Smoorenburg, 1992).

Hearing-impaired people with mild to moderate cochlear hearing loss do not generally have difficulty in understanding connected discourse in a quiet, nonreverberant room. However, they may have

some difficulty for isolated nonsense syllables. Subjects with severe losses can have considerable difficulty with speech in quiet. Turner and Robb (1987) tried to determine whether this difficulty could be explained in terms of audibility. They studied the identification of synthetic consonant-vowel syllables composed of one of the six stop consonants /b, d, g, p, t, k/ followed by the vowel /a/. A model of filtering in the peripheral auditory system was used to estimate the portion of the speech spectrum that was above the threshold of audibility for a given presentation level. Several presentation levels were used for each subject. They tested four normally hearing subjects and five subjects with moderate to severe hearing losses.

For the normally hearing subjects, recognition improved monotonically as the audible portion of the stimulus increased, and performance was perfect when most of the spectrum was above threshold. For four out of the five hearing-impaired subjects, recognition scores were consistently lower than for the normally hearing subjects at a given level of audibility. For these subjects scores remained below 100%, even in cases where audibility was close to 100%. For the remaining subject, the hearing loss at low frequencies was too severe for even 50% audibility to be achieved. Overall, these results strongly suggest that reduced audibility is not sufficient to explain the relatively poor consonant recognition of the hearing-impaired subjects. It is noteworthy, however, that even presentation levels of 100 dB SPL were not sufficient to provide 100% audibility for subjects with severe losses (although appropriate frequency-dependent amplification could ensure 100% audibility at overall levels below 100 dB SPL).

Taken together, the results reviewed above strongly suggest that one or more factors other than audibility contribute to the difficulties of speech perception experienced by those with moderate or greater cochlear losses. This is especially true in situations where the stimuli are presented at high levels and/or in background noise. In other words, the difficulties arise partly from abnormalities in perception of sounds that are above the threshold of audibility. For those with mild losses, audibility may be the dominant factor.

THE USE OF SIMULATIONS TO ASSESS THE IMPORTANCE OF PSYCHOACOUSTIC FACTORS IN SPEECH PERCEPTION

Another approach to assessing the importance of suprathreshold discrimination abilities for speech perception is to simulate the effect of one specific aspect of hearing impairment by processing the stimuli in a way that mimics the effect of this aspect.

The processed stimuli are then used in tests with normally hearing subjects. Provided the simulation is accurate, this makes it possible to study the effect of that aspect in isolation.

Simulating the Effects of Recruitment

The effects of loudness recruitment can be simulated by splitting the signal into a number of frequency bands and applying dynamic range expansion in each band before recombining the bands. Effectively, this simulation does the opposite of what happens in a normal cochlea, where stimuli are filtered along the basilar membrane and fast-acting compression is applied at each CF. The goal of the simulation is to process the level of the stimulus in each band, on a moment-by-moment basis, so as to create loudness sensations in a normal ear that would resemble those produced in an impaired ear with recruitment.

Villchur (1974) used a three-band system to process speech so as to simulate the effects of recruitment associated with severe hearing loss. The stimulus in each band was processed using a fast-acting expander. Subjects with unilateral hearing impairments judged processed stimuli presented to their normal ear to sound "similar" or "very similar" to unprocessed stimuli presented to their impaired ear. The intelligibility of the processed speech was not measured in formal tests. However, Villchur concluded that recruitment is a sufficient cause for loss of intelligibility in cases of severe hearing loss.

In a later study, Villchur (1977) used a 16-channel system with computer-controlled attenuators to achieve the dynamic range expansion. A severe, sloping hearing loss was simulated. The intelligibility of speech in quiet at a level of 94 dB was adversely affected by the processing for both isolated words and sentences. The intelligibility of words and sentences in white noise was also adversely affected by the processing; the percent correct decreased from about 69% to 50%. An even greater effect was found for speech in speech-shaped noise. It should be noted that the deleterious effects of the processing may be partly attributed to reduced audibility of the speech rather than to distorted loudness relationships. Villchur's main point is that severe recruitment makes it difficult or impossible to restore audibility of the weaker parts of the speech by linear amplification without the more intense parts becoming uncomfortably loud.

