Trends In Amplification

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Directional Hearing Aids

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When asking individuals with hearing impairment to identify the situations for which the most communication difficulty is encountered, listening in noise seems to be nearly universally mentioned. In fact, I would venture that if queried, almost no patient would refuse the offer of improved levels for sounds of interest in relation to the level of background noise. Improving the signal-to-noise ratio (SNR) for listeners with hearing impairment has long been a goal of various amplification schemes. The reasons for attempting to improve the SNR delivered to hearing aid wearers are obvious given clear evidence of reduced speech recognition with increasingly less favorable SNRs. Moore (1989) suggested that individuals with normal hearing require a SNR of at least +6 dB for satisfactory communication. Unfortunately, data overwhelmingly show that individuals with sensorineural hearing loss generally require even more favorable SNRs for satisfactory communication than listeners with normal hearing (Carhart and Tillman, 1970; Cooper and Cutts, 1971; Dirks et al, 1982; Groen, 1969; Killion, 1997; Plomp, 1976; Schum, 1996; Sutter, 1985). Children with hearing impairment appear to be even more negatively effected by poor SNR (Boothroyd *et al.,* 1996; Crandell, 1993; Crandell and Smaldino, 2000; Finitzo-Hieber and Tillman, 1978), leading the American Speech-Language-Hearing-Association (Bess *et al.,* 1996), and others (Berg, 1993; Bistafa and Bradley, 2000; Blair, 1990; Smaldino and Crandell, 1995), to recommend SNRs of at least +15 to +30 dB in educational settings. Unfortunately, most classrooms have SNRs between –6 and +6 dB, making learning in such environments difficult (Bess *et al.,* 1984; Crandell and Smaldino, 2000).

Results from past investigations offer clear evidence that listening in poor SNRs is a significant problem for listeners with sensorineural hearing loss. Listening problems in noisy environments can be devastating, and lead some listeners with hearing loss to avoid difficult listening situations, resulting in withdrawal and greater isolation, potentially impacting their overall quality of life in a negative way (Jackson, 1997; McCay, 1996; Mulrow *et al.,* 1990).

Hearing aids represent a common rehabilitation method for listeners with sensorineural hearing loss. Hearing aids using standard (omnidirectional) microphones, while effective at increasing audibility for speech and other sounds, are largely ineffective in improving inadequate SNR conditions whether they use analog (Killion and Villchur, 1993; Killion, 1997; Plomp, 1978; Tyler and Kuk, 1989; Van Tasell, 1993; Verschuure *et al.,* 1999) or digital (Ricketts and Dahr, 1999; Walden *et al.,* 2000) techniques. In fact in some

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cases, SNR may be made worse by some hearing aid styles that have single omnidirectional microphones (Beck, 1983; Ricketts, 2000a). Listeners wearing hearing aids that have standard microphones are likely to require more favorable SNRs than listeners with normal hearing to accommodate the noise-related problems associated with their poorer hearing thresholds. This difficulty is highlighted in Figure 1 in which the aided sentence reception threshold as a function of pure tone average (PTA) hearing threshold for 70 listeners collected as part of two different investigations have been plotted (Ricketts *et al.,* 2001; Ricketts, unpublished data). The regression line through these data and the linear regression of the unaided data are also plotted. These data were all collected in a single listening environment designed to emulate a noisy restaurant, and the speech recognition results from these studies were averaged across three or four hearing aid models of the same style to obtain a more stable representation of performance. The test hearing aids included analog, digitally programmable analog, and digital signal processing (DSP) instruments. The SNR necessary for these listeners to achieve 50% correct sentence recognition performance was measured using two ten-sentence blocks of the Hearing In Noise Test (HINT; Nilsson *et al.,* 1992; Nilsson *et al.,* 1994).

Although there is considerable variance present in the data shown in Figure 1, as reflected by the moderate correlation (r = 0.59), two trends are apparent. First, in good agreement with previous data, listeners with greater hearing loss require more positive SNRs for equivalent sentence recognition performance (i.e., 50%). Second, and more pertinent to the current discussion, the relationship between SNR and hearing loss is remarkably similar across the aided and unaided listening conditions. Specifically, the average HINT performance for aided and unaided conditions is nearly identical for listeners with better hearing, suggesting that these hearing aids provided no SNR benefit over the unaided ear. For listeners with greater hearing loss, HINT performance is greater for the aided, as compared to the unaided condition; however, this is likely due to increased audibility for these listeners, rather than an improvement in SNR. These results are also in good agreement with surveys that indicate that listeners with hearing impairment commonly experience problems in noise, even when fit with amplification. For example, Kochkin (2000a) reported that 25% of individuals who own hearing aids but do not wear them cite poor performance in background noise as the major reason for hearing aid rejection.

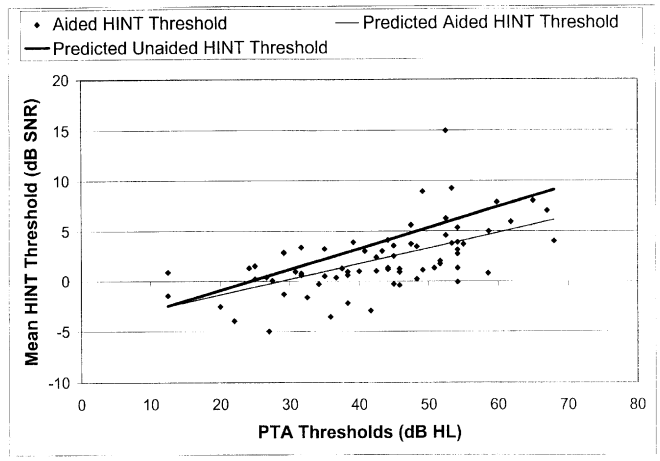

Figure 1. Mean aided HINT sentence reception thresholds, as a function of PTA hearing threshold, for 70 listeners. The predicted, aided, linear regression of these data is also plotted. For comparison purposes the linear regression of these same subjects' predicted unaided sentence reception thresholds, as a function of PTA hearing threshold, are also plotted.

Improving SNR through Amplification Systems

While existing hearing aids using omnidirectional microphones have generally failed to improve SNR, two different microphone techniques can improve SNR across a variety of listening situations. One technique places the microphone close to the sound source of interest. Without a doubt this method, exemplified by frequency modulated (FM) systems, provides the most SNR improvement (as much as 16 to 20 dB in noisy environments—Hawkins, 1984). While FM systems are *the best option* for SNR improvement in many environments, there are some potential drawbacks including concerns related to ease of portability, cosmetics, and reduced or absent overhearing abilities (Flexer, 1996; Lewis, 1991; Ricketts and Dittberner, 2002).

A second method, directional hearing aids, are advocated as a potential method for improv-

ing SNR in some noisy environments, while providing greater portability (since hardware external to the hearing aid is not needed) and alleviating some of the monitoring difficulties associated with FM systems fit to children. Directional hearing aids are designed to improve SNR based on the spatial location of the signal of interest relative to unwanted signals. Even though the magnitude of the improvement in SNR provided by directional hearing aids (approximately 3–6 dB) is much smaller than that reported for FM systems, they can still provide improved speech recognition across a range of noisy environments when compared to omnidirectional amplification.

