Performance Requirements for Hearing Aids

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Abstract — The requirements for hearing aid frequency responses, SSPL characteristics, compression limiting, directional response, and internal noise are examined. On the basis of the assumptions made, aids would best serve the hearing impaired if they had flat very-low-frequency gains (and various mid- and high-frequency gains), SSPL responses adjustable in shape and level, directional responses, and internal noise levels below a recommended criterion level.

Introduction

The work reported in this paper had as its aim the determination of specifications for hearing aids to be issued by the National Acoustic Laboratories (NAL) of Australia. It examines the performance range needed in a family of hearing aids if a reasonable proportion of the hearing impaired population is to be adequately fitted. The performance range recommended is not far removed from that which can be achieved using existing components in a behind-the-ear hearing aid case. Since there is no unanimous agreement about the basis on which parameters such as gain, frequency response, maximum output, type of compression, and directionality of microphone should be selected, it is obviously impossible to be dogmatic about what constitutes the ideal range of hearing aids. The recommendations made here arise, in part, from the fitting procedures in current use at the National Acoustic Laboratories (NAL), in part from specially conducted experiments, and in part from consideration of basic theoretical principles.

Specifically, the required range of frequency responses is determined by applying the threshold-based selection procedure published by Byrne and Tonisson in 1976 (1) to a representative sample of audiograms. The saturated sound pressure level (SSPL) requirements were obtained by assuming that the SSPL should not exceed the client’s loudness discomfort level (LDL), and by performing LDL measurements with a hearing-aid-type transducer and coupling system on a representative sample of hearing impaired people. It is pointed out that the resulting complex SSPL requirements can be met with a single-channel hearing aid if the compression circuitry is appropriately arranged. Recommendations about the directional characteristics of aid microphones are based largely on measurements made on a KEMAR anthropometric acoustic manikin, while recommendations about the maximum tolerable internal noise of hearing aids are based on a specially performed experiment which determined the maximum tolerable signal-to-noise ratio for narrow- and wide-band random noise added to continuous discourse.

Gain and Frequency Response

Probably the most important specifications for a hearing aid are its gain and its frequency response. Although the two terms really relate to the same thing, i.e., the response curve showing gain at each frequency, it is convenient to separate the curve into two components. "Frequency response" is usually taken to mean the shape of the response curve, and "gain" is used to refer to the overall degree of amplification. Gain is usually quantified by the maximum gain, or by the average gain at specified frequencies, or the gain at 1 kHz or 2 kHz. Both gain and frequency response are presently selected at NAL on the basis of the procedure described by Byrne and Tonisson (1), which requires only threshold loss as its input data. It is thus possible, using this procedure, to predict what range of responses will be required if audiograms of the population to be fitted are available.

The data used to determine the gain and frequency response requirements consisted of pure tone audiograms obtained from the case reports, selected at random, of 229 children, 249 war veterans, and 219 old age pensioners fitted with hearing aids.

For each audiogram, the required open access 2-cc coupler (HA, configuration, ANSI S3.7-1973) gain at each octave frequency from 250 Hz to 8000 Hz was calculated by means of a slightly modified version of the August 1976 revision of the method described by Byrne and Tonisson in (1). In conductive and mixed-loss cases, one-quarter of the air-bone gap was added to the gain that would have been required had the loss been purely sensorineural.

In cases where the threshold exceeded the limit of the audiometer, the hearing threshold level (HTL) was assumed to be 5 dB greater than the audiometer limit.

Cumulative frequency distributions of the required gain at each frequency are given in Figure 1. The required gains vary from 0 dB (at 250 Hz and 500 Hz) up to 90 dB (at 1 kHz and 2 kHz). The cases requiring these very large gains had significant conductive components in the hearing
FIGURE 1
Cumulative frequency distributions of required open access 2-cc coupler gain for the entire sample at frequencies 250 Hz to 8 kHz.

FIGURE 2
Cumulative frequency distributions of required slopes at octaves 250 Hz/500 Hz, 500 Hz/1 kHz, 1 kHz/2 kHz, 2 kHz/4 kHz and 4 kHz/8 kHz for the NAL client population.
loss. Frequency response requirements were assessed mainly by examining the range of slopes required in each octave between adjacent audiometric frequencies (Fig. 2). The 500 Hz to 1 kHz octave obviously requires the greatest slope, on average. The 250 Hz to 500 Hz octave is noteworthy in that the cumulative curve does not span a large range; that is, 95 percent of the population can be fitted by a slope between -8 and +12 dB/octave with the median (50 percent) case requiring only +2 dB/octave. (A negative slope indicates a response whose gain decreases as frequency rises.) Also noteworthy is the large span required in each of the three octaves above 1 kHz.

