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Technical improvements to enhance the performance of completely-in-the-canal (CIC) hearing aids

A ***Hearing Review*** survey showed that since its introduction in 1993, completely-in-the-canal (CIC) hearing aids accounted for about 16% of the total hearing aid sales for 1995 and 1996 (Skafte, 1997). Their aesthetic and acoustic advantages (see Staab, 1996; Mueller & Ebinger, 1996 for a review) may have been responsible for the immediate acceptance of this style of hearing aid. There were speculations as to whether this style of hearing aid would, like in-the-ear (ITE) hearing aids, become the major style of hearing aid dispensed in the USA (Kirkwood, 1996).

Unfortunately, in the same ***Hearing Review*** survey, it was shown that CIC growth actually dropped from 68% a year ago to only 42% (Skaft, 1997). This drop in CIC growth rate should send an important warning to the hearing aid industry. If CIC hearing aids are to achieve the same acceptance as ITE hearing aids; if CIC hearing aids are to be successful in expanding the hearing aid market by tapping in more new hearing aid wearers as envisioned by Kochkin (1994), cosmetic and acoustic advantages are not sufficient. A performance solution is also critically important (Kirkwood, 1996). In-the-ear style hearing aids replaced the sale of BTEs because they achieved design sophistication and performance comparable to BTEs, and not simply because of their cosmetic appeal alone. Hearing impaired consumers may demand a discreet fit from their clinicians, but they also require performance from their hearing aids in order to keep using them. In this paper, we will explore which areas of technical improvements may further enhance the performance of CIC hearing aids.

Precision in hearing aid specification

Most clinicians select the hearing aid settings for potential wearers by taking their audiometric thresholds and applying them to a prescriptive formula. Correction factors are then added to account for transformation between sound field to real-ear, differences in transducers, couplers, venting etc. For example, a correction factor like CORFIG (Killion and Mosner, 1980) adds to the coupler response to achieve the desired insertion gain; RECD (Seewald et al., 1996a) adds to the coupler output to achieve the desired real-ear output. These transformations and corrections assume average ear characteristics.

It was indicated in Widexpress 11 (Ludvigsen and Topholm, 1997) that the conventional approach of determining audiometric thresholds with transducers may vary in real-ear SPL by as much as 20 dB across listeners. Because hearing aid gain is specified from threshold data, such variability means that one may prescribe inappropriate real-ear gain or compression settings for the "non-average" hearing aid wearer. At the same time, the use of average correction factors to select the desired coupler characteristics undermines the individual variability in ear canal size, eardrum impedance, leakage and venting effects and may lead to different output than is desirable. For example, Fortune (1997) showed that vent size could change the effective compression ratio on a compression hearing aid.

The problem of real-ear variability exists in all styles of hearing aids. However, the consequences are especially intolerable in a CIC for several reasons. First, conventional CICs, because of size constraint, are more limited in their electroacoustic flexibility than conventional BTEs and ITEs. Any miscalculation on the part of the clinician (or the manufacturer) can not be easily corrected by adjusting the appropriate potentiometer on the hearing aid. Instead, either the CIC has to be returned for modification, or the wearer has to tolerate the inappropriate settings. Secondly, most conventional CICs do not have a wearer adjustable volume control (VC). If the wearers find the loudness on a conventional BTE/ITE to be inappropriate, they can adjust the VC up or down. Such cannot be done, or can only be done inconveniently, in most CIC hearing aids. Thirdly, real-ear measurements can be made easily in a BTE/ITE hearing aid to evaluate its fit. It is more difficult to perform such measurements in a CIC accurately because of potential compression of the probe tube, and/or leakage of low frequency amplification (Staab, 1996). Although functional gain measurements have been routinely used, the interpretation of such results may be different from real-ear gain measurements because of room noise, circuit differences etc. (Seewald et al., 1996b).

Programmable CICs allow the clinicians external control of parameters on the CIC. This permits more electroacoustic changes than with a conventional CIC. This is invaluable especially if the initial settings on the CIC are inappropriately specified due to errors in threshold measurement and deviation between average and individual correction factors. Furthermore, the use of a remote control may compensate for the lack of a VC on the hearing aid.