The remainder of this article will focus on issues related to directional hearing aid technology and fitting. Can directional hearing aids improve the effective SNR for listeners with hearing impairment? Yes, but there are several limitations to this technology. In the following I will endeavor to describe the potential benefits and limitations of this technology and provide tools for quantifying directional hearing aids in the laboratory and clinic. Both behavioral (*directional benefit and performance*) and electroacoustic (*directivity*) directional properties will be discussed.

Directional Hearing Aids: Some History

Directional hearing aids were first introduced to the US market in 1971, and Rumoshovsky described a directional in-the-ear (ITE) instrument in 1977. By 1980 directional hearing aids represented almost 20% of the total hearing aids sold (Mueller, 1981). Their use steadily declined during the 1980s, despite numerous studies that suggested or measured additional benefit from directional hearing aids (Arentsschild and Frober, 1972; Frank and Gooden, 1973; Hawkins and Yacullo, 1984; Hillman, 1981; Lentz, 1972; Madison and Hawkins, 1983; Mueller and Johnson, 1979; Mueller *et al.,* 1983; Nielsen, 1973; Nielsen and Ludvigsen, 1978; Sung *et al.,* 1975).

There are several factors that may have contributed to the decline in sales of the first generation of directional hearing aids. In contrast to modern directional hearing aids, instruments of the 1970s and 1980s were limited by relatively large microphone size, little use in custom (in-the-ear and smaller) hearing aids, and limited ability to switch between directional and omnidirectional modes (Preves *et al.,* 1999; Ricketts and Dittberner, 2002). While some directional hearing aids allowed for switching through the use of a sliding cover that could be used to prevent sound from entering the rear port, the sliding cover was usually quite small, leading to potential problems for users with poor dexterity (Christensen, 2000). In addition to these limitations, early directional hearing aid technology had not developed to the point that it provided very large increases in directivity when compared to omnidirectional hearing aids. The poor directivity of these instruments resulted, in part, from attempts to maximize directivity in the free field, rather than when placed on the head (Killion *et al.,* 1998; Ricketts and Dittberner, 2002). It is well known that the angle-specific pattern of attenuation is significantly altered by the presence of the head and pinna.

The design of modern directional hearing aids has been modified to include a more in situ approach. That is, in the case of either single or dual microphones, many current manufacturers tune the response of the directional microphone system to provide maximum directivity across frequencies in situ. It is assumed that this design philosophy, coupled with more technologically advanced microphones, and methods are responsible for the better directivity reported for many modern directional hearing aids.

The interest in modern directional hearing aids began with the introduction of the first modern twin microphone hearing aid, the Phonak Audiozoom, in the early 1990s. This device and related research revealing excellent directional benefit in noisy environments (Valente *et al.,* 1995) is viewed by many as one of the primary events that have lead to the renewed popularity of directional microphones. A second major event impacting the renewed directional popularity was the introduction of the Etymotic D-Mic in 1997. The D-Mic, a directional + omni design, differed from previous directional microphones in that both the omnidirectional and directional microphones and microphone preamplifiers were housed within a single capsule. This design allowed for several hearing aid manufacturers to easily place a directional microphone in the faceplate of existing ITE products. The introduction of the D-Mic allowed many manufacturers to incorporate directional hearing aids in their product line without the expense and time involved with designing directional systems in a product specific manner. Consequently, the D-Mic may have al-

lowed for a greater number of manufacturers to quickly bring directional hearing aid products to the market, thus increasing the visibility of directional hearing aids in general. Since the introduction of the D-mic, Etymotic has continued to introduce increasingly smaller versions (ie, the sD-mic and the cD-mic), allowing for further use of directional microphones in increasingly smaller hearing aid shells.

How Directional Microphones Work

In general, the directional portion of directional hearing aids can be considered the front end or input stage to the amplification device. That is, the directional properties are applied to the incoming signal before the signal is further processed for the listener with hearing loss (ie, amplification, filtering, compression, etc). This is an important distinction in that the directional effect is applied independently of other signal processing. Consequently, the SNR advantage provided by a particular directional microphone design is expected to be of the same magnitude *regardless of other signal processing within the hearing aid,* whether it be analog or digital, compression or linear.

One common convention is to refer to devices that sample sound at only two locations as *first-order* directional microphones while referring to devices that sample at more than two locations as *second, or higher order* directional designs (including microphone arrays). A discussion of microphone arrays is beyond the scope of this manuscript and the interested reader is referred elsewhere (Ricketts and Dittberner, 2002).

Directional hearing aids operate by comparing incoming sounds sampled at two inlet ports (separated by 4–12 mm) located on the case of the instrument. Directional processing can be achieved using a single microphone and an acoustical phase shifting network (pressure gradient approach) or the electronic output of two separate omnidirectional microphones (Bauer, 1987; Ricketts and Mueller, 1999; Thompson, 1999; 2002). A schematic of the single microphone approach is shown in Figures 2A, B. As you can see there are two independent microphone ports (openings) with two sound pathways leading to either side of the diaphragm. Depending on the source location of a sound, it will arrive at the two microphone ports at different instants in time. For example, a sound arriving from directly behind the rear microphone port will have a travel time before it reaches the front microphone port. This travel time, often referred to as external delay, will be linearly dependent on the distance between the microphone ports, with greater separation resulting in greater travel time. The length of tubing that separates the opening on the case from the microphone diaphragm is often not the same for the front and rear openings. If however, if we make the assumption that these lengths are equivalent, we can calculate the external delay by dividing the port separation by the speed of sound. For example, a 12 mm port separation would result in an external delay of (12 mm ÷ 344 m/sec) 35 μs.

The use of the physical configuration shown in Figure 2A, sounds arriving directly from the side would enter both microphone ports at exactly the same time. Since the distance and travel time of the sound to either side of the waveform are equal, the sound will travel down both the front and rear openings and the sound pressure will reach either side of the diaphragm at the same time. This effectively cancels some of the energy, providing attenuation to signals arriving directly from the sides. Since travel time to the diaphragm for sounds arriving from directly in front or behind the microphone is different depending on whether the sounds travel down the front or rear opening (due to the external delay), the sound will not arrive at either side of the diaphragm at the same time and little or no cancellation will occur. This will result in a pattern of attenuation described as bidirectional or Figure 8.

In hearing aid wearers, competing noise signals in many environments are more likely to be behind or surrounding the listener, rather than directly off to the side. Consequently, it is usually of interest to provide attenuation at angles other than directly to the side. Using a similar single microphone scheme, a fine mesh screen (mechanical filter) can be placed along the sound pathway between the rear microphone port and the diaphragm (Figure 2B). This filter will delay the travel of sound resulting in an internal delay. By changing the value of this internal delay, different attenuation patterns can be achieved. For example, if the internal delay is set to a value similar to that of the external delay, sounds arriving from directly behind the listener will reach either side of the diaphragm at the same

