Bone-Conduction Amplification with Completely-in-the-Canal Hearing Aids

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Abstract

Recent advances in miniaturization have provided clinicians with hearing aids that can be comfortably inserted as far as the bony portion of the ear canal. It is possible to take advantage of these deeply inserted hearing aids in new ways. For example, the physical vibrations of microphone and receiver components may be used to improve hearing aid gain through bone conduction. Three cases are presented that will introduce this phenomenon for two transcranial CROS fittings and for one unilateral otosclerosis fitting. In each case, functional gain measurements under headphones were obtained with the hearing aid receivers acoustically plugged. Considerable gain was still present. Potential benefits, ramifications, and side effects are discussed.

Key Words: Probe microphone measures, transcranial CROS, unilateral hearing loss

Abbreviations: BTE = behind the ear; CIC = completely in the canal; CROS = contralateral routing of signals; HFA = high frequency average response values at 1000, 1600, and 2500 Hz; ITE = in the ear; OSPL90 = output sound pressure level with a 90 dB SPL input; REIG = real ear insertion gain; REIR = real ear insertion response.

he transcranial CROS approach to fitting a total unilateral hearing loss using full concha in-the-ear (ITE) hearing aids was formally described by Sullivan in 1988. Case examples and evaluation protocols for transcranial CROS fittings have subsequently appeared sporadically in the literature (Chartrand, 1991; Valente et al, 1995; Valente, 1995; Bauman and Braemer, 1996). Several authors (Sullivan, 1988; Chartrand, 1991; Valente et al, 1995; Valente, 1995) describe fitting procedures for powerful ITE and/or behind-the-ear (BTE) hearing aids. They advocate these devices on the basis that high sound pressure levels are required in the canal of the dead ear to generate adequate crossover amplification to stimulate the contralateral cochlea.

A different approach has been suggested by some investigators. Welling et al (1991) and

Pulec (1994) have used the Audiant bone-conductor implant on the side of the dead ear, the rationale being that the interaural attenuation for bone-conducted (mechanically coupled) stimuli is much lower than that of air-conduction (acoustically coupled) stimuli from the ear canal of the dead ear to the contralateral cochlea. Such a fitting should therefore cause fewer problems with acoustic feedback than a powerful acoustic hearing aid fitting. More recently, two investigators (Bauman & Braemer, 1996) have described a fitting protocol using completelyin-the-canal (CIC) hearing aids for transcranial fittings. They note that the deep insertion of the CIC instruments allows them to generate much higher sound pressure levels in the residual ear canal between the hearing aid and the tympanic membrane than can be obtained with much more powerful amplifiers in ITE or BTE hearing aids. They report that these higher sound pressure levels are the basis for the greater transcranial crossover amplification to the contralateral ear by CIC than by either ITE or BTE aids. They state that if the CIC is properly made to provide full contact with the walls of the ear canal beyond the second bend into the bony portion of the canal, then acoustic feedback will be minimized and sound pressure levels in

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the residual canal will be maximized to provide the best possible crossover amplification (improved acoustic coupling).

In this paper, three cases of CIC fittings are described for the purpose of illustrating the following points. First, CIC fittings can generate considerable crossover amplification. Second, CIC fittings provide such crossover amplification because the full contact to the bony portion of the ear canal improves the mechanical coupling to the bone, creating a second channel of crossover amplification. The mechanical coupling enables the vibration of the hearing aid components to be transmitted by bone conduction to the contralateral cochlea. Therefore, it is the mechanical coupling and not the higher sound pressure levels in the residual ear canal (acoustic coupling) that improves the crossover amplification of the CIC fittings. Third, if the mechanical coupling is responsible for a significant portion of the crossover amplification, then some of the verification techniques that work for ITE and BTE fittings, such as probe microphone measurements, are not an appropriate verification tool for CIC transcranial CROS fittings. Fourth, if mechanical coupling plays a significant role in transcranial crossover amplification for some fittings, then prefitting assessment procedures that rely on functional crossover measurements to prescribe frequency/gain characteristics cannot ignore that role. Fifth, it should be possible to take advantage of this mechanical coupling to fit large conductive hearing losses such as otosclerosis with CIC hearing aids instead of powerful ITE or BTE hearing aids. An excellent description of the problems in assessing boneconduction amplification, and the possible benefits in functional gain, can also be found in Carlsson and Hakansson (1997). Finally, whenever mechanical coupling is the primary mode for transmission of the signal then perhaps the hearing aid receiver can be replaced by a small vibrating device to further reduce acoustic feedback while still generating sufficiently high levels of amplification to be useful to the wearer.