A more direct method to minimize individual variability and real-ear errors in threshold measurements is to find the proper hearing aid settings based on in-situ thresholds established with the actual hearing aid in the wearer's ear. The hearing aid generates the test signal, and any modification of the magnitude of this test signal by the wearer's ear canal volume, eardrum impedance, venting etc. will be reflected in the magnitude of the threshold. Consequently, no conversion between earphone data and real-ear data, or application of individual or group correction factors will be necessary. Eliminating these sources of variability could mean a more accurate specification of hearing aid settings for the wearer and lead to higher satisfaction. Ideally, the hearing aid system must also have multiple channels in order to achieve the flexibility and precision in target response specification.

Better signal processing strategy

It was mentioned earlier that because of space limitation, CIC hearing aids typically do not allow for easy or convenient volume adjustment. The choice of a signal processing algorithm must consider this limitation as well.

Each of the two major forms of signal processing methods used today has its desirable features and limitations. Linear processing provides equal amplification to inputs at all levels. This preserves the intensity and temporal relationship between sounds. However, the wearer may need to adjust the volume in different acoustic environments in order to ensure audibility and comfort.

A form of non-linear processing, fast-acting wide dynamic range compression (WDRC) reduces amplification as the input sound pressure level increases. Compared to linear processing, this form of amplification may provide better audibility for low input signals and more comfort for high input signals without volume adjustment. A potential limitation of this type of processing is the reduction of intensity contrast and distortion of the temporal waveform of the input signal (Van Tasell, 1993). For hearing impaired people with poor psychophysical resolution, especially those with poor speech understanding in noise, this form of processing may be less desirable than linear pro-cessing. Interested readers are referred to Kuk (1996a) for a review of the different types of pro-cessing.

Although both forms of signal processing methods are used in CIC hearing aids, the choice of WDRC as the signal processing method seems reasonable because it allows better audibility and more comfort (than linear) without the need to adjust the VC (Bentler, 1994). However, it also means that performance may be compromised for some hearing aid wearers, especially in noisy situations.

One approach for achieving more audibility and comfort (than linear) while minimizing the drawbacks of fast-acting wide dynamic range compression is to use slow-acting compression with a low compression threshold. The low compression threshold ensures better audibility of low input sounds (than a higher CT), and the slow-acting compression puts the hearing aid in a linear processing mode, and at a gain level appropriate for the acoustic environment. Consequently, even without a VC CIC wearers may still enjoy audibility and comfort without intensity distortion of the input signal.

Quieter hearing aid

Johnson and Danhauer (1997) reported that people with more severe hearing losses are less sensitive about the look of a hearing aid. A report by Kochkin (1994a) shows that only 1 out of 20 mildly hearing impaired consumers, but 1 out of 4 moderately hearing impaired persons wore a hearing aid. Stigma was the most frequently cited reason for non-purchase (HIA, 1990). Kochkin (1994a,b) rationalized that because of its cosmetic appeal, CIC may be the tool to attract this group of hearing impaired non-wearers to using a hearing aid. Indeed, this may be the tool to expand the hearing aid market.

Kochkin (1994a,b) listed many potential obstacles that the industry needs to work on in order to realize this goal. There is at least one more obstacle - circuit noise. Hearing aids, like all other electronic devices, produce circuit noise. Because the microphone transduces acoustic signals for amplification, any noise that it produces will also be amplified to an extent determined by the gain setting on the hearing aid. In a CIC, this output level is further accentuated by the reduced residual volume of the ear canal. If this output is below the hearing aid wearer's hearing threshold, it will not be perceived. Otherwise, the wearer may complain of "rushing noise", "static" etc. in quiet environments.

People with a mild degree of hearing loss are more likely to have regions of normal hearing. Consequently, the circuit noise may become audible at all times which may cause them to reject the hearing aid. A "quieter" microphone, a larger residual volume (i.e., shorter CIC), and/or some mechanism to ensure that the amplified microphone noise output is below the wearer's threshold will be necessary in order to minimize the level of circuit noise.

Performance in noise

Most hearing aid wearers need to communicate in noise. Kuk (1996b) showed that as many as 34% of a group of hearing aid wearers (median age = 70 years, N=100) needed to communicate in noisy situations on a regular basis. Ninety-eight percent of the same group needed to communicate in noise at least some of the time. If the typical CIC wearer is younger and more socially active (Kochkin, 1994b), one may conclude that hearing in noise is one requirement of most, if not all CIC wearers.