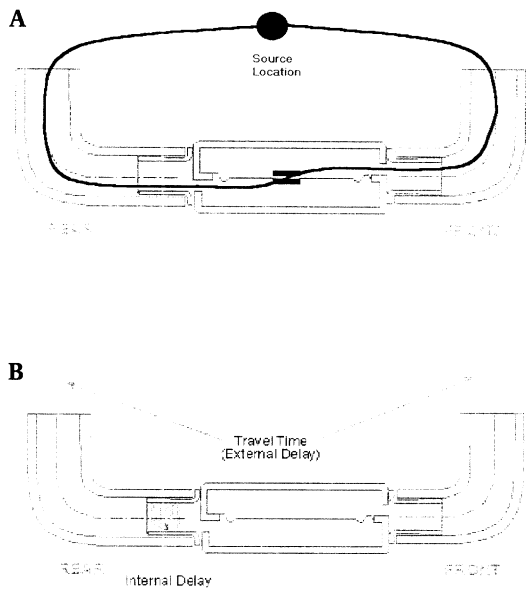

Figure 2. Schematics of directional microphone designs with **(A)** and without **(B)** a mechanical delay.

time and the greatest cancellation will occur for this angle (directly behind the hearing aid). The resulting attenuation pattern is commonly described as *cardioid.* Based on these design principles, we see that the specific relationship between internal and external delay can be adjusted to provide different, desired attenuation patterns. Figure 3 provides an example of the effect of changing internal delay on the directional pattern. In this figure, the magnitude of increasing attenuation as a function of angle is plotted, with greater attenuation occurring nearer the center of the figure. These polar plots are commonly used to describe the directivity of microphones and other devices. The theoretical directional patterns are plotted in Figure 3 assuming a port spacing of 7 mm and represent those obtained for a 2000-Hz stimulus. The magnitude of the cancellation for any particular angle is frequency-dependent. That is, the same microphone design will result in directional patterns that differ in magnitude and shape, depending on the frequency of interest.[1]

[1]The reader should note that the three internal delays presented in the Figure 3 example would result in bidirectional (0 ms), hypercardioid (10 ms), and cardioid (20 ms) patterns in the free field given the fixed 7 mm microphone spacing.

Directivity can also be achieved with two omnidirectional microphones and many of the same design principles apply. That is, the shape and magnitude of the pattern of attenuation are still dependent on the relationship between internal and external delay. The two microphone designs differ however in that the electronic output generated by the front and rear microphones are combined electronically for cancellation (a technique generally referred to as beamforming), rather than relying on acoustic cancellation at the diaphragm as in the single microphone approach. Since two different electronic signals are present the internal delay can be applied directly to the voltage from the rear microphone. It is important to note that both single and dual microphone designs rely on the same relationships to obtain directivity. Consequently, the limits of attenuation possible for single and dual microphone approaches are theoretically identical.

Single versus Twin Microphones for Directivity

The majority of current directional hearing aids implement a method for switching between directional and omnidirectional modes. In the case of hearing aids using two omnidirectional microphones to achieve directivity, the omnidirectional response is achieved by turning off (or ignoring) the output from the rear microphone, leaving the front omnidirectional microphone to function

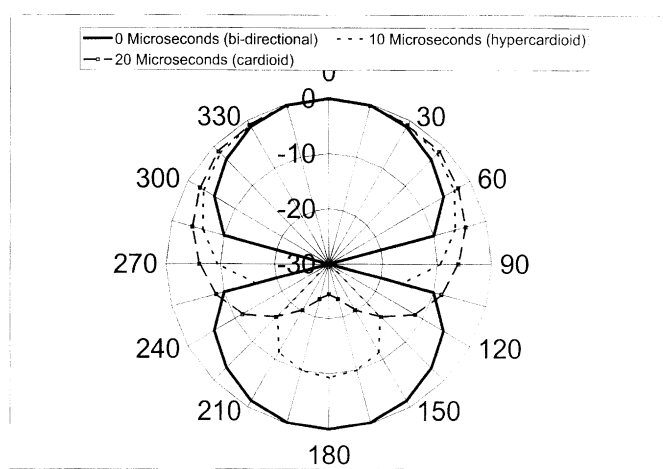

Figure 3. The theoretical, free field polar patterns that would be achieved assuming internal delay values of 0, 10, and 20 μs and a port spacing of 7 mm. These polar patterns represent those obtained for a 2000-Hz stimulus.

as the sole source of sound input. These directional hearing aids are generally referred to as *dual microphone* or *twin microphone.* Hearing aids using a directional microphone (a single microphone to achieve directivity) usually also incorporate a second, omnidirectional microphone for the omnidirectional response. These systems, commonly referred to as *directional + omni* and also contain two microphones. Directional + omni systems are easily identifiable because there are three microphone ports (two for the directional microphone and one for the omnidirectional microphone) visible on the hearing aid case. One unfortunate consequence of both designs including two microphones is that some manufacturers and researchers have chosen to refer to directional + omni as twin-microphone or dual-microphone directional hearing aids. While it is true that these instruments do include two microphones, only one is active at a time, in contrast to a dual microphone directional system. This overlap in nomenclature may lead to some confusion, although it is of limited importance given the similarity in directivity possible with both designs.

In the case of dual omnidirectional microphone design for directivity, it is important that the two microphones are matched in terms of the output they provide. A clear example of the negative impact of microphone mismatch is shown in Figure 4. In this figure the change in theoretical *directivity index* (DI) as a function of microphone mismatch is plotted based on mathematical predictions. The DI, described in detail later in this article, is the most common laboratory measure applied to the directivity of hearing aids. It can be calculated as the ratio of sound intensity (hearing aid output) for a sound source at 0 degrees azimuth (both the vertical and horizontal planes) to the intensity that would be produced in response to a diffuse sound source[2] of the same total acoustic power (Beranek, 1949; 1954). A hypercardioid design and microphone spacing of 10 millimeters are assumed for this example. In this example, the rather small mismatch of 1 dB results in a reduction in directivity of 4 dB at 500 Hz, with the greatest negative impact on directivity present in the low frequencies. Unfortunately,

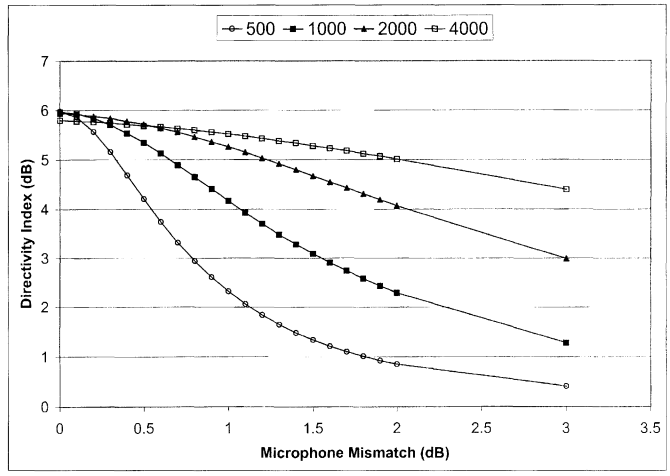

Figure 4. The theoretical change in DI values associated with increasing microphone mismatch across four frequencies. A hypercardioid design and microphone spacing of 10 mm are assumed for this example.

this negative impact is further exacerbated as port spacing is reduced. In Figure 5, the additional reduction in DI given a 1 dB microphone mismatch as port spacing is decreased from 12 mm to 4 mm is clearly evident. Obviously the importance of obtaining microphones that are closely matched for use in directional microphone hearing aids is imperative. Unfortunately, there is some concern that microphone sensitivity may drift over time, potentially causing a mismatch that was not present at the time the hearing aid was manufactured (Thompson, 1999). While the potential for microphone drift exists, the magnitude of this drift, and whether it actually occurs in hearing aids in real-world settings has not been demonstrated. Thompson (2002) contends that microphone drift rarely, if ever occurs. One might imagine, however, in the absence of true microphone drift, one or the other microphone ports may become partially or completely clogged by debris, resulting in a mismatch in sensitivity. Because of the potential problems with debris, audiologists are encouraged to check, and if necessary, clean the microphone ports during routine, periodic hearing aid evaluations. Regular electroacoustic evaluation of directivity (ie, front-to-back ratio), is encouraged to evaluate any changes in directional microphone operation.