Duchnowski (Reference Note 1) used digital signal processing to implement a 14-band system. The gain in each band was adjusted dynamically to reproduce the elevated absolute thresholds and abnormal loudness growth associated with cochlear

hearing impairment. In one experiment, the processing was adjusted to simulate the hearing losses of the subjects tested by Zurek and Delhorne (1987); these subjects had mild to moderate hearing losses. The test stimuli were consonant-vowel syllables presented in quiet and in various levels of speech-shaped noise. Generally, the pattern of results obtained using the simulation with normally hearing subjects matched the pattern obtained in the impaired subjects of Zurek and Delhorne. Thus, for these subjects, the threshold elevation and associated loudness recruitment appear sufficient to account for their difficulties in understanding speech.

In a second experiment, the hearing loss of subjects with more severe losses was simulated. When a flat frequency-gain characteristic was used, the pattern of results from the impaired subjects closely matched the results from the normal subjects listening to the processed stimuli. However, when high-frequency emphasis was employed to increase the audibility of high-frequency components, the impaired subjects generally performed more poorly than their normal counterparts listening to processed stimuli. This suggests that some factor other than threshold elevation and loudness recruitment contributed to the speech perceptual problems of the hearing-impaired subjects.

Moore and Glasberg (1993) simulated threshold elevation and loudness recruitment by splitting the input signal into 13 frequency bands and processing the envelope in each band so as to create loudness sensations in a normal ear that would resemble those produced in an impaired ear with recruitment. They simulated three types of hearing loss: flat moderate (a 50 dB loss at all frequencies), flat severe (a 67 dB loss at all frequencies), and sloping (a 33 dB loss at low frequencies increasing to 67 dB at high frequencies). Examples of sounds processed in this way are available on a CD (Reference Note 5). Moore and Glasberg also assessed whether there are deleterious effects of recruitment after the attenuative component of a hearing loss has been corrected as far as possible by linear amplification, as would be used in a conventional hearing aid. To do this, they ran a set of conditions in which the stimuli were subjected to the frequency-dependent gain recommended by the National Acoustic Laboratories' (NAL) revised procedure (Byrne & Dillon, 1986).

For speech in quiet, the simulation produced, as expected, a reduction in the ability to understand low-level speech. However, speech at sufficiently high levels was highly intelligible in all conditions. Also, linear amplification according to the NAL prescription gave high intelligibility for speech at normal conversational levels. Thus, linear amplification was rather effective in improving the intelli-

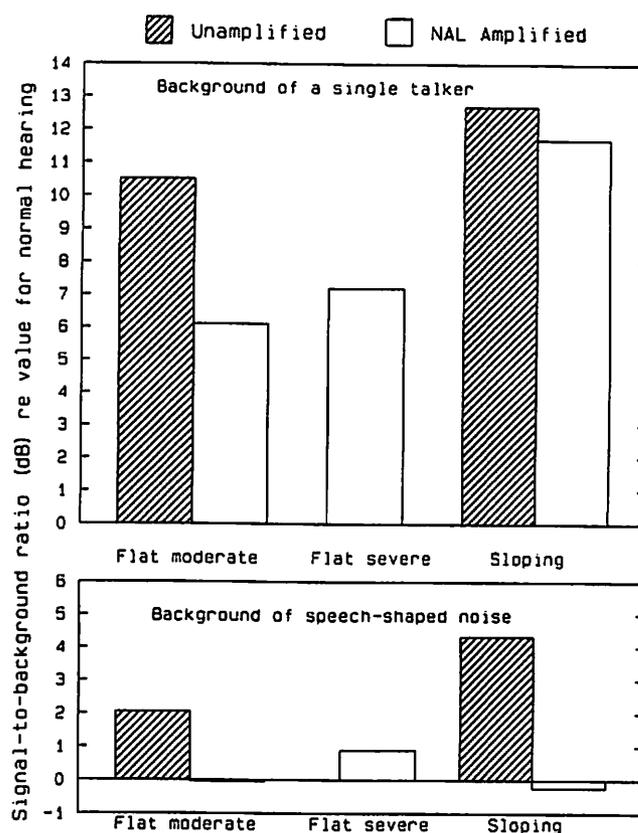


Figure 9. Results of Moore and Glasberg (1993) (upper panel) and Moore et al. (1995) (lower panel). The speech-to-background ratio needed for a given level of intelligibility is plotted relative to the value for normally hearing subjects. No score is shown for the condition simulating a severe flat loss without amplification because the target speech was inaudible for that condition.

gibility of speech in quiet, although it did not allow speech to be both intelligible and comfortable over a wide range of sound levels; speech with input levels above about 70 dB SPL was judged to be unpleasantly loud in the conditions involving linear amplification.