CASES

Case 1

The first case, a 48-year-old woman (AP), suffered a complete unilateral hearing loss during the removal of a left-sided acoustic neuroma. On the right ear, pure-tone thresholds were 10 dB HL or better from 250 to 4000 Hz and 30 dB HL at 8000 Hz. AP purchased a left CIC



Figure 1 Two-cc coupler measurements of AP's hearing aid. [--] 60 and 90 dB SPL pure-tone sweeps at user volume setting. [--] 60 and 90 dB SPL pure-tone sweeps at the same volume setting but with the hearing aid receiver blocked with putty (i.e., no acoustic output).

hearing aid that maintained full contact with the walls of her ear canal to the second bend and was unvented. The hearing aid was adjusted to her most comfortable loudness use setting and she wore it at that setting for a 1-month period. The frequency/gain characteristics of the linear CIC at her use gain were measured in an Audioscan RM500[®] electroacoustic test box using a 2-cc coupler. The acoustic outputs of the hearing aid for 60 dB and 90 dB swept pure tones are shown



Figure 2 Transcranial crossover amplification for AP under headphones. "U" = unaided thresholds, "A" = aided thresholds, "B" = aided thresholds with no acoustic output from the hearing aid.

as solid lines in Figure 1. Functional gain measurements under TDH-39 headphones were obtained upon her return at the end of the 1-month trial period. The results are shown in Figure 2. The series of points designated by the letter "U" represent pure-tone crossover thresholds that were obtained by stimulating her right cochlea from the left unaided ear. After obtaining the unaided thresholds, the hearing aid was placed in her ear under the left TDH-39 headphone and the aided pure-tone crossover thresholds were measured. Those thresholds are designated by the letter "A" in Figure 2. The hearing aid receiver was then thoroughly blocked with plasticine putty to eliminate any acoustic output from the hearing aid. A second set of aided crossover measurements was obtained with the acoustically sealed hearing aid to determine how much of the aided crossover amplification was due to mechanical coupling. The acoustically blocked aided pure-tone crossover thresholds are designated with the letter "B" in Figure 2. The functional crossover amplification in the low frequencies, 250 Hz and 500 Hz, is dramatically reduced, as would be expected with no acoustic output. However, there was still considerable functional crossover amplification from the hearing aid in the frequency range from 1000 Hz to 6000 Hz-as much as 40 dB at 2000 Hz. The hearing aid was then removed and its acoustic output remeasured using the 60 and 90 dB swept pure tones in the test chamber while the receiver was still blocked with plasticine putty. The results of those measurements are shown as dashed lines in Figure 1. The difference in acoustic output from the unblocked to the blocked condition was a decrease of approximately 45 dB across the entire measured frequency range (250-8000 Hz).

Case 2

The second case, a 37-year-old man (GP), suffered a complete unilateral hearing loss during the removal of a right-sided acoustic neuroma. On the left ear, pure-tone thresholds were at 5 dB HL or better from 250 to 8000 Hz. GP's first attempted transcranial CROS fitting was a right-sided, full-shell, unvented, linear ITE hearing aid. The peak output sound pressure level with a 90 dB SPL input (OSPL90) was 123 dB SPL with an average output of 118 dB SPL. The highfrequency average (HFA), full-on gain characteristic of the aid was 54 dB with a peak gain of 63 dB at 2000 Hz. At the end of a 30-day trial with this aid, GP felt that it was of little benefit. He



Figure 3 Two-cc coupler measurements of GP's ITE hearing aid. [—] 50 and 90 dB SPL pure-tone sweeps at highest volume setting before audible acoustic feedback.

increased the gain until the onset of audible acoustic feedback, then backed it off slightly. His ITE was measured in the sound-treated test chamber at the gain setting obtained in this fashion. The acoustic outputs of the hearing aid for 50 dB and 90 dB swept pure tones are shown as solid lines in Figure 3. The HFA gain at this setting was 44 dB with a peak of 53 dB at 2000 Hz.