Even if microphone drift proves to be a significant problem, technology exists to counteract

[2]A diffuse sound field is defined as having statistically uniform energy density and for which the directions of propagation of the noise waves are randomly distributed (Harris, 1991).

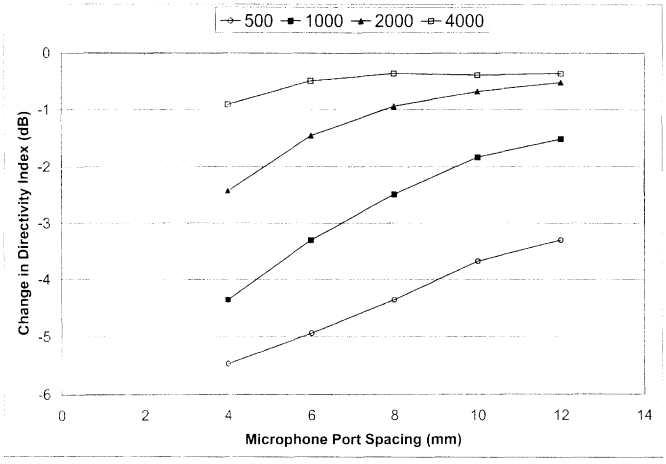

Figure 5. The theoretical change in DI values associated with a 1 dB microphone mismatch and a range of port spacing values. A hypercardioid design is assumed for this example.

it. In anticipation of potential microphone drift problems, a few manufacturers have introduced circuitry in their directional DSP hearing aids intended to counteract independent changes in microphone sensitivity. Specifically, some modern DSP circuits can be used to calibrate the relative outputs from the two microphones. Slightly more or less gain can be added to the output of one of the microphones prior to combining the two outputs for directivity. If A/D conversion (digitizing of the analog waveform) occurs prior to the combination of microphone outputs for cancellation, DSP processing can be used to adjust microphone output and counteract problems with microphone mismatch.

The stability of a single microphone directional design over time is also of interest. The timing of the two inputs to either side of the diaphragm depends on the properties of the mechanical filter and the distance traveled. While changes in travel distance over time are certainly unlikely, changes in the properties of the mechanical filter are certainly possible. For example, it is conceivable that dirt and moisture from the environment may affect the physical properties of the filter so that the magnitude of the internal delay is affected, in turn potentially impacting directivity. Unfortunately, as with the dual microphone design, the stability of the directivity of single microphone directional systems in real-world settings has not been systematically investigated.

Low Frequency Roll-Off

Due to differences in their physical design, the use of directional versus omnidirectional microphones will have an impact on a hearing aid's frequency response (Thompson, 1999). Specifically, directional microphones are less sensitive in the low frequencies than their omnidirectional counterparts, for sounds that arrive on-axis. This difference in frequency response between the low and high frequencies exhibited by directional microphones is due to differences in phase matching. The magnitude of the output from a directional microphone system is dependent on differences between the sounds arriving from the two microphone ports, with more similar signals resulting in less microphone output. Since low-frequency signals sampled at the two ports will be more similar in phase than high-frequency signals, a relative reduction in output for these low frequencies will occur. An example of this relationship, assuming a dual microphone directional system is shown in Figures 6A, 6B. A single waveform is assumed to originate directly in front of the listener. In this example, it is assumed that the waveform travels unimpeded through the front microphone port to the microphone and is converted to an electronic signal. This same waveform arrives at the rear microphone after the time necessary for external delay is converted to an electronic signal and then is further delayed by the internal delay mechanism. For this example, we are assuming the total delay (external + internal) is 35 μs and the port spacing is approximately 6 mm. At this point the two waveforms are combined to form a single output waveform. The two waveforms just prior to combination are shown in the figures as the front input and rear input waveforms, with the combined waveform displayed as the output. In Figure 6A we can see that the phase shift caused by the 35 μs delay is quite small relative to the wavelength of the 400 Hz signal. Consequently, the combined output signal is significantly reduced in intensity. In contrast, much greater phase misalignment is present for the 4000 Hz, and the resulting combined output is of much greater intensity signal (Figure 6B).

The frequency at which the low frequency roll-off begins is predictable on the basis of the spacing between the microphone ports with increasingly smaller separation resulting in the reduction in sensitivity occurring at increasingly *higher* frequencies. That is, the closer the micro-

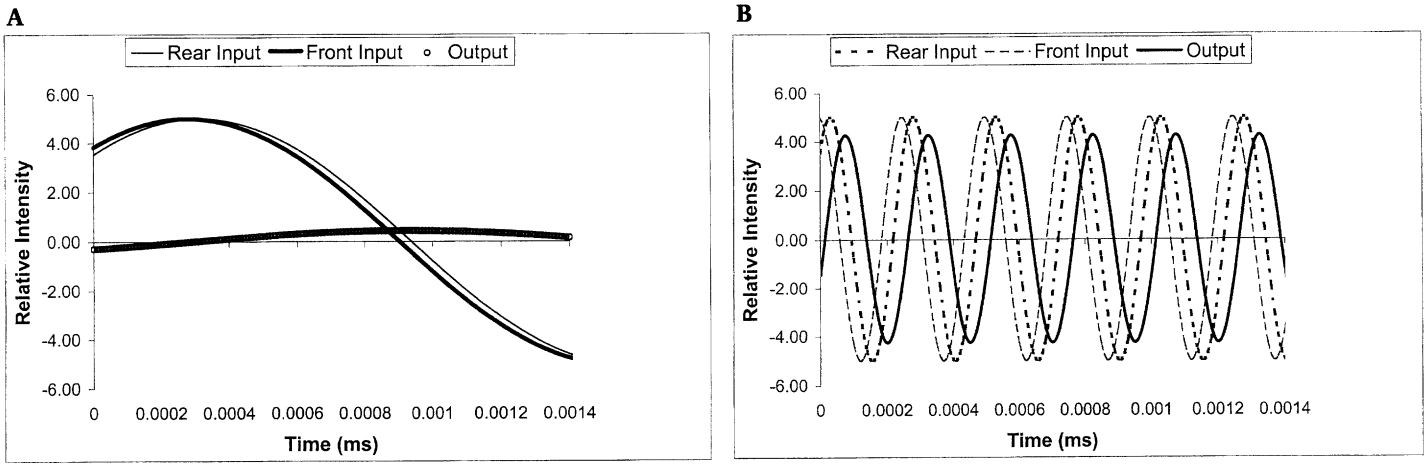

Figure 6. The relative reduction in hearing aid output in the low frequencies, assuming a dual-microphone directional system for the frequencies of 400 Hz **(A)** and 4000 Hz **(B)**.

phone ports, the greater the potential for reduced audibility of low-frequency sounds unless gain compensation is provided. Regardless of port spacing, the magnitude of low frequency roll-off is relatively constant at approximately 6 dB per octave. A theoretical example of this roll-off is shown in Figure 7. In this example, the frequency response of directional microphones (dual-microphone) with port separations of 12 mm and 6 mm are compared to those of a single omnidirectional microphone. In this case, the directional roll-off for the directional microphone with 12 mm port spacing leads to a reduction in sensitivity of about 17 dB at 500 Hz relative to the omnidirectional. However, the fact that roll-off begins at a substantially higher frequency for the directional microphone with 6 mm port spacing leads to an additional 6 dB reduction (a total of 23 dB) at this same frequency. The reader who is interested in the low-frequency roll-off in directional microphones as well as the previously discussed impact of microphone spacing, internal delay, and frequency on theoretical free-field directional patterns is referred to the excellent interactive Polar Primer available from Gennum Corporation (www.frontwave.com).