For speech presented at a fixed input level of 65 dB SPL against a background of a single competing talker, simulation of hearing loss produced substantial decrements in performance. This is illustrated in the upper panel of Figure 9, which shows the signal-to-background ratio required to achieve a given level of intelligibility relative to the ratio required for the same intelligibility in the control condition. Linear amplification according to the NAL prescription improved performance markedly for the conditions simulating flat losses but was less effective for the condition simulating a sloping loss. Performance in this condition remained well below that in the control condition, even after linear amplification. A possible explanation for this is that the NAL-prescribed gains for the condition with the

sloping loss were 4 to 10 dB less at high frequencies than the gains for the flat severe loss, even though the high-frequency absolute thresholds were similar for the two cases. The gain for the condition with the sloping loss may have been insufficient to restore audibility of weak high-frequency sounds such as "p" and "k."

In a second study, Moore, Glasberg, and Vickers (1995) used similar processing conditions, but the speech was presented in a background of speech-shaped noise instead of a single competing talker. The input level of the speech was fixed at 65 dB SPL, while the level of the background noise varied from 65 to 74 dB SPL. For the conditions simulating threshold elevation and recruitment, the speech to noise ratios had to be higher than in the control condition to achieve similar levels of performance; this is illustrated in the lower panel of Figure 9. However, when linear amplification according to the NAL prescription was applied before the simulation, performance improved markedly and became close to that for the control condition.

The differences between the results using speech-shaped noise and those using a single talker as the interfering sound can be understood in the following way. People with normal hearing can take advantage of spectral and temporal dips in a background of a single competing talker. Hence, the speech-to-background ratio required to achieve a given level of performance is markedly lower than when the background is speech-shaped noise. However, "dip listening" requires a wide dynamic range. Loudness recruitment, either real or simulated, reduces the available dynamic range. If the level of the speech is set so that the more intense parts of the speech are comfortably loud, the weaker parts may be inaudible. Therefore, people with recruitment cannot exploit "dip listening" as effectively as normally hearing people can. When a background of speech-shaped noise is used, "dip listening" is of much less importance because the noise does not contain dips of sufficient magnitude or duration. Hence, speech intelligibility depends more on the higher-level portions of the speech, and these are less affected by reduced dynamic range. Furthermore, linear amplification, which ensures that the higher level portions of speech are audible, is effective in compensating for the simulated recruitment, except when the hearing loss is severe.

Simulating Reduced Frequency Selectivity

To simulate reduced frequency selectivity, the basic idea has been to smear or broaden the spectra of test signals so that the excitation pattern produced in a normal ear resembles the pattern that

would be produced in an impaired ear using unprocessed signals. Such spectral smearing does not simulate all of the effects of reduced frequency selectivity. Specifically, the time patterns at the outputs of the auditory filters are affected by reduced frequency selectivity in a way that is not reproduced by the simulations. Instead, the simulations alter the time patterns of the stimuli in a complex way that is a byproduct of the specific processing used. Essentially, the simulation may be regarded as mimicking the consequences of reduced frequency selectivity in terms of place coding in the auditory system but not in terms of time coding.

In recent studies, digital signal processing techniques have been used to perform spectral smearing. Generally, short segments of the signal are transformed into spectra using a short-term fast Fourier transform (FFT). Modifications are then made in the spectral domain, and the modified spectra are transformed back into time signals using an inverse FFT. This is repeated for a series of overlapping segments, and the resulting processed segments are added together. This is referred to as the overlap-add method (Allen, 1977). The method has been used in several studies (Baer & Moore, 1993 and 1994; Celmer & Bienvenue, 1987; Howard-Jones & Summers, 1992; ter Keurs, Festen, & Plomp, 1992 and 1993).

ter Keurs, Festen, and Plomp (1992) smeared the spectral envelope of each segment while leaving its fine structure unchanged. They smeared the spectrum of speech and noise separately and then added the speech and noise together. The SRT in noise increased once the smearing bandwidth was increased beyond about 1/3 octave. Vowel identification was affected more than consonant identification.

In a second study, ter Keurs, Festen, and Plomp (1993) compared the effects of the smearing using either speech-shaped noise or a single talker as the background sound. For both types of masker, SRTs increased when the smearing bandwidth was increased. For unsmearing speech, SRTs were 5 to 7 dB lower when the background was a single talker than when it was speech-shaped noise. This difference decreased as the smearing bandwidth was increased. Hence, the effect of the spectral smearing on SRTs was greater for the speech masker than for the noise masker.

