Guest editorial

Hearing aids: Past, present, future: Moving toward normal conversations in noise[1]

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We have heard puzzling things lately regarding hearing aids. One rumour suggests that a new type of signal processing will improve the recognition of speech in noise, even though research has failed to demonstrate a significant improvement for any new type of signal processing — compression, digital signal processing, or other — over clean linear amplification whose frequency response and gain have been properly adjusted for the test conditions (provided the test conditions are constrained to constant presentation levels and the subjects are not profoundly hearing impaired). Those same research results are sometimes interpreted as an argument for linear amplification by those who forget that in the real world individuals rarely have the controls necessary to provide the treble boost for low-level sounds required for best intelligibility, or a research audiologist ready at hand to set those controls.[2] In the USA, we have official government pronouncements from the FDA (Food and Drug Administration) that all hearing aids use basically the same type of circuits and that hearing aids cannot help in noise (Kessler, 1993).

The interesting common element in these comments is that they are all directed at the principal remaining unsolved problem reported by hearing aid wearers: that of hearing in noise. In this editorial, I argue the following points.

First, it is true that a properly adjusted linear hearing aid might provide roughly as good intelligibility in noise for most hearing aid wearers as sophisticated compression hearing aids. This is provided that the linear aid has the appropriate compromise frequency equalization, is regularly readjusted by its user each time the listening situation changes, and it has none of the traditional defects which reduce available speech information (narrow bandwidth, distortion, irregular frequency response, or frequency response inappropriate to the hearing loss).

Second, it is true, in particular, that there is little laboratory evidence that improved speech intelligibility will result from a hearing aid automatically adjusting its gain and frequency response in order to restore normal loudness relationships, attempting to reject stationary noise, or any other signal processing yet tested. (In practice, however, this statement really applies to laboratory research in which, for example, one-third-octave-band equalization is used to provide the 'linear' equalization appropriate to *each* test presentation level. As the required equalization often depends dramatically on level, as shown by Skinner (1976), the results of many such experiments should perhaps not be labelled 'linear' but instead 'experimenter-controlled-TILL' processing.)

Third, there are pleasant advantages for the user, nonetheless, in having the hearing aid automatically adjusting its gain and frequency response, not necessarily because it improves intelligibility in noise but because it eliminates the inconvenience of frequent volume control adjustments without *degrading* the intelligibility in noise.

Fourth, compared with the *unaided* condition, hearing aids can provide dramatic help in noise (despite Kessler's official pronouncements), i.e.

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[2]In the real world, there is often a substantial practical advantage to TILL (treble increases at low levels) processing over the old-fashioned only-one-frequency-response-you're-stuck-with-it linear processing.

dramatic improvement in the signal-to-noise ratio (SNR) required for conversational speech, whenever the level of speech in noise is low enough so that neither can be heard clearly without the hearing aid. Once cocktail-party levels (85 dBA) or party-with-orchestra levels (95 dBA) are reached, the best the hearing aid can do for *most* individuals is stay out of the way (Killion and Villchur, 1993). (Individuals with severe-to-profound loss may need amplification even under those conditions.)

With regard to the third point, one churchgoer wearing linear hearing aids said to me, 'Oh yes, I have to adjust my aids six or seven times each service'. This is hardly surprising, because quiet speakers at such a service may produce listening levels of 55 dB SPL, while lustily sung hymns with organ accompaniment can produce 100 dB SPL. Any linear aid set to provide good audibility for the 20 dB HL and quieter cues in 55 dB SPL (40 dB HL) speech will almost certainly overload without grace for 100 dB SPL inputs! I have personally measured a range of 55–95 dB in normal group conversations, between offhand remarks and forceful argument. At a recent Chicago Symphony Orchestra concert, the sound levels on Ravel's *Rhapsodie Espagnol* ranged between 45 dB and 95 dB at centre balcony seats. Martin (1973) dramatized the dilemma facing the wearer of linear hearing aids.

Providing they have low distortion, a smooth wideband frequency response, and appropriate variable recovery times, the only known disadvantage to aids that automatically adjust the gain and frequency response as needed is that they can improve the audibility of quiet sounds so much that they are sometimes considered noisy by new users. This problem tends to disappear with time as the wearers relearn how to localize and recognize those sounds so that the automatic brain-operated noise squelching circuits can operate again (Gatehouse and Killion, 1993). The writer as amateur violinist is gratified to report that world-class violinists in the Chicago Symphony Orchestra and elsewhere are now successfully using such hearing aids in concert as well as in daily life.