The cumulative slope curves were broken down in various ways. Cumulative slope distributions are shown separately for each population type in Figures 3(a) to 3(e). The population needs are different only in the 2-kHz to 4-kHz and the 4-kHz to 8-kHz octaves. In the first of these, the curve for the child population is displaced to the left of the other two, indicating that, on average, children’s hearing losses do not deteriorate from 2 kHz up to 4 kHz as much as do adult losses. (This is not too surprising, given the noise-induced and/or presbycusic origins of most adult hearing losses, especially among the older population served by NAL.) Between 4 kHz and 8 kHz the pensioner population requires a greater positive slope than the other two, again conforming to expectations based on

FIGURE 3a, 3b
Cumulative frequency distributions of required slopes for the three sub-populations at (a) 250 Hz/500 Hz and (b) 500 Hz/1 kHz.

Please turn the page for Figures 3c, 3d, and 3e representing frequencies to 8 kHz as follows: (c) 1 kHz/2 kHz, (d) 2 kHz/4 kHz, and (e) 4 kHz/8 kHz: Pensioners (solid), Veterans (dashed), Children (dot-dashed).
FIGURE 3c, 3d, and 3e
Cumulative frequency distributions of required slopes for the three sub-populations at (c) 1 kHz/2 kHz, (d) 2 kHz to 4 kHz, and (e) 4 kHz to 8 kHz: Pensioners (solid), Veterans (dashed), and Children (dot-dashed).
the relative nature of presbycusis and noise-induced hearing losses.

The second breakdown of the curves investigated the different requirements for high- and low-gain aids. Responses requiring more than 52 dB maximum gain at any frequency were classified as high-gain aids: the basis for this classification is that this is the 50 percent point on the maximum gain cumulative distribution curve. It was found that low- and high-gain aids require a similar range of response slopes within each octave. However, distribution curves for the low-gain aids were slightly steeper in each octave, indicating that high-gain aids need to have somewhat more flexible response shaping.

The last breakdown examined the dependencies between the slope in one octave and slope in adjacent octaves for individual response curves. For example, if a particularly steeply sloping response is required in the 1 kHz to 2 kHz region, then is it possible to narrow the range of slopes required in adjacent octaves? Unfortunately, little or no correlation was found between the slopes required in adjacent octaves. This means that a universal hearing aid would need one tone control adjustment for every octave over which the response is to be fitted. Furthermore, if the slope within each octave is quantized into 6-dB steps, the total number of response options can be calculated to be 34,020. Of course, that could be greatly reduced by deciding to fit less than 100 percent of cases, or by planning to fit accurately only to 4 kHz instead of to 6 or 8 kHz—for example, fitting 95 percent of the population to 4 kHz requires “only” 1176 different response shapes. The authors believe that the 6-dB step size used above is a reasonable one in that a useful range of slopes (such as 24 dB/octave) can be achieved in a practical number of switch steps. We know of no research data that allows us to predict how far from optimum a response can be before the aid wearer is significantly adversely affected.

While the cumulative slope curves are extremely useful, they do not enable the shape of any individual response curves to be seen. For this reason, a clustering analysis was performed to produce a more easily handled number of representative response curves. So that differences in the shapes of the curves could be readily observed, the responses were normalized before clustering so that each had 0 dB gain at 500 Hz.

Results for the three subpopulations, each grouped into 10 clusters, are shown in Figure 4. One striking feature of these responses is the shape in the 250 Hz to 500 Hz octave. There is a clear requirement for most aids to have a flat or near-flat response in this region. The requirement is poorly met by most existing aids, which make use of

![Figure 4](image-url)
tone control circuits incorporating series capacitors. This causes a response in which the generally steeper slopes above 500 Hz are continued down indefinitely. It is possible that including a low-frequency shelf in the response of aids may resolve the present paradoxical situation whereby decreasing the low-frequency cut-off of an aid's response increases the quality of reproduction—but decreases intelligibility: see Punch, 1981 (2). That is, with a low-frequency shelf, it is likely that amplification can be extended down to very low frequencies at a level sufficient to be audible but insufficient to cause upward spread of masking. The requirement for near-equal gain at 250 Hz and 500 Hz is not peculiar to the Byrne and Tonisson selection procedure, it will result both from MCL-based procedures and from dynamic range bisection procedures, because neither hearing losses nor MCLs tend to vary much between 250 Hz and 500 Hz for the majority of clients. Furthermore, it doesn't matter whether or not the procedure includes a correction for the speech spectrum shape, because the average speech spectrum is also flat between 250 Hz and 500 Hz, as is pointed out by Byrne, 1977 (3).