Since a reduction in low-frequency gain is associated with switching to directional mode, it may seem logical to fully compensate for this change. The decision of frequency response compensation, however, is made more difficult when the internal noise levels of the microphone are also considered. Unfortunately, while directional microphones have a reduced low-frequency response for incoming signals, this reduction does not positively impact microphone (internal) noise. In fact, in dual-microphone systems, the microphone noise of the two omnidirectional microphones is additive, resulting in an increase in microphone noise of 3 dB over a single omnidirectional microphone. When gain is provided to compensate for the directional low-frequency roll-off, that additional gain is also applied to the microphone noise floor. This increase in gain has the potential to increase microphone noise to a level that is audible, or perhaps even bothersome, to a listener if the listening environment becomes quiet enough. The amplitude expansion processing available in some DSP hearing aids represents a technology that can be used to offset the potential for increased audibility of microphone noise when listening in directional mode. Expansion can be thought of as the opposite of compression and works by reducing gain for increasingly lower level sounds below a fixed expansion threshold. As a consequence, hearing aids using expansion circuits are able to provide less gain for soft sounds (including microphone noise) in comparison to hearing aids that lack expansion, but are matched in all other ways (Bray and Ricketts, 2000).

Summary of Directional Hearing Aid Design Principles

It seems prudent to summarize our discussion of the design principles of directional hearing aids

to draw a few conclusions. Only select directional hearing aid design issues that were considered to be of the most potential interest to the readership were considered in the previous section. For further details and discussion of related topics, the interested reader is referred to the excellent review by Thompson (2002).

It is well known that directional microphones are able to selectively attenuate sound based on the angle of sound incidence. This selective attenuation is achieved by taking advantage of the timing differences present when the same sound is sampled at two or more locations. To operate, directional hearing aids must sample sound at two locations. This is accomplished through the use of a single microphone using two ports (opening to either side of the diaphragm), or two separate microphones. Different patterns of attenuation are then possible by varying the relationship between internal and external delay. Since the same physical relationships are used when achieving directivity through a single or dual microphone design, the theoretical limits of directivity of these two schemes are identical. It is also important to keep in mind that directional microphones differ in their frequency response compared to their omnidirectional counterparts, resulting in a reduction in low-frequency gain. This reduction in low-frequency gain may be compensated for by applying more gain through the hearing aid amplifier or microphone pre-amplifier; however, this gain will also be applied to the microphone noise floor.

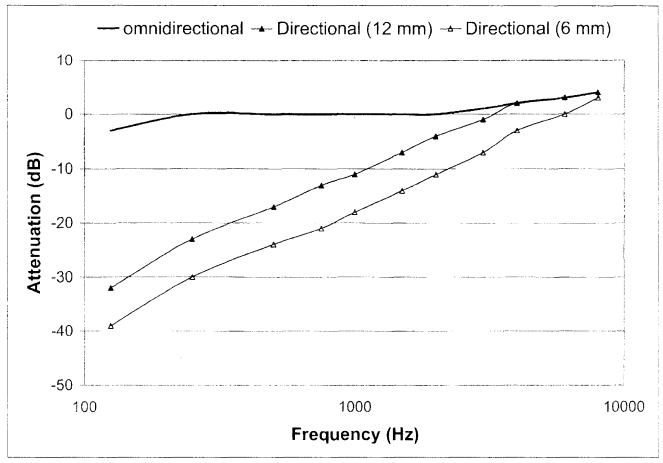

Figure 7. The relative frequency response of directional microphones (dual-microphone) with port separations of 12 mm and 6 mm versus that of a single omnidirectional microphone.

Verification and Validation of Directional Hearing Aids

Although directional hearing aids appear to be a sound theoretical idea, it is obvious that quantification of their operation is necessary to determine how they operate in practice. In the following section, quantification procedures will be described to lay the terminology groundwork for later discussions of real-world performance and benefit. There are two general categories of methods for assessing the SNR advantage provided by directional instruments: electroacoustic and behavioral evaluation. The general term *directivity* is commonly used to describe electroacoustic evaluation of directional properties. Conversely, the term *directional benefit* can be used to describe situations in which a person using a directional mode performs better than when using an omnidirectional mode. It is important that we differentiate between the measurement of directional benefit and performance (Ricketts and Mueller, 1999). Directional research across hearing aid brands sometimes reveals little correlation between listeners' relative performance with directional hearing aids and directional benefit (Ricketts, 2000b; Ricketts and Dhar, 1999; Ricketts *et al.,* 2001). This is not surprising since performance and directional benefit are assumed to reflect different aspects of the relationship between speech understanding and the hearing aid processing system. Specifically, performance (absolute score) is influenced by the hearing aid as a whole, including not just the directional microphone, but also all other signal processing and frequency shaping properties. In other words, which hearing aid is best? In contrast, it is assumed that directional benefit (difference score) reflects the impact of the directional microphone on the hearing aid processing system. That is, directional benefit is assumed to mainly reflect differences in the electroacoustically measured directivity of directional and omnidirectional instruments (Ricketts, 2000b; Ricketts *et al.,* 2001). In other words, how good is the directional microphone?

Measurement of Directivity

Although there are many ways to describe directivity, the three that are most commonly used

with directional hearing aids include *front-to-back ratio* (FBR), *directional patterns,* and the DI. FBR is differentiated from the other two methods as the only clinically viable directivity procedure. In contrast, directional pattern measurements and DI calculations require the use of an anechoic room. Directional patterns of hearing aids are commonly measured in a single (horizontal) plane and are graphically realized by a two-dimensional polar coordinate system such as in the *polar plots* shown in Figure 8. These single plane polar plots provide a quick and easy way to compare the directivity of hearing instruments. Directional patterns are also sometimes obtained in all planes of reference allowing for the visualization of the exact amount of attenuation provided by a directional hearing aid in response to signals presented from all directions in three dimensional space (Figure 9). Directional patterns involving all planes are obviously more representative in terms of describing the directivity of an instrument than directional patterns in only the horizontal plane; however, three-dimensional directional patterns require significantly more time and equipment to construct. Two-dimensional plots require less time and effort to construct, but since no information is available relative to the vertical plane, they may provide somewhat limited information about the attenuation properties

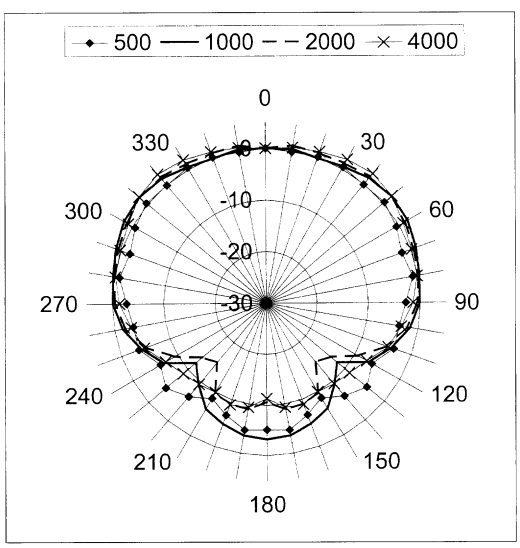

Figure 8. A polar plot constructed from the frequency specific (500, 1000, 2000, and 4000 Hz), free-field directional pattern measured from a single hearing aid in the horizontal plane.