Soundfield functional gain testing was attempted using warble-tone stimuli. Frequencymodulated pure tones were presented via a wallmounted speaker at a distance of 1 meter and from 0 degrees azimuth. Unfortunately, his interaural attenuation was so great that he could hear each warble-tone stimulus through his better ear in the unaided condition at levels below his true crossover thresholds. This condition persisted even when his better ear was blocked. It was therefore impossible to measure true aided threshold improvements in the sound field.

GP switched from the ITE to a CIC fitting that was in contact with the walls of his ear canal to the second bend. The frequency/gain characteristics of the CIC in the test box for 60 and 90 dB SPL swept pure tones are shown as solid lines in Figure 4. As with AP aided versus unaided functional crossover gain was assessed under TDH-39 headphones. His unaided crossover thresholds, designated by the letter "U," are shown in Figure 5. His aided crossover thresholds are designated by the letter "A." He had measurable functional crossover amplification (>10 dB) from 1500 to 4000 Hz. The hearing aid receiver was then thoroughly blocked with plasticine putty to eliminate any acoustic output from the hearing aid. A second set of Ļ



Figure 4 Two-cc coupler measurements of GP's hearing aid. [--] 60 and 90 dB SPL pure-tone sweeps at user volume setting. [--] 60 and 90 dB SPL pure-tone sweeps at the same volume setting but with the hearing aid receiver blocked with putty (i.e., no acoustic output).

aided crossover measurements was obtained with the acoustically sealed hearing aid to determine how much of the aided crossover amplification was not acoustic. The acoustically blocked aided pure-tone crossover thresholds are designated by the letter "B" in Figure 5. Figure 5 shows that there was reduced but still considerable functional crossover amplification from 1500 to 4000 Hz. There was a fairly consistent

Figure 5 Transcranial crossover amplification for GP under headphones. "U" = unaided thresholds, "A" = aided thresholds, "B" = aided thresholds with no acoustic output from the hearing aid.

drop of 10 dB from the unblocked to the blocked functional crossover amplification conditions.

The hearing aid was then removed and its acoustic output remeasured using the 60 and 90 dB swept pure tones in the test box while the receiver was still blocked with plasticine putty. The outputs of the blocked hearing aid for 60 and 90 dB SPL swept pure tones are shown in Figure 4 as dashed lines. There was a decrease of 30 to 55 dB from the unblocked to the blocked conditions for the range of 250 to 8000 Hz. Figure 4 shows that there was no acoustic output whatsoever when the aid was measured in the blocked condition. Clearly, the acoustic measurements obtained in the 2-cc coupler are poorly related to GP's functional gain results. A more direct acoustic measurement of the aided output in his ear canal was undertaken to ensure that there was also no acoustic output from the hearing aid when it was in place for the functional crossover gain assessment. The aided real-ear insertion response (REIR) was measured with the hearing aid blocked and unblocked. The solid line in Figure 6 shows REIR of the CIC for a 60 dB SPL swept pure tone. This measurement was obtained just after the unblocked functional crossover gain measurement. There are 30 to 40 dB of real-ear insertion gain (REIG) at each frequency from 1500 to 4000 Hz. The dashed line in Figure 6 shows the REIR for the CIC with the receiver blocked by plasticine putty. This measurement was obtained just after the unblocked REIR and just before the blocked functional crossover gain assessment. There was an



Figure 6 REIR measurements of GP's hearing aid with a 60 dB SPL swept tone input. [---] REIR at user volume measured in his "dead" ear canal. [--] REIR in his "dead" ear canal with the hearing aid receiver blocked (i.e., no acoustic output).

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acoustic insertion loss of 5 to 15 dB at all frequencies from 1500 to 4000 Hz. The average decrease in acoustically measured REIG from the unblocked to the blocked conditions was generally 35 to 40 dB from 1500 to 4000 Hz.