Indirect evidence of the need for an extended low frequency response comes from speech discrimination tests conducted by Plant as reported in 1984 (4). In those studies, aided severely-hearing-impaired subjects showed poor discrimination between nasal sounds and voiced stops, and poor discrimination of vowel duration (i.e., long versus short vowels) whenever the first formant was lower than 400 Hz. Both of these types of errors are readily explained if the low frequency components of the speech signal are inaudible or sufficiently low in level to be masked by higher frequency components. The importance of low frequency components in sentence discrimination has been shown by Speaks (5). This also is understandable in terms of the suprasegmental information known to be carried by low frequency components.

Overall, we feel fairly confident that an extension of the low frequency response of hearing aids by use of a low frequency shelf will be beneficial to many hearing aid wearers. Even with present aids, low-gain aids often achieve a low frequency shelf at 0 dB gain when worn, because of transmission of low frequency sound in through the vent or around a loosely fitting mold. Unfortunately, this mechanism is of no use to those who require a low frequency gain of more than 0 dB.

In concluding this section, we would like to put a limitation on the data presented. The required responses and range of slopes are clearly only as accurate as the Byrne and Tonisson selection procedure. The procedure often needs to be modified when applied to profoundly impaired individuals. Many such individuals are fitted with an aid with a fairly flat response up to about 1 kHz and may not be able to make much use of amplification above that frequency. This type of response curve is not represented in the present data. Validation studies (for less than profoundly impaired individuals) on the procedure are currently being undertaken by Byrne. A complete analysis is not yet available, but a preliminary analysis of the data indicates that some change to the procedure is likely. While overall required gain will probably continue to be estimated to vary at about half the rate of the hearing loss, the required range of slopes does not appear to be as great as that predicted by the present procedure. In particular, the steep slope which the procedure invariably predicts in the 500 Hz to 1 kHz octave is likely to be less for many clients.

### Saturated Sound Pressure Level

In comparison with the selection of gain and frequency response, the selection of the optimum maximum power output (MPO) or saturated sound pressure level (SSPL) of a hearing aid for each individual client has received very little attention, either in the research literature or in hearing aid fitting practice. Yet it is quite clear that the SSPL curve of an aid is very important. One has only to subtract the gain of an aid from its SSPL to realize that it does not take a very intense sound in the environment to drive some aids into saturation. Under this condition it seems possible that the shape and level of the SSPL curve could be an even more important attribute of the aid than the gain curve.

If the SSPL of an aid is excessively high for a particular client, then one of several things can happen. First, the output level accompanying relatively intense input sounds may be sufficiently high to cause the aid wearer discomfort or even pain. Second, an excessively high SSPL curve may cause damage to the person's remaining hearing. There have been documented cases where this has occurred, despite the fact that the person experienced no discomfort from the aid: see Hawkins, 1982 (6).

If the SSPL has been set too low, then the usable dynamic range of hearing for that person has been unnecessarily reduced. This will result in the removal of some amplitude cues from the speech signal, with possible loss of information and a decrease in the variety of sound levels to which the aid wearer is exposed. Further, if the aid does not incorporate compression limiting, then distortion will occur every time the aid saturates (which will be more often than necessary). For an individual with a very restricted dynamic range, a low SSPL setting may even cause speech in particular frequency regions to be always inaudible, irrespective of the input level or the volume-control position.

If one temporarily excludes the hearing damage constraint, then how should SSPL be selected? In the absence of any contrary evidence, we will make the assumption that at each frequency, SSPL should not exceed loudness discomfort level (LDL) at that same frequency. It seems likely that for aids with "peaky" SSPL curves, the peak SSPL will be able to exceed the LDL at that frequency, since for reasonably broadband input signals, output levels equal to the peak SSPL will never be reached. For the purposes of this report, it will be assumed that loudness discomfort is the primary constraint on SSPL selection. LDL's were obtained from 71 sensorineurally impaired children and from 45 adults with mixed losses: see Dillon\[b\]MCL abbreviates "most comfortable loudness."
One may query whether aids in the bottom right hand corner of the gain-MPO plot need be fitted. That is, is a high-gain, low-SSPL aid ever required? We feel that the answer is an unequivocal yes! Examination of the data obtained from the hearing impaired children revealed that 33 percent of the measured LDL-threshold pairs indicated a dynamic range of less than 15 dB. These people with very restricted dynamic ranges, and usually with large losses, require an aid that will be in saturation much of the time. The alternatives are to provide them with an aid that guarantees that the signal at that frequency is often inaudible—or else is usually uncomfortable or even painful. While it may be argued that people with large sensorineural losses have reduced frequency and temporal resolution, and cannot extract all the information present in a given frequency region, it does not seem fair to assume they can extract no information. On the other hand, it can be safely assumed that either an inaudible signal, or an aid rejected due to loudness discomfort, will lead to a lack of aided benefit.