CIC hearing aids offer many acoustic features that may enhance speech understanding in noise. Because of the recessed microphone, wind noise is reduced by an average of 7 - 10 dB compared to ITE hearing aids (Fortune and Preves, 1994). In addition, 6 to 8 dB concha resonance at and above 4000 Hz is expected from the recessed microphone position (Chasin, 1994). The reduced residual volume also leads to a higher output, with more in the high frequency (10 - 15 dB) than in the low frequency (4 - 8 dB) (Agnew, 1994; Chasin, 1994; Staab, 1996). The emphasis on the higher frequency components of the input signals (both speech and noise) may lead to better audibility of speech sounds in quiet. When one considers this in conjunction with the shadow effect cast by the pinna to high frequency sounds that originate from behind the wearer (Agnew, 1994), one can appreciate why more CIC wearers reported satisfactory use of their hearing aids in noise (38%) than those wearing conventional ITEs (30%, Kochkin, 1996). Apparently, these natural acoustic benefits are not sufficient to sustain the growth and acceptance of CIC hearing aids. Indeed, if one compares this satisfaction rate (38%) to that of single channel multimemory hearing aids with an omnidirectional microphone (46%) or multiple microphones (67%, Kochkin, 1996), one must conclude that additional noise management methods should be included in CIC hearing aids to improve their performance in noise.

The use of multiple microphones to achieve a directional advantage, and the use of directional microphones are not possible in a CIC today because of space and distance (between microphone openings) requirements for such systems. On the other hand, there are many relatively effective digital signal processing (DSP) algorithms which require only a single omnidirectional microphone to achieve noise reduction. Ludvigsen (1997) reported on a DSP speech enhancement algorithm that utilizes the difference in dynamic properties between speech and noise signals to achieve differential amplification for different types and various levels of signals. Such an approach may further enhance the performance of CIC hearing aids in noise beyond that provided by the natural acoustic benefits of CIC hearing aids.

Hall and Sandlin (1997) asked 20 wearers of high-end programmable hearing aids to compare their hearing aids with SENSO BTEs and ITEs in listening environments with different background noise levels. Eighteen subjects rated the SENSO to be "good", "very good", or "excellent" in all the noisy situations; whereas the number of subjects who gave their programmable hearing aids similar high ratings decreased as the listening situations became more difficult.

Feedback

Because of its tighter fit and lower gain, feedback may not be a problem for a CIC when it is fit to wearers with a mild-to-moderate degree of hearing loss. However in cases where the CIC is used for a more severe loss, or as a transcranial CROS (Bauman and Braemer, 1996), or when mandibular movement threatens the fit of the hearing aid, feedback could occur. It is worthwhile to manage the feedback problem so that this style of hearing aid can be extended to people with various degrees of hearing loss.

Several approaches have been advocated to minimize feedback in hearing aids. Use of a notched filter, reducing the high frequency output, shifting the frequency and phase, and limiting gain in specific frequency regions have been suggested. Interested readers are referred to the summary by Agnew (1996) on this topic. Some of these methods should be incorporated in CIC hearing aids.

Practical realization: DSP for CIC

This wish list is difficult to accomplish in its entirety even in an ITE style hearing aid using analog technology. Such circuit complexity would result in a fairly good size chip, and drain the battery at a high rate. Size, current drain, and circuit noise have always played an important role in hearing aid design. Fortunately, with the use of today's Very Large Scale Integrated (VLSI) chip technology, and with careful planning and design, digital technology can be used to achieve all the stated functions in a chip that is suitable for CIC use. Westermann and Sandlin (1997) outlined how DSP can achieve these goals without encountering the same degree of size, drain, and noise constraints as with analog technology. They have also summarized the algorithms used in the SENSO family of hearing aids. With the exception of a T-coil and its associated A/D converter, the SENSO CIC preserves all the features of the SENSO chip, which measures 9 mm^2 and drains current at a rate of 0.6 mA.

The hearing aid industry is entering an exciting age with digital technology. Many of the limitations that once prevented the full realization of a high performance hearing aid in a small size are becoming less difficult to overcome. These improvements, when implemented in a CIC hearing aid, not only allow the industry to meet the cosmetic demands of the hearing impaired consumers, but also their performance demands. With digital technology, cosmetic appeal can go hand-in-hand with performance.

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