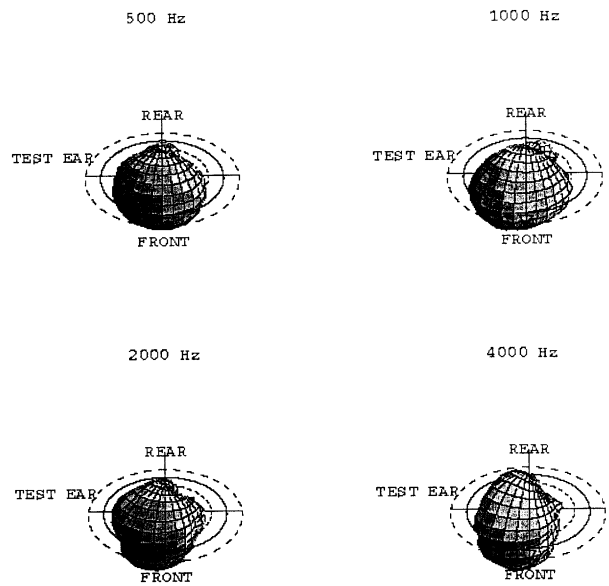

Figure 9. Frequency specific, spherical coordinate representations of the angular attenuation provided by a single directional hearing aid in response to signals (500, 1000, 2000, and 4000 Hz) presented from all directions in three dimensional space. Reprinted with permission (Etymotic Research, 2000).

of hearing aids in environments for which sound sources arrive above and below the horizontal plane through the listeners' ears. The interested reader is referred to Ricketts and Dittberner (2002) for detailed methods for the measurement of directional patterns.

The magnitude of relative hearing aid output is plotted as a function of the distance from the center of the sphere. That is, a smaller sphere is reflective of greater average attenuation. In addition, the angles of greatest attenuation, usually referred to as nulls, are displayed as indentations in the sphere. A brief visual inspection of Figure 9 reveals that the tested hearing aid provided much greater average attenuation for 4000 Hz than 500 Hz signals. This pattern is commonly seen in many in-the-ear directional hearing aids. The angular resolution of measurement must be chosen prior to measurement of either two- or three-dimensional directional patterns. The angular resolution simply describes the magnitude of angular separation between each individual measurement. A measurement increment of 10 degrees is probably appropriate for current directional microphone technology incorporated in hearing

aids. It is also common to construct polar plots for several key frequencies (eg, 500, 1000, 2000, and 4000 Hz), because the directivity of a hearing aid is usually not equal across frequency. Observe, for example in Figure 8, at a 180-degrees azimuth there is only a 5 dB attenuation for a 500 Hz signal, but 11 dB attenuation for a 2000 Hz signal (relative to 0-degrees azimuth).

While directional patterns can provide detailed information relative to the attenuation provided by a hearing aid across angles, it is sometimes difficult to visualize the total impact of this attenuation in specific listening environments. Fortunately, DI provides a single number calculation that is representative of the frequency specific spatial attenuation properties that are displayed in directional patterns. The DI of hearing aids is of interest since it is assumed that it approximates the *effective SNR* for a condition in which the signal of interest originates directly in front of the hearing aid wearer and a fully diffuse noise field of the same total acoustic power is present. That is to say that (not surprisingly), the magnitude of directivity is related to the magnitude of directional benefit (Killion *et al.,* 1998; Mueller and Johnson, 1979; Ricketts and Dittberner, 2002; Sung *et al.,* 1975). DI in most amplification systems designed for the hearing impaired varies from approximately –3 dB to perhaps +12 dB in some microphone array systems. Hearing aids that are equally sensitive to sound arriving from all angles (true *omnidirectional*) will have a DI = 0dB. The reader should be aware, however, that omnidirectional hearing aids will not be truly omnidirectional when placed on the head. When a hearing aid is generally more sensitive to sounds arriving from directly in front of a listener, in comparison to sound arriving from all other angles, the DI will be positive. In the unfortunate case for which sensitivity to sound is generally poorer for sounds arriving from directly in front of a listener, in comparison to sound arriving from all other angles, the DI will be negative. Since DI is assumed to provide a reasonable estimate of effective SNR some predictions concerning speech recognition can be made based on DI. For example, a hypercardioid microphone would be expected to provide better attenuation for diffuse noise in the free field than a cardioid design since these designs have theoretical DI values of 6 dB and 4.8 dB, respectively.

While the DI appears to be a well-accepted measure, differences in methodology across laboratories can impact DI values. Specifically, DI values varying by as much as 1.2 to 1.95 dB (depending on frequency) have been reported for the same open ear (Knowles Electronic Manikin for Acoustic Research—KEMAR) condition (Bentler and Dittberner, 1999). In practice, the DI of hearing aids is usually calculated from two-dimensional directional patterns, three-dimensional directional patterns, or diffuse field versus free field measures.

The calculation method proposed by Beranek (1949) and shown below is usually used when a three-dimensional directional pattern data is used. This formula assumes a constant radius and equal division of the surface regions and thus requires no weighting.

$$DI = 10Log_{10}\left\{\frac{4\pi|P_{ax}|^2}{\int_0^{2\pi}\int_0^{\pi}|P(\theta,\phi)|^2|\sin\theta|\,d\theta\,d\phi}\right\}$$

In this formula, $|P(\theta,\phi)|^2$ is the magnitude of the mean squared sound pressure at all horizontal (θ) and vertical (ϕ) measurement angles; $|P_{ax}|^2$ is the magnitude of the on-axis (0-degrees azimuth) mean square sound pressure; and $|\sin\theta|$ is the absolute value of the sine of each measurement angle θ. The double integral defines the regions off-axis. One integral covers the horizontal region from 0 to 360 degrees, whereas the other integral covers the vertical region from 0 to 90 degrees. This, as can be deduced from the equation, covers an entire surface region of a sphere. DI measurement and calculation may be simplified to only use directional pattern data collected in the horizontal plane if symmetry along the vertical axis is assumed. That is, hearing aid output in response to the input measured at each angle in the horizontal plane is assumed to be the same as that measured anywhere along the intersecting vertical plane (in practice a deviation from the height of the hearing aid microphone).