The goal of ensuring an audible but comfortable signal in all frequency regions can be further helped by examining the shape of individual LDL-versus-frequency contours. Similarities and differences in the shapes of the contours can best be appreciated if the LDLs are all normalized to one value at a particular frequency. Figure 6 shows such data for both the child and adult samples. For either set

**FIGURE 5**
Required SSPL (assumed equal to client LDL) versus required gain (calculated from the pure tone threshold using the Byrne & Tonisson selection formula) for combined child and adult data at 2 kHz.
of data, there is a 40 dB spread in the curves by 2 kHz. If SSPL is to be equated to LDL, then obviously the shape of the SSPL curve has to be capable of being shaped by the same amount. When one considers the large range of LDL shapes, and the impracticability of fitting them with a single SSPL shape, there appears to be a strong need for controllable-shape SSPL curves. This need will be strongest for medium- and high-power aids, since the people wearing these aids will normally have the most reduced dynamic ranges.

The need for some type of dynamic range reduction system for many clients is unquestioned, as some of the data in the previous section have shown. While peak clipping can achieve the desired reduction, it does so at the expense of increased distortion and a consequent loss in intelligibility. Compression systems, by virtue of gain reduction for higher level sounds, achieve the range reduction with little perceptually relevant distortion and little consequent loss in intelligibility. The lack of undesirable side effects is obtained only if the parameters of the compression system are appropriately chosen.

The most critical parameter is the recovery time, which on the evidence available, should be somewhere in the range of 60-to-120 ms and probably nearer the bottom than the top: see Walker and Dillon, 1982 (8). Release time is defined as the time taken for the output to settle to within ±2 dB of its final value after a 25-dB step decrease in the input level. The lower limit is set by the need to prevent the gain from varying excessively during each voicing period and thus generating distortion products. The upper

![FIGURE 6](image-url)

LDL versus frequency for (a) child data and (b) adult data. All LDL values have been normalized to 0 dB at 250 Hz to illustrate the range of curve shapes encountered.
limit is set by the need to prevent gain reductions, which occur during intense vowels, from persisting during following weak consonants, thus reducing their intelligibility.

Selection of the optimum attack time seems less critical. Values too small will cause unnecessary gain reduction during extremely brief noise impulses, which will then cause the signal immediately following to be less audible. Values too large will not provide sufficient protection against high-level sounds with sudden onsets, unless the aid incorporates an additional limiting device. Attack time, defined similarly to release time but for a 25-dB increase in input level, should probably be in the range of 2-to-5 ms, but the evidence is not available to positively exclude values anywhere between 0.5 ms and 20 ms.

Compression systems can be categorized into three useful types: (i) compression limiting, (ii) whole-range syllabic compression, and (iii) slow-acting automatic volume control. The first two have attack and recovery times short enough to allow the gain to vary significantly during a single syllable. The last two operate in the compression mode during moderate level or even low level sounds, while the first one operates in the compression mode only when the sound level is so intense that a gain reduction is necessary to prevent uncomfortably or dangerously loud sounds from occurring. There is no audiological evidence that any one of the three is superior to the others in terms of speech intelligibility or quality; there is also no logical reason why all three types shouldn't be combined in a single aid. However, for various practical reasons, we recommend that compression limiting be the system chosen. Its advantage over the slow-acting system is that it inherently provides protection against brief intense sounds, thus removing the need to provide additional circuitry to perform that essential function. Its advantage over the whole-range compressor is that a fitting procedure is available. That is, the compression limiter is used to select the aid SSPL which, in turn, is equated to the client's LDL.

It should not be assumed that the more complex full-range compression systems will eventually prove to be superior. Both types of compression remove some of the speech cues that are associated with the waveform envelope. While these cues are not an important source of information for normal hearers (compressed speech is highly intelligible), they do provide important cues for the more severely impaired, as was reported by Risberg & Lubker in 1978 (9). Whole-range compression has the advantage that there is a large input range over which the intensity variations are only reduced rather than removed, but has the disadvantage that there is almost no input level at which they are left untouched. Compression limiting has the advantage that low- and moderate-level inputs are left uncompressed, but the intensity variations are removed almost completely from higher level inputs. Compression limiting also has the advantage that, because there is no gain increase for low level signals, the amplitude modulation of background noise during pauses in the signal of interest will not be a problem. This effect becomes more noticeable as the signal-to-noise ratio deteriorates.

Viewed somewhat differently, whole-range compression systems, including slow-acting volume controls, may be beneficial by keeping all speech at a near-optimum loudness level so that the speech can be kept near the peak of the hearing impaired person's performance-intensity function. However, all environmental sounds above the (low) compression threshold are brought up, or down, to this level, possibly creating fatigue and a feeling of detachment from the true "noisiness" of the environment. Only the appropriate research will enable the superior system to be chosen.