Consistent with the above arguments, Villchur (peronal communication) reported on the results of 18 tournament-procedure adjustments of high-quality two-channel wide-dynamic-range-compression hearing aids, where part of the tournament involved choice of compression ratios (including 1:1 i.e. linear operation). All of the subjects had a clear preference for increased low-level gain (compression); none of the subjects chose a reduction in low-level gain to linear operation, which they were free to do.

Hearing aids past

Figure 1 shows what I believe to be true: Some time ago the distortion, narrow bandwidth, irregular response, and inappropriately adjusted (or available) frequency response of hearing aids garbled or muffled so much speech information that hearing in noise was almost impossible. Tillman, Carhart and Olsen (1970) determined the signal-to-noise (babble) ratio required for 50% word recognition scores of normal and hearing-impaired subjects listening through the best hearing aid they could find in the 1960s. That hearing aid worsened the SNR required by their hearing-impaired subjects by 18 dB. Their normal-hearing subjects required a 12 dB greater SNR with the same aid, indicating that it was indeed the hearing aid's limitations that were reducing the listeners' ability to understand speech in noise. During that time, everyone — with either normal or impaired hearing — could hear better at a loud party *unaided*.[3]

Over the years, improved hearing aid design and hearing aid fitting (surely as important as the design in many cases) have removed the previous
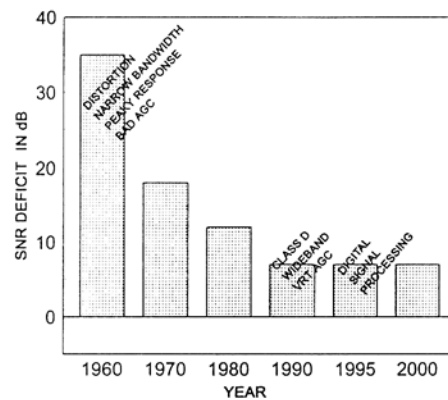

Fig. 1. Hearing in noise with hearing aids: better circuits helped a lot, but there is no 'better' left to do.

[3]The noise level at a typical cocktail party is 80–85 dBA, at which level most hearing-impaired individuals could receive enough information unaided to make that preferable to the garbled information available through a hearing aid.

defects from hearing aids. By 1990, Class D amplifiers, wide bandwidth (16 kHz bandwidths, in fact), smooth real-ear response, appropriate variable-recovery-time compression, and improved fitting targets and real-ear measurement techniques had brought the SNR down to what appears to be an irreducible floor. (Variable-recovery-time compression had been introduced before 1990. In 1993, a Fikret-Pasa study confirmed that variable-recovery-time compression limiting could remove degradations in SNR brought on by traditional fast-recovery AGC systems.) By 1990, we had reached the point where hearing-impaired individuals routinely heard better in noise *with* the better hearing aids available; sometimes 5–10 dB better in low-level noise and no worse even in high-level noise (Killion and Villchur, 1993).

The introduction of the newer digital signal processing hearing aids did not provide a further reduction in SNR for understanding speech, although their increased fitting flexibility may make it easier to *reach* that minimum in some cases.

More on noise

It may be worthwhile to say a few more words about the futility of trying to somehow assist the brain in processing signals in noise. Until recently, the world's most powerful computer was the Thinking Machine, with 64 000 cpu chips working simultaneously to solve problems in an attempt to simulate the parallel processing of the brain. The Thinking Machine is so powerful that it can do human-like things, such as recognize a face — in half an hour! A baby can do it in half a second. If a 64 000-computer-chip computer, which takes so much power that it requires a special electrical service, can only achieve 1/3600 the processing power of a baby's brain, we cannot soon expect a single-chip computer operating on flea power to be able to know when you wish to listen to what John is saying but want to ignore what Graham, Anu, Phillipe, Tony, and Celeste are saying, and then when a few minutes later you wish to switch to listening to Celeste.