In contrast to DI calculations based on free-field only measures, calculations based on free-field versus diffuse data require access to both an anechoic chamber and a reverberation chamber. Specifically, DI is calculated as the output pressure from a hearing aid (P) in response to a signal presented from the reference angle in a free field (anechoic chamber), versus the output in a diffuse field (reverberation chamber). DI is then calculated using the following formula (adapted from Roberts and Schulein, 1997):

$$DI = 10Log_{10}\frac{P_{diffuse}}{P_{free}}$$

A complete discussion of the intricacies of DI calculation are beyond the scope of this paper; however, a number of authors have argued over calculation specifics (Beranek, 1949; Bobber, 1974; Davis, 1973; Dittberner, 2001; Gerzon, 1975; Wilson, 1973). While the measurements and calculations necessary to derive DI by the methods mentioned here certainly differ in time consumption and complexity, both the diffuse and free-field, three-dimensional calculations are assumed to yield equivalent results for most modern hearing aids (Roberts and Schulein, 1997). Calculation of DI based on two-dimensional polar patterns has been shown to lead to erroneous conclusions in some situations (Dittberner *et al.,* 2001; Egge *et al.,* 2001); however, two-dimensional DI calculations also provide a reasonable approximation of true three-dimensional measures (Roberts and Schulein, 1997). The interested reader is referred to Ricketts and Dittberner (2002) for detailed information regarding measurement of directivity and DI calculation.

Killion and associates (1998) have advocated an articulation index weighted DI (AI-DI) as an enhancement to the traditional DI. It has been proposed that the AI-DI provides a reasonable estimate of the improvement in speech recognition in noise afforded by directional hearing aids. The AI-DI uses band importance weightings to assign more importance to the directional advantages for the frequencies most important for speech intelligibility. The premise of AI-DI is based on evidence from AI theory that various frequency regions of speech differ in their importance for understanding (DePaolis *et al.,* 1996; French and Steinberg, 1947; Pavlovic, 1987). Consequently, improved directivity in the most important regions of speech should be weighted more heavily, providing a more accurate estimate of the impact of the directivity of instruments on speech recognition. To date, the majority of reported AI-DI calculations have been based on speech weighting from the Mueller-Killion Count-The-Dots speech spectrum (Mueller and Killion, 1990), although any AI band importance function could conceivably be used. The band importance functions, taken from the Mueller-Killion index are 500 Hz = 20%; 1000 Hz = 23%; 2000 Hz = 33%; and 4000 Hz = 24%. An example calculation of the AI-DI versus a simple DI average is provided below to further explore its impact.

AI-DI Weightings	Sample DIs
500 Hz .20	1 dB
1000 Hz .23	1 dB
2000 Hz .33	4 dB
4000 Hz .24	2 dB

To obtain the average DI we would take (1 + 1 + 4 + 2) = 8 and then divide by 4 (frequencies) to find an average DI = 2.0 dB. To calculate the AI-DI we take [(0.2 * 1 dB) + (0.23 * 1 dB) + (0.33 * 4 dB) + (0.24 * 2 dB)] = 2.2 dB. So it can be seen in this example that AI-DI is 0.2 dB greater than a simple average. From the weighting values and this example the reader can see that the result of applying AI-DI is that more emphasis and importance are given to the directivity in the high frequencies. This example demonstrates that the impact of the four-frequency AI-DI calculation is quite small (0.2 dB). The difference between AI-DI and simple-average DI values, however, could potentially be greater if applied to, for example, one-third octave frequencies, rather than only these four octave frequencies. While the AI-DI certainly has intuitive appeal and would provide a relatively easy method for comparing directivity of hearing instruments, it has yet to be systematically investigated.

Measurement of Directivity: FBR

The equipment and time involved in measurement of DI is certainly well beyond what could be expected to be reasonable in routine clinical practice. Fortunately, there is a clinical method for quantifying directivity that is simple to perform using most commercially available probe microphone equipment. The front-to-back ratio (FBR) is the frequency-specific difference between the output level of a hearing aid in response to a sound source placed directly in front of a listener (0 degrees azimuth) versus that measured for the same sound source placed directly behind the listener (180 degrees azimuth).

Due to the potential interaction between the placement of the sound source and the angle of the polar null of a particular design, the FBR can provide misleading results if used for comparisons of absolute directivity across hearing aids. For example, most directional hearing aids with a car-

dioid pattern will have much larger FBRs than those with hypercardioid patterns, even though the DI associated with the cardioid pattern is usually smaller (as previously described). This is due to the fact that the cardioid pattern has a polar null (angle of greatest attenuation) at 180 degrees. The FBR is therefore not recommended for making comparisons across different hearing aids models.

The measurement of FBR using probe microphone equipment was previously described by Mueller (1992). While two separate real-ear aided responses (REARs) can be measured and the FBR calculated as the difference, it is possible to use the majority of commercial probe microphone equipment in a novel way to measure FBR directly as explained in the following.

1. Seat the hearing aid wearer on a swivel chair a fixed distance from the probe-microphone loudspeaker (usually 18 in. to 1 m will be sufficient, depending on the specific probe microphone system). Obviously, the closer the speaker is to the patient the less difficulty one may have with poor SNR biasing the results. Placing the speaker two close, however, can lead to increased variability for on-axis measures due to changes in head shadow.
2. Disable the reference microphone.
3. Measure the output of the hearing aid with the test loudspeaker directly behind the hearing aid wearer (180 degrees azimuth). Record this as the unaided response (REUR/REUG). It is important to be sure that the patient's head is fixed and facing directly away from the loudspeaker.
4. Swivel the patient to directly face the loudspeaker (0 degrees azimuth) and again measure the hearing aid output. This time, save the measure as the aided response (REAR/REAG). Again, make sure the patient's head is fixed and this time it should be directly facing the loudspeaker. *Since the reference microphone is disabled, it is important that the audiologist ensures that the distance from hearing aid to speaker is the same for both measurements.* Some clinicians may find it useful to use a length of string attached to the loudspeaker to ensure that loudspeaker to hearing aid distance remains constant across all FBR measures.
5. Most probe microphone systems will automatically calculate insertion gain as REAR-REUR (or REAG-REUG); however, following the methodology above what the system calculates as REIG is actually FBR.

Once FBR measures are made for several hearing aids, clinic or patient/instrument specific normative values can be generated for comparison to future measurements. These data can be used to easily assess the functioning of the directional microphone in general, or the influence of patient specific factors such as venting. Repeat measurement of FBR at periodic hearing aid checks can also be useful for verifying complaints of reduced directivity. Measurements at angles other than 180 degrees can also be useful, especially when evaluating directional instruments that exhibit the greatest attenuation for sounds other than those arriving from 180 degrees (ie, instruments with hypercardioid patterns). In addition, continual measurements can be made while the patient is being swiveled, if the probe microphone system being used is capable of real time output and measurement. This allows for visualization of the angles of greatest attenuation (polar nulls).

One other suggestion for clinical FBR measures is to always measure using the same compression parameters, or (when possible) set to linear processing. Compression will impact FBR in the same way that it does other traditional directivity measures, resulting in an apparent reduction in the true value of directivity. For example, see Figure 10 to examine the impact of compression on the FBR of the directional microphone. If an input of 65 dB SPL is sent from directly in front of the patient (left hand panel), the microphone will provide no attenuation and the compression circuit sees 65 dB SPL, and assigns gain accordingly. If instead, the same signal arrives from directly behind the listener, the directional microphone will attenuate the signal 12 dB (righthand panel). In this case, the compression circuit sees a 53 dB SPL input. If this is a low-threshold WDRC circuit with 2:1 compression, it will provide 6 dB more gain for this signal than for the 65 dB SPL input that arrived from the front. Consequently the FBR will be measured as 6 dB rather than 12 dB. In more general terms, this example shows that the true magnitude of attenuation provided by directional hearing aids with low-threshold compression will be underestimated by FBR.