Ideally, the I-O curve should be completely flat in the limiting region. The compression ratio should certainly be greater than about 5 so that outputs significantly above the chosen SSPL value are never obtained.

The properties of circuits employing compression amplifiers are critically dependent on the relative locations of the compressor amplifier, the point at which the voltage is sensed to provide the compressor control voltage, and of the rest of the amplifier subcircuits. The monitored signal can, in principle, be fed either backward or forward to the variable-gain amplifier. We will consider only feedback arrangements because the only compression amplifiers available for hearing aid use are configured to work in that manner. A completely general circuit arrangement thus appears as in Figure 7(a). However, subcircuits have the same effect whether they are placed in blocks $H_{1A}$ or $H_{1B}$. A simplified block diagram is thus shown in Figure 7(b). The block $H_1$ includes the microphone and all amplifiers, filters, controls, etc., before the feedback point, while the block $H_2$ includes all amplifiers, filters, controls, etc. (as well as the earphone, tubing, and earmold) after the feedback point. It is then straightforward to show that the gain of the aid in the linear region is equal to $H_1H_2$ and that the SSPL is equal to $kH_2/G$. The compression threshold referred to the input is simply equal to the SSPL divided by the gain in the linear region: the compre-
sion threshold referred to the input is thus not an independent design choice. This approach bypasses the argument about whether the compression threshold should be frequency dependent, and what the nature of that dependency should be.

From the above discussion, it is clear that any component in block $H_1$ affects the gain, any component in block $G$ affects the SSPL, and any component in block $H_2$ affects both gain and SSPL. From the fitter’s point of view, it is most convenient to have noninteracting controls, and therefore the tone controls and fitter-adjusted gain control should, preferably, be in block $H_1$. Similarly, control of SSPL shape and of overall level should be in block $H_2$.

The optimum location of the volume control has been argued in some depth elsewhere: see Dillon and Walker, 1983 (10). There it was argued that, for compression-limiting devices, output-controlled compression (from the point of view of volume control operation) is the necessary configuration. That is, the volume control should precede the feedback point and so must also be in block $H_1$.

**Directional Response Characteristics**

The problems caused to the hearing impaired by background noise have long been noted. Directional microphones appear to be the only available device completely worn by the aid user which can offer a partial alleviation of this problem. Hearing aids incorporating directional microphones have now been available for more than 10 years. Despite their objectively measured ability to improve the signal-to-noise ratio (SNR) when the signal and noise are spatially separated, the majority of aids sold still do not incorporate directional microphones. A review by Mueller published in 1981 (11) gives a good summary of behavioral tests comparing directional and nondirectional aids. The majority of studies have shown that, in noisy situations, directional hearing aids offer superior speech intelligibility to the wearer. The remainder of the studies have indicated no difference between the two aid types. The directional properties of hearing aids seem to disappear when the aids are measured electroacoustically in reverberant environments, because the sound energy is incident from all directions, irrespective of the location of the signal source. However, in reverberant situations, intelligibility is increased if the ratio of direct-to-reverberant sound can be increased. Provided the aid wearer is oriented toward the signal of interest, the directional microphone will always emphasize the early-arriving direct sound compared with the later-arriving reverberant sound. Directional hearing aids can thus be expected to offer superior intelligibility even in reverberant environments, although the superiority may be less marked in highly reverberant environments.

Obviously, not all directional aids exhibit the same degree of directionality. Equally obviously, their benefit is greater, the greater their directionality. One common but fairly poor measure of directionality is the ratio of the front sensitivity to back sensitivity. This is not a suitable measure because an angle of 180 degrees (from the front) may happen to coincide with a null in the polar response curve or it may fall on the peak of a minor lobe. Measures such as the unidirectional index, directivity index, or “front to total random” ratio are better because they are not sensitive to the exact location of the response nulls or minima.

Each of the methods of quantifying directivity involves obtaining a ratio by comparing sensitivity in some specified direction with the sensitivity averaged over another range of directions. In this paper, these ratios will be expressed in dB. The additions involved in taking the averages are performed on a power law basis. The variation of sensitivity with direction is obtained from a polar diagram which shows the sensitivity for every angle of the incoming wave, usually with the sound source specified as lying in the same horizontal plane as the microphone. The definition of several useful measures of sensitivity are as follows (with 0 degrees representing the direction in front of the observer, and 90 degrees the direction to the side of the observer on which the hearing aid is mounted):

**Directivity index:** sensitivity at 0 degrees compared with sensitivity averaged over all directions.

**Unidirectional index:** sensitivity averaged over the frontal semicircle compared with sensitivity averaged over the rear semicircle.

**Maximum to average:** sensitivity in the most sensitive direction compared with sensitivity averaged over all directions.