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Case 3

The final case, a 35-year-old woman (HA), had left-sided otosclerosis. Her complete puretone audiogram is shown in Figure 7. It revealed normal hearing sensitivity in her right ear from 250 to 4000 Hz and a 35 dB HL hearing loss at 8000 Hz. She showed a profound mixed loss on the left side, which was predominantly conductive. Note that she had no measurable air-conduction thresholds below 1000 Hz, probably due to an overmasking dilemma. In Figure 7, the unconnected line of Xs represents her unmasked pure-tone thresholds through the left headphone. The unconnected line of Ns represents her unmasked pure-tone thresholds with her hearing aid in her ear and with no battery in place. The purpose of that measurement was to provide a more direct comparison to her aided thresholds, to be discussed later in Figure 8.

HA was fitted with a deeply inserted, nontapered, unvented CIC. The linear hearing aid



Figure 7 Audiogram for HA obtained under TDH-39 headphones. Both right and left ear air-conduction and bone-conduction pure-tone thresholds are shown. The symbol "N" shows the unmasked air-conduction thresholds obtained with the hearing aid in place in the left ear without a battery. Deviations from the regular unmasked thresholds from the left side "X" are due presumably to the presence of the hearing aid shell in the canal.



Figure 8 "N" = unmasked thresholds from the left headphone with the hearing aid in place and no battery in the aid, "A" = aided thresholds for pure tones from the left headphone, "B" = aided thresholds for pure tones from the left headphone with no acoustic output from the hearing aid.

had a full on HFA OSPL90 of 115 dB SPL in a 2-cc coupler and a peak gain of 50 dB with a 20-dB slope from 2000 Hz down to 500 Hz. At HA's use setting, where functional gain was assessed, the peak gain was 37 dB in the 2-cc coupler. Aided pure-tone thresholds for HA are shown in Figure 8. The line of Ns once again marks her unmasked thresholds with the hearing aid in place and no battery. The connected line of As indicates her aided thresholds for pure tones from the left headphone. We hesitate to refer to these as functional gain measures for reasons that are included in the discussion section. As with AP and GP, the acoustic output of the hearing was blocked and the aided pure-tone thresholds were obtained again. The acoustically blocked thresholds are shown as the connected line of Bs. In this case, the bulk of the amplification is maintained without acoustic output from the hearing aid. Mechanical coupling would seem to be to the primary mode of transmission of the amplified signal.

DISCUSSION

I n the case of AP, the amount of functional crossover gain measured under TDH-39 head-phones initially appeared to agree well with acoustic measurements of the same CIC hearing aid in a 2-cc coupler. When the receiver port

Journal of the American Academy of Audiology/Volume 9, Number 1, February 1998

of the hearing aid was acoustically sealed, the measured functional crossover was still present at 1000 to 6000 Hz. However, there was no measurable acoustic gain at all in the 2-cc coupler. Much of the functional crossover gain reported by AP and measured under headphones must be transmitted by other than acoustic means. We suggest that our efforts to provide a tight acoustic seal by obtaining a long, unvented CIC instrument whose medial end terminates with full contact at the bony portion of the ear canal has provided enhanced mechanical coupling to the temporal bone. It is the vibration of the hearing aid components mechanically coupled to the temporal bone that provides one source of transmission for transcranial crossover amplification. The peak frequency of mechanical transduction is probably the resonant frequency of the microphone and receiver combination for that given hearing aid. The tuning of the mechanical transduction may also be related to the resonant frequency of the temporal bone itself. The second source of transmission is the acoustic signal generated by the hearing aid. The difference in functional crossover amplification between the blocked and unblocked conditions is due to the loss of the acoustic contribution when the hearing aid is sealed.