What both digital and analog signal processors *can* do is make certain the brain is not starved of information by the operation of the hearing aid. In the old days, hearing aids often distorted and restricted the available information so badly that they made things *worse*. What Villchur (1993) argued and demonstrated was that clarifying the

speech was the first order of business, asking rhetorically, 'Would you prefer clear speech in clear noise or muffled speech in muffled noise?'

Once we stop *creating* a problem with our hearing aid designs, and provide a suitable 'selective amplification' type of frequency response, we find a ceiling effect. We can worsen the situation by distorting, restricting, or otherwise abusing the incoming speech and noise, or by inappropriately compensating for a frequency-dependent hearing loss, but we can't improve things beyond *not* doing those things. (I speak here of laboratory conditions, where the gain and frequency response are adjusted to maximize intelligibility, not of real-world conditions where the automatic adjustment of gain and frequency response may prove of great practical value.)

Indeed, Bentler (1997) recently reported preliminary results indicating that, for high-level (83 dB SPL) presentations of the SIN test, an 1800's speaking tube provided an SNR for 50% correct that was about equal to that of two recent all-digital hearing aid designs (slightly better than one, slightly worse than the other). The speaking tube had a moderately wide bandwidth, reasonably smooth treble-emphasis frequency response (partially damped by the resistive part of the eardrum impedance), and 0.0% distortion. In effect, we have returned full circle in our recent designs!

Some 20 years of digital noise-reduction research, on the other hand, has shown that it is indeed possible to reduce the noise, but the speech is sufficiently degraded in the process that there is no net improvement in signal-to-noise ratio required for adequate intelligibility (Hochberg *et al.*, 1992; Ludvigsen *et al.*, 1993). We achieve a hollow victory over noise.

Is there anything useful left for us to do? Indeed there is. The problem of hearing in noise remains, and is roughly proportional to the degree of hearing loss.

SNR versus hearing loss

Figure 2 shows the SNR required as a function of hearing loss by a variety of subjects. Two points are evident: a greater SNR is required on the average as greater hearing losses are encountered, and a large individual variability is seen. Some subjects with 40–50 dB loss require large signal-to-noise ratios for a 50% correct score on words in sentences. On the other hand, some subjects with 50–60 dB loss can perform at almost

normal levels. The presumed explanation for these differences is that the former subjects have extensive inner hair cell damage, while the latter have a hearing loss characterized by nothing but outer hair cell loss. In some cases, this may be due to different types of noise-induced hearing loss. Borg *et al.* (1996) suggest that high-intensity noise causes extensive damage to inner hair cells as well as outer hair cells; long-term lower-level noise causing the same audiometric loss may show predominantly outer-cell loss.

By paralysing the motion of outer hair cells with furosemide, Ruggero and colleagues (Ruggero and Rich, 1991) have shown a loss of 40 dB or more in basilar membrane velocity for low-SPL inputs, with no change in velocity for high-SPL inputs. This suggests an explanation for the finding that a loss of sensitivity for quiet sounds may occur while high-level sounds are heard normally: the sensitivity loss was due to outer hair cell damage only. A variety of psychoacoustic and physiological experiments have indicated normal cochlear function for high-level sounds in the presence of mild-to-moderate threshold loss, as summarized by Killion (1979).

The *inner* hair cells, on the other hand, provide the neural inputs to the brainstem. Loss of inner hair cells means loss of information flow to the brain, starving the brain of the full information flow it needs to separate speech from noise.

The important thing about Fig. 2 is that a steady progression of hearing loss as measured by the audiogram is accompanied, on the average, by a steady progression of hearing loss as measured by the signal-to-noise ratio required to carry on conversational speech.[3]

Figure 3 shows the *average* deficit in S/N as a function of hearing loss, obtained from the smoothed curve in Figure 2. Those with mild-to-moderate loss required 4–6 dB greater SNR
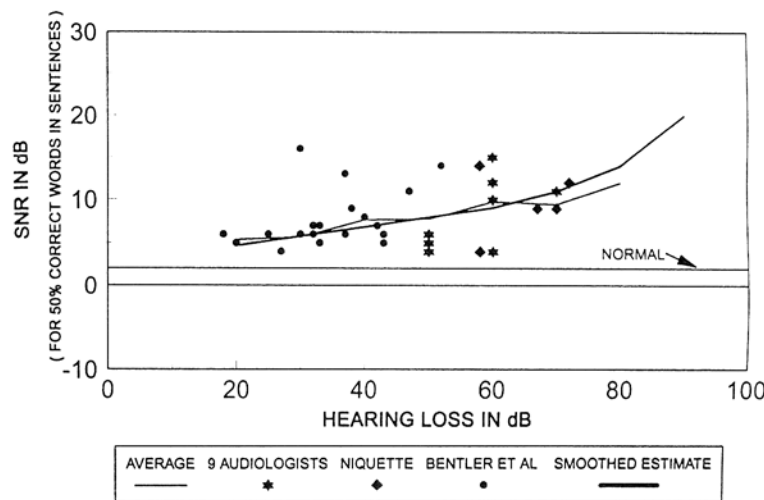