**Front to total random:** sensitivity averaged over front semicircle compared with sensitivities averaged over all directions.

**Front ¼ to rear ¼:** sensitivity averaged over 0 degrees to 90 degrees compared with sensitivity averaged over 90 degrees to 360 degrees.

These methods can be used to quantify the relative advantages of the three main contenders for “directional” hearing aids. The first contender is a nondirectional hearing aid placed on a head with the directionality imparted by head-baffle and head-shadow effects. Unfortunately, at 2 kHz, the sensitivity maximum occurs at about 70 degrees from the front, and the frontal sensitivity is about 5 dB down from this maximum. Hillman (12) pointed out in 1981 that this makes it difficult for the aid wearer to simultaneously achieve the best signal-to-noise ratio and view the lips of the person speaking in a conversation.

The second contender is a conventional directional hearing aid with parameters selected to give good on-the-head performance.

The third contender is some arrangement that makes use of the person’s own directional aid, the pinna. This is done in in-the-canal aids (and to a lesser extent in in-the-ear aids) by locating a nondirectional microphone near the ear canal entrance. However, it is also possible to take advantage of the pinna with behind-the-ear aids if the microphone is set in the earmold rather than in the aid case. For all of these methods, the directionality is frequency-dependent because, for a given microphone, it depends on the relationship between signal wavelength,
inlet port spacing, head size, and pinna size. Figure 8 shows the directivity index for each of the three options at the octave frequencies from 250 Hz to 4 kHz. Data for the unaided ear were obtained from Shaw, 1974 (13), and refer to pressure at the eardrum. Up to 6 kHz, eardrum pressure and ear-canal entrance pressure both display the same directional dependency. Data for the two “hearing aids” were obtained by mounting a Knowles EB 1864 microphone in a hearing aid case (with an effective port spacing of 20 mm) which was in turn mounted on KEMAR’s head.

Directionality for the non-aided ear and the non-directional aid are similar up to 2 kHz because both are equally affected by the head. By 4 kHz, the pinna dimensions are significant and the unaided ear receives a boost for frontally incident sounds and an attenuation for sounds from the rear, while the non-directional aid situated above the pinna receives no such benefit.

The directional aid is clearly superior to both the others up to 2 kHz. By 4 kHz its directionality is no better than that of the non-directional aid (due largely to the low-pass filter in the rear port used to create the internal time delay, and partly to the fact that the port spacing is an appreciable part of a wavelength).

Overall, these measurements show that up to at least 2 kHz the directional aid can provide a SNR about 4-to-5 dB better than either a non-directional aid or in-the-ear microphone placement. By 4 kHz, in-the-ear microphone placement is superior, although the results shown refer only to placement at the ear canal entrance. The benefit obviously becomes less as more and more of the concha is filled with earmold and/or aid.

The value of the 4-to-5 dB improvement in SNR should not be underestimated. Under noisy conditions, an improvement of this much can increase speech intelligibility scores by as much as 20 percent. It is difficult to achieve such an increase by any other means (e.g., optimization of frequency response or complex signal processing).

It is well known that localization in the vertical plane is made possible by the spectral shaping caused by reflections from, and coupling with, the convoluted folds of the pinna. Use of a directional microphone located above the pinna will destroy those cues, and the new set provided will almost certainly not be as strong. However, the cues for vertical localization are high-frequency cues, certainly above 3 kHz and probably above 6 kHz, and it is unfortunate that, it is in the high-frequency region that present aids have their poorest performance—and hearing losses tend to be greatest. The reduced frequency-resolution of hearing impaired people will probably make the use of high-frequency spectral cues more difficult as well. On balance, it seems that, at present, most hearing impaired people are probably better served by a directional microphone than by a non-directional microphone in the ear.

While some aids on the market allow the user to close the rear port to remove the directionality, this does not seem to be a necessary option from the point of view of achieving different types of directional responses, as the vast majority of clients simply leave the switch in the directional position*. However, it does seem a worthwhile fitter-operated option, as the aid’s response can then be measured in a standard test box or single-port microphone coupler while in its non-directional mode, and a standard correction applied to determine its directional performance. Additionally, those people the

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*Private communication: H. G. Mueller, Ph. D., Supervisor Audiology Section, Walter Reed Army Medical Center, Washington, D.C.
fitter believes would be better served by an extended flat low-frequency response than by a directional aid could be fitted with the rear port closed off. The optimum directional pattern (i.e., the shape of the polar diagram) depends on where the signal is and where the noise is—and obviously these vary from moment to moment. However, if one assumes that the noise is randomly incident from all directions, and the wanted signal only from directly in front, then the hypercardioid response is optimal. Alternatively, if the wanted signal can come from anywhere in the front hemisphere, then the supercardioid response is optimal. These terms, which are defined with reference to their identifying equations in Table I, represent particular responses from an infinite family of responses all having the general sensitivity equation $S = S_0(1 + R \cos \theta)$ where the value $R$ varies among the members of the family.