Another point that we have tried to address here is the relationship of supra-aural earphone measurements to normal soundfield listening situations. It is possible that the pressure of the headphone may affect the physical coupling of the CIC to the ear canal. In the case of HA, we measured the crossover thresholds with the hearing aid in place but without the battery. The theory was that if the supra-aural headphones were adding pressure to the CIC, pushing it against the bony portion of the ear canal, then there would be better thresholds for the "no battery" condition (N) than for the "unaided" condition (U) in Figure 8. We thought that perhaps the CIC might couple the vibrating supraaural headphone directly to the temporal bone. However, there was no appreciable difference in the two sets of thresholds in Figure 8. The implication is that the aided functional crossover measurements are probably accurate, but the evidence is not conclusive. A more rigorous assessment technique will be required to put this question to rest.

The measurements made with GP's hearing aid confirm the findings obtained with AP. There are two channels of transmission from each hearing aid to the contralateral cochleas of these two people. One channel, albeit the less dominant one, starts with the acoustic output of the aid in the ear canal. The second, in these cases the predominant one, is the direct stimulation of the temporal bone via mechanical coupling of the hearing aid, which is physically in contact with that bone. In both cases, the total blockage of the acoustic output of their hearing aids still yielded considerable functional crossover amplification to the contralateral cochlea. The existence of mechanical coupling was further substantiated in the case of GP by REIG gain measurements. The REIG assessment verified an acoustic insertion loss when there was still considerable functional crossover amplification after the hearing aid receiver was blocked with plasticine putty.

This portion of GP's assessment raises another issue. Bauman and Braemer (1996) argued that CIC hearing aids provide better crossover amplification than ITE or BTE instruments. They reasoned that CIC instruments created a smaller residual ear canal volume than ITEs or BTEs. The smaller volume of air allows the CICs to generate higher sound pressure levels than the more powerful amplifiers available in the larger instruments. The REIG measurements obtained during GP's assessment indicate that it is the second mode of coupling (mechanical) provided by deeply seated CIC hearing aids and not increased acoustic sound pressure levels that is responsible for the improved crossover amplification that can be obtained with CICs. The mechanical coupling is far less likely to occur for most wearers of ITE or BTE instruments because the larger, more powerful components that they contain will not fit far enough into most ear canals to make physical contact with bone. Those hearing aids are limited only to acoustic coupling and therefore obtain less crossover amplification with more powerful components. There is one exception and that is the use, by one manufacturer, of a BTE hearing aid where the receiver has been removed from the casing and placed in a CIC shell inside the wearer's ear canal. This approach couples the benefits of the deeply inserted receiver to the increased power, flexibility, and battery life of the BTE.

One final point can be made with regard to all of the acoustic verification techniques used for both of these cases. It is obvious that there are two modes of crossover amplification with deeply seated CIC transcranial fittings. The predominant mode, mechanical coupling, cannot be assessed acoustically using a probe microphone system. In each case, the contribution of mechanical versus acoustic modes of

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transmission is different. Possibly, the extent and/or placement of contact for a specific hearing instrument within the bony ear canal may determine the contribution of the hearing aid's mechanical vibration to function crossover gain. Possibly, the extent to which the hearing aid components are damped or isolated from the hearing aid shell also affects the mechanical coupling. In that case, the choice of manufacturer becomes crucial since different manufacturers of CICs use different techniques to damp mechanical vibration; some are more effective than others. Perhaps the resonant characteristics of the individual temporal bone play a part. A further factor that cannot readily be assessed clinically is the resonant frequency of the specific microphone and receiver components of the device. If the frequency of mechanical vibration were known for a specific hearing aid or, even better, if it could be tuned and/or damped by the clinician, this would be a powerful fitting tool, especially for some conductive hearing losses. These two cases do not provide insight into those questions. What they do show is that acoustic verification techniques (probe microphone measurements) will not provide the whole picture as to what is happening with these fittings and are not recommended as part of the verification protocol. It is also clear that preassessment protocols that measure functional crossover amplification will be of limited value unless the precise type and extent of mechanical coupling can be simulated before the actual hearing aid is made.