Fig. 2. Signal-to-noise required for 50% word recognition scores as a function of hearing loss. SIN test blocks 3, 4, 5 and occasionally 6 and 8 were used to obtain these data.

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[3]The data in Fig. 2 were taken at 83 dB SPL (70 dB HL) for subjects with mild-to-moderate loss, and at the 'Loud but OK' level (just below 'Uncomfortably Loud') for subjects with greater loss. At these high levels, the data of Skinner (1976), Dirks (1982) and van Buuren *et al.* (1995) suggest that little improvement would be obtained by shaping the frequency response of the sound-delivery system, which was an audiometer. The SIN (speech-in-noise) test that was used consisted of a 4-talker babble (three females and one male) and a female talker delivering the IEEE sentences at signal-to-noise ratios of 0, 5, 10, and 15 dB, with five key words scored per sentence. (The CD version of the SIN test is available from Auditec of St Louis, St Louis, MO, USA.) By plotting the per cent correct for the words in each sentences versus SNR, the SNR for 50% correct can be estimated. Those numbers are plotted in Fig. 2. Normals typically require a 1–2 dB SNR for 50% correct on this test. In the 0 dB SNR blocks, it is possible, more or less, to follow any one of the five talkers, which is consistent with our experience at parties, where we can voluntarily move our attention from the talker in front of us to a more interesting talker nearby, making use of what Broadbent (1958) called 'selective listening'.
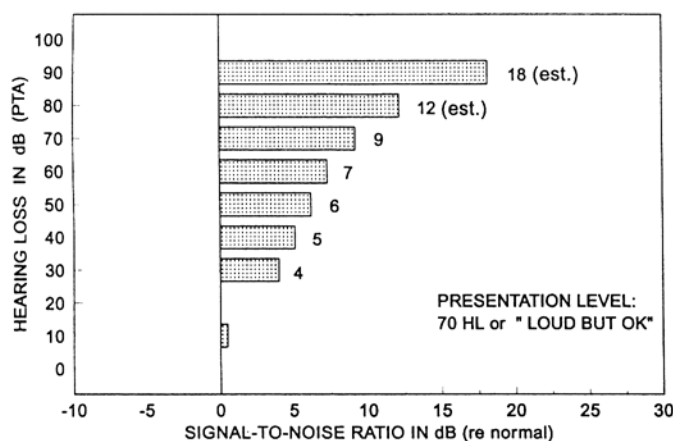
Fig. 3. Smoothed-average SNR versus HL data from Fig. 2.

than normal. Those with moderate-to-severe loss required 7–9 dB. Those with severe-to-profound loss required 12–18 dB greater SNR. Figure 3 illustrates the average problem remaining *after* the hearing aid circuit has been made defect free and adjusted for maximum intelligibility in noise.

Some solutions

Fortunately, there are hearing aids available and under development that can solve the problem of hearing in noise, even for those with profound hearing loss. In the near future, the main limitation to dramatically increasing the SNR will be not expense but inconvenience. Rather than attacking the problem indirectly with sophisticated digital filtering algorithms, which can't possibly — at present — know which of several talkers at a party should be considered noise and which should be considered the desired signal to be enhanced, these new developments attack the problem directly with what might be called a head-through-the-wall approach. There is nothing intrinsically new about these solutions, which can improve the signal-to-noise ratio by 5, 10, or even 20 dB. What is new is their practicality with today's electronic technology.

Figure 4 shows the in situ polar plot of an improved first-order directional microphone designed for ITE (in the ear) hearing aids. The two plots were measured at 2 and 4 kHz, both on KEMAR. They provide a calculated directivity index of approximately 6 dB at those frequencies.