However, these response shapes are obtained only when the aid is measured in isolation in a free field; when the aid is worn they are modified substantially by the aid-wearer’s head. Madaffari in 1983 published polar responses measured on KEMAR for aids with various values of $R$(14).

<table>
<thead>
<tr>
<th>Response</th>
<th>Equation</th>
<th>$R$</th>
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<tbody>
<tr>
<td>Omnidirectional</td>
<td>$S_0$</td>
<td>0</td>
</tr>
<tr>
<td>Cardioid</td>
<td>$S_0(1 + \cos \theta)$</td>
<td>1</td>
</tr>
<tr>
<td>Super-cardioid</td>
<td>$S_0(1 + \sqrt{3} \cos \theta)$</td>
<td>1.73</td>
</tr>
<tr>
<td>Hyper-cardioid</td>
<td>$S_0(1 + 3 \cos \theta)$</td>
<td>3</td>
</tr>
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Various directivity figures of merit have been abstracted from those responses (Fig. 9). Some of the indices (such as maximum sensitivity relative to average sensitivity) are little affected by the value of $R$ chosen. Overall, however, there is a tendency for the indices to be maximized for responses near that of the cardioid. It is thus recommended that a value of $R$ of about 1.0 be adopted, but as the maximum is rather broad and occurs at different values of $R$ for different figures of merit, any value in the

![FIGURE 9](image)

**FIGURE 9**
Various directivity figures of merit plotted as a function of the variable $R$ in the sensitivity equation. The values of $R$ are those applicable for the aid in isolation, while the actual directivity measurements are for the aid mounted on KEMAR. A double set of values is present for $R = 0.76$ because the four smallest values of $R$ were obtained with a 6.3-mm port spacing and the four largest with a 14.4-mm port spacing. All points are averages of the results at 1, 1.6, and 2.5 kHz.
range 0.7 to 1.4 must be considered satisfactory. In practice various values of R are readily obtained because R is equal simply to the ratio of the external delay between microphone ports to the internal microphone delay in the rear port path (14).

**Maximum Internal Aid Noise**

The work reported in this section was performed in an attempt to determine the level at which the internally generated noise in hearing aids becomes a problem. Noise in hearing aids is most usefully expressed as “equivalent” noise needed to be applied at the input in order to produce the same output noise. The equivalent input noise is most fully specified if it is measured on a band-by-band basis. That is, the output noise is measured in separate (usually 1/3-octave) frequency bands, and from the measurements is subtracted the gain at the center frequency of each band. The result is a complete 1/3-octave spectrum of the aid noise referred to the input. This method obviously gives more complete information about the aid’s noise, but how can it be used to arrive at a subjectively determined criterion?

If we were to know the spectrum (and level) of the signal input to the aid, the equivalent input noise spectrum could be used to determine the SNR (at each frequency) received by the aid user. Fortunately, we are in a position to estimate the average input signal, since it will have the same shape as the long-term spectrum of speech; e.g., see Byrne, 1977 (3). (This estimate assumes that the primary use of hearing aids is for speech communication.)

Having now characterized the typical input signal, we need to determine a criterion SNR at which the aid will be deemed “just acceptable.” To find this, an experiment was conducted in which eight normal-hearing subjects were asked to adjust the level of a wide-band and various narrow-band noise signals. The wide-band noise was shaped to have the same long-term rms spectrum as the speech, and the narrow-band noise was filtered from this by using octave-wide filters at 250, 500, 1000, 2000, and 4000 Hz. A tracking procedure was used in which the background noise level was controlled by the subject via a recording attenuator. The subjects were instructed that they were to press the button whenever the noise became objectionable for longterm listening and to release it when the noise was no longer objectionable. All subjects were experienced at the tracking procedure, and all were aware that the purpose of the experiment was to establish a criterion for the maximum allowable internal aid noise. The subjects, with one exception, were tested twice for each of the six noise stimuli.

**Noise criterion test results**—From a comparison of the test and retest data (for the seven subjects tested twice), it was quite obvious that all except one were able to establish a noise criterion and stick to it. Test and retest for each stimulus were separated by about 15 minutes but the standard deviation of the test-minus-retest differences, averaged across six subjects and all six noise stimuli, was only 2.4 dB. This is even lower than that expected for threshold. The seventh subject had an average test-retest standard deviation of 11.5 dB, but as his overall results were very close to the average results of the rest of the group, his data were retained. The results for each subject, expressed as the just-acceptable noise relative to the signal, are shown in Figure 10. Despite the internal consistency of each subject, there are reasonable differences among the SNRs deemed acceptable for the various subjects. The results for one subject are obviously quite different from the rest; these were excluded from further consideration. The remaining subjects were all in agreement that a poorer SNR can be tolerated at the lowest and highest frequencies. A SNR of about 30 dB is needed to satisfy most of the subjects at the 3 mid-frequencies, with the value falling to 18 dB at 250 Hz and to 25 dB at 4 kHz. This is represented in Figure 10 by the
heavy line with its dotted extrapolations to lower and higher frequencies.