Baer and Moore (1993) measured the intelligibility of speech in quiet and in speech-shaped noise using the overlap-add method to perform smearing of the magnitude spectrum. When speech in noise was used, the speech was mixed with the noise before processing. The procedure for simulating impaired frequency selectivity used a realistic form of

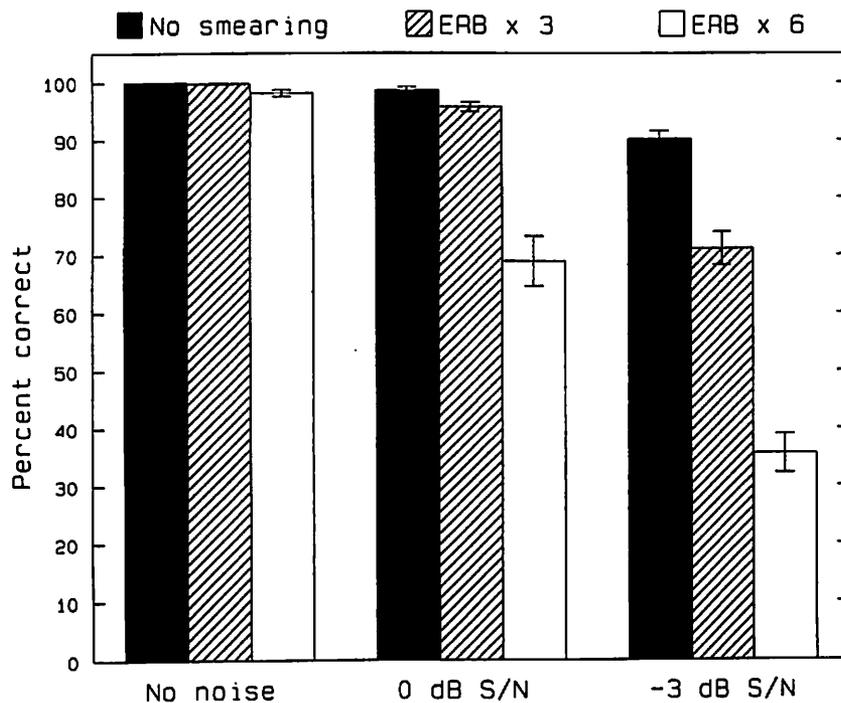


Figure 10. Results of Baer and Moore (1993) showing the percentage of key words correct for three amounts of spectral smearing and three different amounts of background noise. The three amounts of smearing were: no smearing (solid bars); simulation of auditory filters with bandwidths three times greater than normal (diagonally shaded bars); and simulation of auditory filters with bandwidths six times greater than normal (open bars). Error bars indicate ± 1 SE.

spectral smearing based on measured characteristics of auditory filters in hearing-impaired subjects. Several different types of smearing were used, simulating specific degrees of broadening and asymmetry of the auditory filter. Examples of sounds processed in this way are available on a CD (Reference Note 5). Some of the results are illustrated in Figure 10. The intelligibility of speech in quiet (the three left-most bars in Fig. 10) was hardly affected by spectral smearing, even for smearing that simulated auditory filters six times broader than normal. However, this may partly reflect a ceiling effect, as scores were close to perfect in all conditions. The intelligibility of speech in noise was adversely affected by the smearing, especially for large degrees of smearing and at a low speech to noise ratio (-3 dB; see the three right-most bars in Figure 10). Simulation of asymmetrical broadening of the lower side of the auditory filter had a greater effect than did simulation of asymmetrical broadening of the upper side, suggesting that upward spread of masking may be particularly important.

In a second study, Baer and Moore (1994) used a single competing talker as the background instead of speech-shaped noise. The results were similar in form to those found using speech-shaped noise. Specifically, performance worsened with increasing smearing, and the worsening was greater at the more adverse speech to background ratio. The results agree with those of ter Keurs, Festen, and Plomp (1993) in showing that the deleterious effects of spectral smearing were greater for a speech masker than for a noise masker. Hence, the differ-

ence in masking produced by speech and noise maskers was less for spectrally smeared than for unprocessed stimuli. This is consistent with the finding noted earlier that, for normally hearing subjects, SRTs are lower when the background is speech than when it is noise, whereas for hearing-impaired subjects the difference is smaller or absent.

In summary, the results of experiments on spectral smearing suggest that reduced frequency selectivity does contribute significantly to the difficulties experienced by people with cochlear damage in understanding speech in the presence of background sounds.