The directivity index (DI) is obtained by comparing the sensitivity of a microphone for 0° (frontal) sound to its sensitivity to random-incidence sound (such as found in a well-designed reverberation chamber). This is the figure of merit that has been used for directional microphones since the 1940s, and provides an estimate of the improvement to be expected in the most difficult cases where multiple echoes bring the noise from the rear around into the front and sides of the microphone.

As the DI often changes with frequency, we have found it useful to use the articulation-index importance functions to weight the DI at each frequency. This produces an average articulation-index-weighted directivity index, or AI-DI number, which will roughly correspond to the improvement in SNR that can be measured in speech tests conducted in real-world situations such as cocktail parties and restaurants (Soede, 1990). The AI-DI figure for the microphone shown in Fig. 4 is approximately 5 dB.

Hawkins and Yacullo (1984) measured an improvement in SNR of 3–4 dB in a normally reverberent environment using a BTE directional microphone with good directivity at 2 kHz and below but with little reported directivity at 4 kHz. Their real-world results are thus consistent with the 5 dB number given above for an improved directional microphone design.

In some cases, the improvement in SNR will be greater than indicated by the calculated AI-DI. For example, if the sources of noise are close at
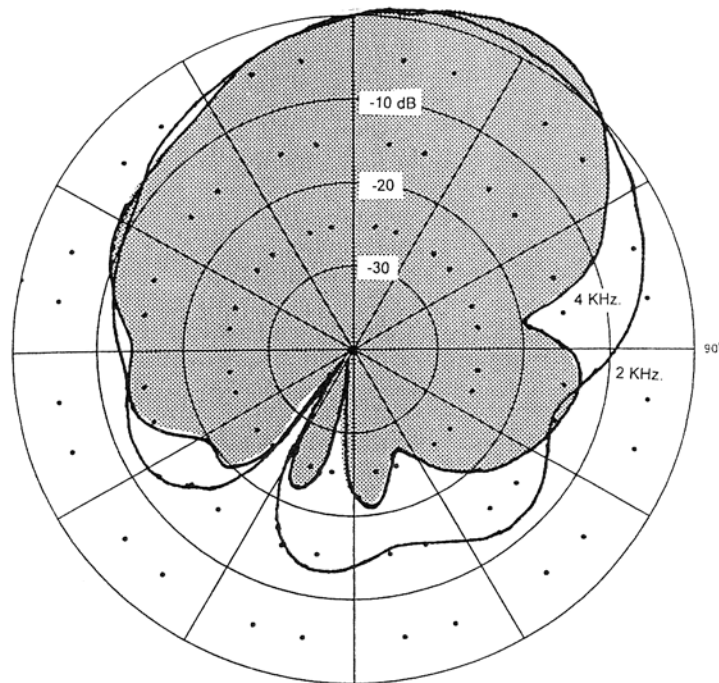
Fig. 4. Polar plot obtained in an anechoic chamber on an ITE directional microphone placed on the KEMAR manikin. The directivity index for these plots is approximately 6 dB. Note that a directivity index of 7 or 8 dB would be obtained if sounds at 30° to the side were considered the desired sounds, which may sometimes be the case for good binaural hearing.

hand and located at relative minimums in the directional microphone's sensitivity, and the nearest reflecting surfaces are relatively far away, improvements of 10–15 dB and greater are possible. As shown in Fig. 4, for example, a 2 or 4 kHz interference located at an angle of 215° would be attenuated more than 30 dB. It is sometimes possible to show improvements of up to 20 dB in the clinic, using what Harvard's FV Hunt (1954) called 'carefully contrived listening tests', but figures derived from semi-anechoic test booths should be cited only with strict caveats. Nonetheless, it seems reasonable to assume that a well-designed directional microphone hearing aid will provide adequate improvement for the typical (although not the exceptional) individual with mild-to-moderate hearing loss; adequate, that is, to bring performance in noise to near-normal levels or better. In any case, all studies have shown an improvement with, and usually a preference for, directional-microphone hearing aids (Mueller *et al.*, 1983).

Soede (1990) demonstrated that when greater SNR improvement is required, subjects with moderate hearing loss could perform just as well as normals in noise when they used array microphones to help reject the noise. Small pencil-size array microphones with suitable wireless signal transmission are under development. These can be worn over the ear, in the hair, be hand held, or set down on the table. The AI-DI of a well-designed array microphone can be 8–10 dB, adequate for most subjects with moderate hearing loss.