In the final case of HA, a new issue is explored, that of fitting profound conductive or mixed hearing losses via mechanical coupling with CICs. This case is of particular interest for one additional reason. Note in Figure 7 the difference in the bone-conduction thresholds in the right versus the left ear. There appears to be better cochlear reserve in the right ear than in the left. There is a difference of as much as 30 dB at 2000 Hz and 4000 Hz. This difference may be larger than the interaural attenuation of the bone-conducted signal from the CIC in the left ear canal. If so, the amplification shown in Figure 8 may be due to a shift in thresholds for both ears, not just the left. Given the potential for overmasking the left ear, there did not seem to be any good way to separate which ear was hearing the pure tones from the left headphone when the hearing aid was acoustically sealed. Reliable thresholds could not be obtained when the right ear was masked. However, the improved thresholds for the aided condition versus the "no battery" condition clearly indicate that a good amplified signal was being perceived. The key issue may be whether interaural phase and timing cues, for example, will disrupt her ability to perform in noise if she hears a diotic presentation through bone conduction in addition to the normal signal in her right ear. More cases will be required to determine the extent to which this may be a problem. Possibly, tests such as binaural fusion or the Staggered Spondaic Words test may provide some insight.

Finally, were these three individuals satisfied with their hearing aids? Each person did offer some anecdotal remarks at the follow-up visits. The first two, GP and AP, who had transcranial fittings, reported success in the situations for which they purchased the hearing aids. Neither of them wear their hearing aids all of the time. Both purchased their devices for specific situations. GP uses her aid while working with children and reports a very natural sound quality as well as the perception of good directional hearing. AP has the distinction of being almost the only wearer of a CROS aid of any kind who prefers to wear the device in noisy situations. He uses it in large computer trade shows where he can talk on the phone through the hearing aid while working on the floor. By far the greatest success was with HA, who reports that she understands what "true stereo hearing" is like for the first time in her life. She wears the hearing aid all day, everyday, and reports excellent results, particularly when listening to music or at the theatre.

CONCLUSIONS

The cases summarized in this paper led to a series of conclusions regarding the use of deeply inserted CIC hearing aids for total unilateral hearing losses and some conductive hearing losses such as otosclerosis:

- 1. Deeply seated CIC hearing aids that terminate at the bony part of the ear canal do provide adequate crossover functional gain to fit total unilateral hearing losses.
- 2. Deeply seated CIC hearing aids often provide better crossover amplification than more powerful ITE or BTE hearing aids because they are mechanically coupled directly to the bony part of the ear canal. The vibration of the hearing aid components

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mechanically stimulates bone-conduction hearing even in the total absence of acoustic output from the hearing aid. Therefore, the CIC hearing aid stimulates the contralateral cochlea via two modes of transcranial coupling, acoustic and mechanical, whereas the larger hearing aids rely solely on acoustic coupling at high sound pressure levels.

- 3. Prefitting hearing aid assessments that rely on strictly acoustic measurement techniques such as probe microphone systems are not recommended for transcranial amplifier fittings involving CIC hearing aids. Strictly acoustic assessments will not provide valid predictions of crossover amplification for the purpose of fitting mechanically coupled hearing aids. Even deeply seated insert earphones cannot adequately simulate the extent and type of mechanical coupling provided by a hearing aid with an acrylic shell.
- 4. Strictly acoustic verification techniques (probe microphones) used in postfitting assessments are not valid for use with hearing aids employing mechanical coupling to the extent described in these cases.
- 5. Wherever the primary mode of coupling is mechanical, it should be possible to replace the hearing aid receiver with a completely sealed mechanical oscillator, decreasing the possibility of significant acoustic feedback.
- 6. It is possible to use a deeply seated CIC hearing aid to fit any conductive hearing loss where there is adequate hearing sensitivity at the level of the cochlea and where it is possible to obtain good mechanical coupling to the bony portion of the ear canal. The purpose of such a fitting would be to overcome the conductive hearing loss via direct bone-conduction stimulation. Normal

contraindications for deep canal fittings in the presence of middle ear effusion, prior mastoid surgery, and damaged tympanic membranes still apply.

- 7. Mechanical coupling may not be indicated for fitting mixed hearing losses in cases where the cochlear reserve of the contralateral ear is substantially better than that of the ear being fitted. Possible complications due to interaural phase and timing differences may degrade performance particularly in background noise.
- 8. Naturally, more research is desirable.

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