APPLICATIONS OF THE FINDINGS TO HEARING AID DESIGN

The Use of Linear Amplification to Restore Audibility

The primary goal of most hearing aids is relatively simple, namely to restore audibility via frequency-selective amplification. Many hearing aids operate essentially as linear amplifiers; over most of their operating range they apply a gain that is independent of level. It became apparent very soon after hearing aids first came into use that it was not practical to use linear amplification to compensate fully for the loss of audibility caused by cochlear damage. The major factor preventing this was loudness recruitment and the associated reduced dynamic range. Say, for example, a person had a

cochlear hearing loss of 60 dB at all frequencies. The highest comfortable level (HCL) for such a person would typically be about 90 to 100 dB HL. A hearing aid that fully compensated for the loss of audibility would apply a gain of 60 dB at all frequencies. However, that would mean that any sound with a level above about 40 dB HL would be amplified to a level exceeding the HCL. In practice, many sounds encountered in everyday life would become unpleasantly loud.

Most hearing aids incorporate a way of limiting the maximum output of the aid so as to avoid discomfort to the user. In many hearing aids this is achieved by electronic peak clipping in the output stage of the aid. Such clipping introduces unpleasant-sounding distortion (Crain & van Tasell, 1994), and in practice most users of hearing aids set the volume control to avoid clipping in everyday listening situations. Even when aids include output limiting, it has been found to be impractical to compensate fully for loss of audibility. Rather, various rules have been developed (Byrne & Dillon, 1986; Lybarger, 1978; McCandless & Lyregard, 1983) that prescribe a gain between one-third and one-half of the hearing loss. Such rules often have the aim of amplifying speech at normal conversational levels so that it is both audible and comfortable in all frequency regions. However, even if the frequency-gain characteristic is appropriate for a "typical" talker, it is unlikely to be optimal for other talkers, for different overall speech levels, or for other (nonspeech) sounds.

A related problem with linear hearing aids is that users often find it necessary to adjust the volume control to deal with different listening situations. The overall level of speech and other sounds can vary considerably from one situation to another (Pearsons, Bennett, & Fidell, Reference Note 6), and people with cochlear damage do not have sufficient dynamic range to deal with this.

The Use of Compression to Compensate for Reduced Dynamic Range

It was suggested many years ago that problems associated with reduced dynamic range could be alleviated by the use of automatic gain control (AGC) (Steinberg & Gardner, 1937). With AGC it is possible to amplify weak sounds more than stronger ones, which results in the wide dynamic range of the input signal being compressed into a smaller dynamic range at the output. Hence, AGC systems are also called "compressors." Although this idea sounds simple, in practice there are many ways of implementing AGC, and there is still no clear consensus as to the "best" method, if there is such a thing.

There is also considerable controversy about the efficacy of AGC systems.

AGC systems have been designed in many different forms, mostly on the basis of different rationales or design goals. For reviews, see Moore (1990) and Hickson (1994). Some systems are intended to adjust the gain automatically for different listening situations. The idea is to relieve the user of the need to adjust the volume control to deal with these situations. Usually such systems change their gain slowly with changes in sound level; this is achieved by making the recovery time of the AGC circuit rather long (greater than a few hundred milliseconds). These systems are often referred to as "automatic volume control" (AVC). Although it is generally accepted that AVC can be useful, relatively few commercial hearing aids incorporate AVC. One reason is that, after a brief intense sound such as a door slamming, the gain drops and stays low for some time; the aid effectively goes "dead." This problem can be alleviated by using an AGC circuit with dual time constants (Moore, Glasberg, & Stone, 1991).

AGC is often used primarily to limit the maximum output of hearing aids, to prevent discomfort and/or circuit overload at high input sound levels. The compression ratio is usually very large, and the compression threshold is usually high. Such systems are known as "compression limiters." Compression limiters usually have a small attack time (<5 msec) so as to respond rapidly to sudden increases in sound level. The recovery time is also usually fairly small (20 to 100 msec). The function of compression limiters is similar to that of the peak clippers described earlier. However, peak clipping causes unpleasant-sounding distortion, whereas the effects of compression limiting are not so noticeable. Hence, compression limiters are quite widely used in hearing aids.

An alternative type of compressor with lower compression ratios and lower compression thresholds has been used in hearing aids in attempts to make the hearing-impaired person's perception of loudness more like that of a normal listener and to ensure that the weaker consonant sounds of speech will be audible without the more intense sounds (e.g., vowels) becoming uncomfortably loud. Such compressors usually have short time constants (typically 20 to 100 msec) and are often referred to as "syllabic compressors" because the gain changes over times comparable to the durations of individual syllables.