Finally, we are all familiar with the 'FM trainer' used in schools, were the teacher wears the microphone/transmitter and the students wear FM receivers. An improved version of this will permit multiple microphones to be given to friends at a restaurant, for example, each of whom transmits a signal to the receiver of the hearing-impaired listener. While theoretically possible for some time, the availability of sub-miniature transmitters and receivers has brought this product near reality. At a restaurant table, for example, the distance from talker to listener is

typically about 100 cm. By pinning a tiny microphone/transmitter on the collar of each talker, the distance drops to about 10 cm, thereby providing a 15–20 dB improvement in SNR. If three microphones are open (four persons in a restaurant, one hearing impaired) the SNR will degrade by approximately 5 dB in the summed signal. If directional microphones are used, however, their noise-rejection properties will improve the situation by about 5 dB. Thus, a practical improvement of 15–20 dB can be obtained, sufficient to allow most subjects with severe-to-profound loss to carry on comfortably in high-noise situations.

Again we are returning full circle. The 1800's speaking tube mentioned above will also provide a SNR improvement of 15–20 dB when used as intended, with the talker's mouth (producing typically 110–115 dB SPL) placed into the funnel.

The future

Figure 5 shows one view of the future in which the three solutions described above are applied to progressively more severe hearing loss.

Given the availability of the SNR solutions shown in Fig. 5, or those discussed recently by Agnew (1996), in the future we should be free to concentrate our circuit cleverness on making the experience of wearing hearing aids more pleasant, and abandon continuing efforts to improve intelligibility by trying somehow to improve on the brain's ability to separate signal and noise.

Changing the past

One of the severe limitations to hearing aid utilization is the large number of hearing aids out there that don't work (at least don't work well in noise). Thus, anyone contemplating obtaining hearing aids probably already believes they won't work: he or she has a relative or friend who can't hear in noise with hearing aids! MarkeTrak studies report only a 45% user satisfaction rating with those older hearing aids (Kochkin, 1995). Villchur has argued that therein lies the true source of the stigma associated with hearing aids, not any cosmetic considerations. If you see someone wearing glasses, for example, you don't offer them a large-print edition of the paper; you assume they can see just fine. When we saw someone with a hearing aid in the past, however, we knew from experience that he or she couldn't hear very well, especially in noise.

If most of the cars on the road were badly rusted and had irredeemably faulty brakes (or irredeemably unreliable electrical systems, like my beloved 1974 MGB), it is unlikely that as many people would be buying new cars each year. By that analogy, as the proportion of hearing-aid wearers who do well in noise increases, we can expect hearing aids to move toward the status of glasses: a nuisance, but a welcome relief from not being able to see well. More recent MarkeTrak studies showed a 91% consumer satisfaction index with a hearing aid having a user-switchable directional-microphone (Kochkin, 1996).
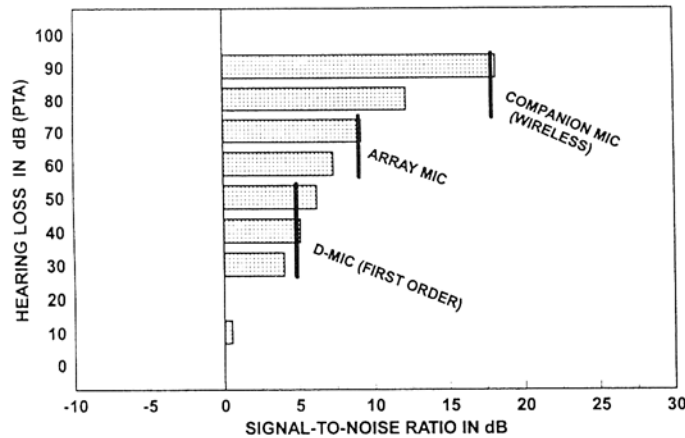

Fig. 5. Bringing people back to normal ability to hear in noisy places. Three technologies make it possible for individuals with mild to profound hearing losses to understand speech in normally noisy social situations.

Hearing aids of the future will surely solve the problem of hearing in noise. If Villchur is right, the stigma will gradually fade.

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