The final criterion input noise can thus be calculated by subtracting the SNR given in Figure 10 from the long-term average speech spectrum with an overall level of 65 dB SPL. The criterion equivalent input noise (in \( \frac{1}{2} \)-octave band levels) is shown in Figure 11. It is equivalent to an A-weighted noise level of 33.7 dB, but not all noise spectra with an A-weighted noise level of 33.7 dB will meet this criterion! (The A-weighted level of a sound is the overall level with the low and high frequency components attenuated to compensate for the reduced sensitivity of the normal human ear to low and high frequency sounds.) For interest, the \( \frac{1}{2} \)-octave equivalent noise levels of the normal human ear are shown in Figure 11 as the dotted line. This curve was calculated by Killion (1976) by considering the normal threshold in quiet to be a masked threshold, with the masking provided by the equivalent input noise of the ear (15).

The criterion outlined above is strictly valid only if the noise present is completely contained within a single octave band, because that is the experimental condition under which it was obtained. We should thus ask whether the criterion SNR in a given frequency region depends on the amount of noise present in other frequency regions. Examination of the data for each subject showed that the just-acceptable level of the 500 Hz and 1000 Hz bands was not greatly affected by whether they were presented in isolation or as part of the wideband noise. Averaged across subjects, a noise level only 1 dB higher could be tolerated when either of these bands was presented alone than when they were presented as part of a wideband noise. From this it follows that the criterion noise level (or SNR) previously suggested is not strongly influenced by the spectral shape of the noise under

**FIGURE 11**
Maximum equivalent noise (measured in \( \frac{1}{2} \)-octave levels) deemed to be acceptable when listening to a speech signal with a long-term rms level of 65 dB sound pressure level (curve), and equivalent input noise of the normal ear (circles).

**FIGURE 12**
Hypothetical examples of an objectionably noisy aid (A) with noise rating of +10 dB, and a quiet aid (B) with a noise rating of -3 dB.
The result suggests that the overall objectionability of an aid's internal noise can be assessed by noting the greatest amount by which the noise in any octave (or one-third octave) exceeds the criterion noise in the respective octave (or third of an octave). Similarly, a negative figure will indicate how far short of objectionable a particular equivalent input noise is. Figure 12 gives two examples: aid "A" would be given a noise objectionability rating of +10 dB and so would probably be considered unsuitable, while aid "B" would be given a rating of -3 dB and be considered sufficiently quiet.

The criterion can be relaxed somewhat in the low-frequency region whenever an effectively vented earmold is to be used. If the sound at the eardrum is dominated by vent-transmitted sound, then noise originating in the aid is clearly of little consequence. In fact, for every decibel by which the combined transmission gain (aid plus vent) lies above the aid transmission gain alone (as modified by the vent, but not including sound transmitted in through the vent), one decibel can be added to the criterion input signal level. As the mold characteristics are known only when individual clients are fitted, a practical approximation for the purposes of aid type testing would be to relax the criterion by 1 dB for every dB by which the aid gain falls below 0 dB transmission gain when measured in an artificial ear (or below -4 dB when measured in a 2 cc coupler). This relaxed criterion will be valid only provided the aid is fitted with a suitably vented earmold.

Existing electret hearing-aid microphones (except for those with sharply rising responses achieved by acoustical means, and directional microphones with very small port spacings) are just capable of meeting the noise criterion proposed here.

Conclusions

Selected aspects of hearing aid electroacoustic performance have been examined. An analysis of the range of slopes required in aid frequency response (based on the Byrne and Tonisson selection procedure) indicated that most hearing aid candidates require as much gain at 250 Hz as they do at 500 Hz. Very few existing aids meet this requirement. Inspection of the range of LDL curve shapes measured on hearing impaired people reveals that, if hearing aid SSPL curves are to be matched to LDL on an individual basis, then adjustable-shape SSPL curves are required in hearing aids. This can be achieved in a single-band aid employing compression limiting. The need for directional characteristics in hearing aid microphones seems well documented, and available data indicate that the response should be approximately cardoid when the aid is measured in isolation if it is to have optimal performance when mounted on the head. Last, a criterion for the maximum tolerable internal noise of hearing aids has been proposed based on experimental data obtained with normal hearing people.

REFERENCES