Several authors (Laurence, Moore, & Glasberg, 1983; Mangold & Leijon, 1979; Villchur, 1973) have proposed that syllabic compression should be applied separately in two or more frequency bands. There are at least two reasons why this might be

beneficial. First, the amount of hearing loss, and usually the amount of recruitment, often varies markedly with frequency; typically, hearing loss and recruitment are both greater at high frequencies than at low. Hence, the amount of compression needed varies with frequency, and this requires that compression be applied independently in more than one band. Second, relatively weak high-frequency components in speech (e.g., k, p, t), which can be important for intelligibility, are often accompanied by, or follow rapidly after, relatively intense low-frequency components. The use of fast-acting AGC in two or more separate bands can ensure that these weak high-frequency components are always audible.

Another rationale for the use of multiband compression as opposed to single-band (broadband) compression is to reduce the effects of interfering sounds (Ono, Kanzaki, & Mizoi, 1983; Rankovic, Freyman, & Zurek, 1992; van Dijkhuizen, Festen, & Plomp, 1991). Consider a hypothetical situation in which a hearing-impaired person is trying to understand speech in the presence of an intense narrowband noise. A broadband AGC system would reduce its gain in response to the intense noise, thereby reducing the audibility of the speech. A multiband AGC system would reduce the gain only in the frequency region of the noise, reducing the masking effect and loudness of the noise without affecting the audibility of the parts of the speech spectrum remote from the frequency of the noise. Van Dijkhuizen et al. (1991) argue that the gain in such systems should change slowly with time.

Several commercial hearing aids incorporate a crude form of multiband AGC based on this rationale. Usually they have only two bands, and the gain in the low-frequency band is reduced when the input level exceeds a certain value. These aids, which are sometimes misleadingly called automatic signal processing (ASP) aids, are based on the assumption that most intense environmental sounds have their energy concentrated at low frequencies. Reduction of low-frequency gain can, in principle, reduce the upward spread of masking from this noise. Evaluations of such aids have given mixed results. Where benefits have been found, they can probably be attributed at least partly to the fact that the ASP circuit reduces distortion that would otherwise be present at the output of the aids for high input sound levels (van Tasell & Crain, 1992).

Research on the Benefits of Multiband Compression

Research into the benefits of multiband compression has given conflicting results. The conflict arises

partly from differences in the way that the compression systems have been implemented and partly from differences in methods of evaluation. Individual differences among the subjects used may also have played a role. Comprehensive reviews of results using multichannel compression have been provided by Braida, Durlach, De Gennaro, Peterson, and Bustamante (1982) and by Hickson (1994). Hickson surveyed 21 studies of multichannel compression and notes that 10 of the studies provide evidence for benefits of the compression. Some general trends can be discerned from the results:

1. For speech in quiet, benefits of compression have often been found in a variety of systems when the speech materials used have covered a wide range of levels as occurs in everyday life (Laurence, Moore, & Glasberg, 1983; Lippmann, Braida, & Durlach, 1981; Moore, 1987; Moore & Glasberg, 1986a and 1988a; Moore, Glasberg, & Stone, 1991; Moore, Johnson, & Pluvina, 1992; Moore, Laurence, & Wright, 1985; Villchur, 1973). When the speech material has been presented at one reasonably high level, and when the speech material has been carefully equalized in level during the recording process (as was the case in many studies), compression does not show benefits over linear amplification (Lippmann et al., 1981).
2. For speech in background noise, benefits of fast-acting compression have sometimes (but not always) been found for systems with a small number of bands (channels) (Laurence, Moore, & Glasberg, 1983; Moore, 1987; Moore & Glasberg, 1986a and 1988a; Moore, Glasberg, & Stone, 1991; Moore, Johnson, & Pulvina, 1992; Moore, Laurence, & Wright, 1985; Ringdahl, Eriksson-Mangold, Israelsson, Lindkvist, & Mangold, 1990; Villchur, 1973). Benefits usually have not been found for systems with a large number of bands (Bustamante & Braida, 1987; Lippmann, Braida, & Durlach, 1981), although Yund and Buckles (1995) found some benefit of increasing the number of bands from one up to eight. There may be a disadvantage in using a very large number of bands, because this tends to reduce spectral contrasts in complex stimuli.
3. The extent of the benefit of compression for the understanding of speech in noise depends upon how the frequency-gain characteristic was chosen for the control condition using linear amplification. Under laboratory conditions using speech in noise with carefully controlled levels and fixed spectra, it may be possible to use linear amplification to make the speech audible and comfortable over a wide frequency

range. Under these conditions there may be little benefit of compression. However, if the linear condition is set up so that speech and/or environmental sounds with reasonably high input levels are not amplified to unpleasantly loud levels, then lower overall gains must be used. Under these conditions, benefits of compression may become apparent because the compression allows more amplification for weaker sounds while not amplifying intense sounds excessively.

4. Most (but not all) of the studies showing benefits of compression for listening to speech in noise have used aids that the subjects could wear in their everyday lives. This is important because if a person has had a hearing impairment for many years, it may take some time for them to learn to use the new cues provided by a compression system (Gatehouse, 1992).

It appears that compression can be beneficial in two ways. Firstly, it can allow speech in quiet to be understood over a wide range of sound levels without adjustment of the volume control and without the speech ever becoming uncomfortably loud. This can be achieved either by using a slow-acting front-end AGC (Moore, Glasberg, & Stone, 1991) or by using multichannel syllabic compression (Moore, Johnson, & Pluvillage, 1992). It may also be achievable using a wideband syllabic compressor, but I am not aware of experimental data indicating this. Secondly, compression can improve the intelligibility of speech in background noise. This only seems to occur when fast-acting compression is used in a small number of bands and when the gain in the control condition with linear amplification is chosen realistically (to reflect the gain that would be usable in everyday life). The improvement is generally small, amounting to a 1 to 3 dB reduction in the SRT in noise, but it is nevertheless worthwhile. There is some reason to believe that larger benefits might be obtained with a background sound of a single talker as opposed to the speech-shaped noise that has been used in many studies (Moore, Glasberg, & Vickers, 1995).

There is little evidence to support the idea that the benefits of compression accrue from restoring the perception of loudness to "normal." Indeed, if fast-acting compression is set up to do this, hearing aid users complain that everything sounds too "noisy." In practice, the aids must be set up so as to "under-compensate" for the loudness recruitment. Compression sufficient to restore loudness perception to "normal" appears to have deleterious effects on speech intelligibility and is not liked by users (Moore, Lynch, & Stone, 1992; Plomp, 1994). It seems likely that the benefits of compression arise

mainly from the fact that compression increases audibility while avoiding discomfort from loud sounds.

Attempts to Improve Speech Intelligibility Using Spectral Enhancement

Impaired frequency selectivity is probably at least partly responsible for the reduced ability of people with cochlear hearing loss to understand speech in noise. Linear amplification and multiband compression do not compensate for the effects of reduced frequency selectivity, although high-frequency emphasis can partially alleviate upward spread of masking. This section considers some forms of signal processing that attempt to compensate for reduced frequency selectivity. Such signal processing is not yet available in wearable hearing aids, but it may become available within the next few years as digital signal-processing chips become more powerful and consume less current.

If reduced frequency selectivity impairs speech perception, then enhancement of spectral contrasts (magnifying the differences between peaks and dips in the spectrum) might improve it for the hearing-impaired person. Several authors have described attempts to improve speech intelligibility for the hearing impaired by enhancement of spectral features. Many studies have reported little or no benefit of such enhancement, but in a few studies small benefits have been found. Simpson, Moore, and Glasberg (1990) describe a method of digital signal processing of speech in noise so as to increase differences in level between peaks and valleys in the spectrum. The processing involves manipulation of the short-term spectrum of the speech in noise using the overlap-add technique. Simpson et al. measured the intelligibility of sentences in speech-shaped noise using subjects with moderate cochlear hearing loss. The results show small but statistically significant improvements in speech intelligibility for the processed speech, typically of 6 to 7%.

Baer, Moore, and Gatehouse (1993) carried out an experiment similar to that of Simpson, Moore, and Glasberg (1990), but the amount of enhancement was systematically varied. Large amounts of enhancement produced decreases in the intelligibility of speech in noise. Performance for moderate degrees of enhancement was generally similar to that for the control condition. This rather disappointing result may have occurred because subjects did not have sufficient experience with the processed speech.

In a second experiment, subjects judged the relative quality and intelligibility of speech in noise processed using a subset of the conditions of the first

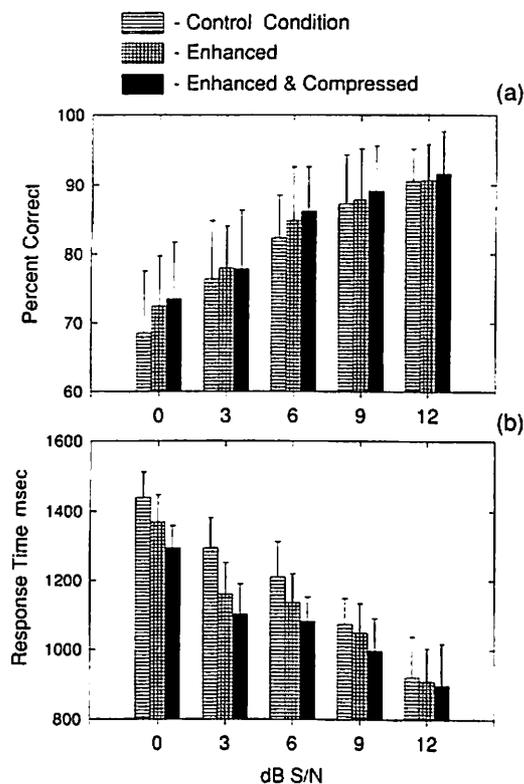


Figure 11. Results of Baer et al. (1993) for three conditions: a control condition, a condition using spectral enhancement, and a condition combining spectral enhancement with a moderate degree of compression. Results are shown for several speech-to-noise ratios. The upper panel (a) shows intelligibility scores, and the lower panel (b) shows response times.

experiment. Generally, processing with a moderate degree of enhancement was preferred over the control condition for both quality and intelligibility. Subjects varied in their preferences for high degrees of enhancement.

A third experiment used a modified processing algorithm with a moderate degree of spectral enhancement and examined the effects of combining the enhancement with broadband fast-acting compression. The intelligibility of speech in noise improved with practice, and, after a small amount of practice, scores for the condition combining enhancement with a moderate degree of compression were found to be significantly higher than for the control condition.

In a fourth experiment, Baer, Moore, and Gatehouse (1993) assessed performance using a sentence verification test (SVT) that measured both intelligibility and response times. The SVT can be administered repeatedly, and it includes a measure of response time which is probably related to ease of listening. They evaluated a subset of conditions from the third experiment. The results are illustrated in Figure 11. There were highly significant benefits of

the processing, which were revealed by increases in percent correct and decreases in response times. Spectral enhancement alone was superior to the control condition, and enhancement combined with compression was superior to enhancement alone. When expressed in terms of equivalent changes in signal to masker ratio, the benefits were about twice as great for the response time measures as for the identification scores, and they were also statistically more robust for the response time measures. This suggests that the major benefits of the processing may be in terms of increased ease of listening rather than in intelligibility. The overall effect of spectral enhancement combined with compression was equivalent to an improvement of speech to noise ratio by 4.2 dB.

In summary, studies of the effects of spectral enhancement have given mixed results. Some studies have shown no benefit, whereas others have shown modest benefits. However, the benefits can show up both in increased intelligibility of speech in noise (percent correct) and in decreased response times. The latter may reflect greater ease of listening. It should be noted that none of the studies on spectral enhancement has used wearable aids. Thus, subjects did not have extensive experience listening to the processed stimuli. It seems likely that such experience may be necessary to allow the full benefits of the processing to be measured (Gatehouse, 1992).

Some Concluding Remarks

Hearing aids have improved considerably in recent years. Aids are available with low distortion and with smooth wideband frequency responses. Many aids, especially the newer programmable aids, have a high degree of flexibility in shaping their frequency-gain characteristic. In principle, it is usually possible to tailor the frequency response of a linear aid to suit an individual patient. Hearing aids that offer reasonably effective compensation for reduced dynamic range are also available. In practice, many hearing-impaired people are still being fitted with hearing aids that have significant distortion, that have an irregular frequency response with an inappropriate shape, and that do not offer effective compensation for reduced dynamic range.

Even the best possible current hearing aids do not restore hearing to normal; especially when listening to speech in noise or competing speech, hearing-impaired subjects listening through hearing aids perform more poorly than do normally hearing people. This situation may be improved in the future through the use of directional microphones and through the use of digital signal processing to en-

hance spectral contrast and/or to improve speech to background ratios. Digital hearing aids also offer the possibility of very precise frequency response shaping and compensation for reduced dynamic range.

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A compact disc (CD) containing audio demonstrations illustrating many of the effects described in this paper may be obtained by writing to the author at Department of Experimental Psychology, University of Cambridge, Downing Street, Cambridge, CB2 3EB, England, enclosing a check for 20 dollars or 12 pounds sterling (payable to B.C.J. Moore). The CD includes simulations of the effects of loudness recruitment and threshold elevation, simulations of the effects of reduced frequency, and simulations of the combined effects of these factors. It also includes simulations of the effects of linear amplification and compression amplification. Finally, it demonstrates spectral enhancement processing and the occlusion effect.

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