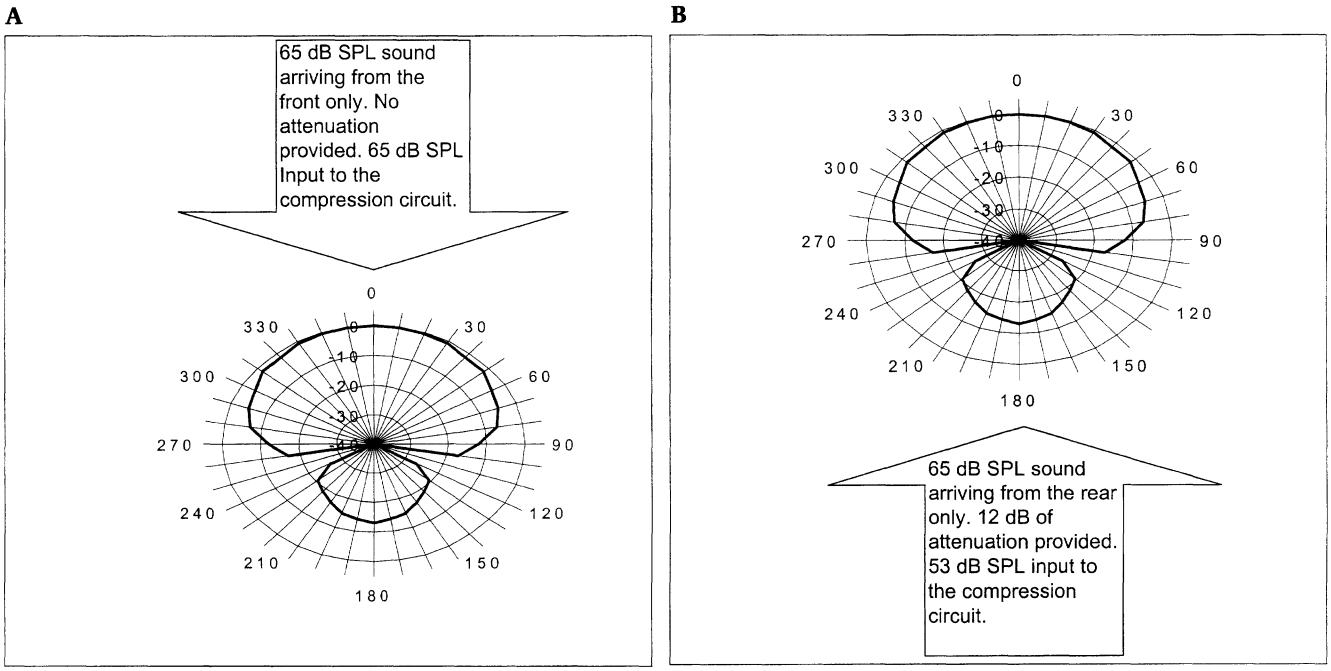

Figure 10. An example of the potential interaction between compression and traditionally measured FBR. Panel **A** displays the measurement of a signal located at 0 degrees azimuth and Panel **B** displays the measurement of a signal located at 180 degrees azimuth.

Behavioral Evaluation of Directional Hearing Aids

While quantifying directivity is of interest, of greater interest clinically is the level of additional benefit directional hearing aids will provide to individuals having hearing loss. As with hearing aids in general, the impact of directional hearing aids can be quantified using objective measures of speech recognition as well as subjective measures of the perception of sound quality, benefit, performance and satisfaction. By far the most common method for assessing the impact of directional hearing aids has been the quantification of changes in speech recognition in noisy environments. As noted previously, the positive impact of directional hearing aids on word recognition has been reported in terms of absolute scores (directional hearing aid performance) as well as difference scores between directional and omnidirectional hearing aid conditions (directional benefit).

Studies that have examined the magnitude of directional benefit provided by modern directional hearing aids often provide quite disparate results. When measured in traditional laboratory settings, directional benefit values ranging from approximately 5.5 dB to 11 dB and 40% to 70% have been reported in the literature. Studies, however, that have evaluated directional benefit in noisy environments designed to emulate difficult real-world conditions *generally* report values less than 6 dB and 40% (Preves *et al.,* 1999; Pumford *et al.,* 2000; Revit *et al.,* in review; Ricketts and Dhar, 1999; Ricketts, 2000b; Valente *et al.,* 2000a; Voss, 1997). While some of the variance in the reported directional benefit values probably is due to true differences across hearing aid models, it is also clear that differences in measurement parameters such as SNR of the test material, speech testing method, and the test environment also can impact results.

1. Test Materials: SNR and Method

Two general signal-to-noise protocols have been used to assess directional benefit. One approach is to select a single or group of predefined SNRs

(Mueller and Johnson, 1979; Ricketts *et al.,* 2001; Voss, 1997). If a single SNR is selected, directional benefit is most often reported as differences in percent correct scores between directional and omnidirectional hearing aid conditions. Obtaining a percent correct score from a single fixed SNR has the advantage of providing straightforward information about improvement that is easy to explain to patients. It may be difficult, however, to determine the proper SNR to select. A second approach is to vary the SNR, and measure the improvement in the threshold signal-to-noise ratio necessary for 50% correct performance (eg, Agnew and Block, 1997; Killion *et al.,* 1998; Madison and Hawkins, 1983; Ricketts, 2000b).

The use of variable SNR tests (eg, speech in noise [SIN] test, Etymotic Research, 1993; hearing in noise test [HINT], Nilsson *et al.,* 1994), while providing useful information concerning the magnitude of directional benefit, can sometimes be difficult to generalize to listening situations in an individual's real-world listening environment. For example lets examine a case involving two patients who are similar to many patients we see clinically—Lucky and his friend Amanda. Lucky and Amanda have similar hearing losses and are fit with identical directional hearing aids. Clinical evaluation using an adaptive SNR test indicates that both Lucky and Amanda receive 6 dB of SNR advantage in directional mode when compared to omnidirectional mode. Lucky reports he finds benefit in the directional mode at the shopping mall, but seems to get little benefit at weekly card games with friends. Amanda reports a contrasting experience in that she thinks that she receives little directional benefit at the shopping mall, but finds significant benefit in the directional mode when playing cards. The reason for this apparent dichotomy can be explained by examining the relationship between these two listeners' performance across various SNRs when fit with an omnidirectional hearing aid (Figure 11) and the SNR present in these two listening environments. Let's assume that the average SNR at the mall is 0 dB and the average SNR at the card game is +10.

Lucky's performance across SNRs (Figure 11) suggests that when listening in the mall he will score approximately 85% in directional mode and 50% in omnidirectional mode, revealing a directional benefit of 35% (see the brackets on Figure 11). In the case of the card game, however, he will score approximately 95% in omnidirectional mode and 100% in directional mode, so that same 6 dB of directional benefit only results in an improvement in speech recognition of 5%. That is, he is already doing very well at the card game and the directional microphone offers little additional benefit.

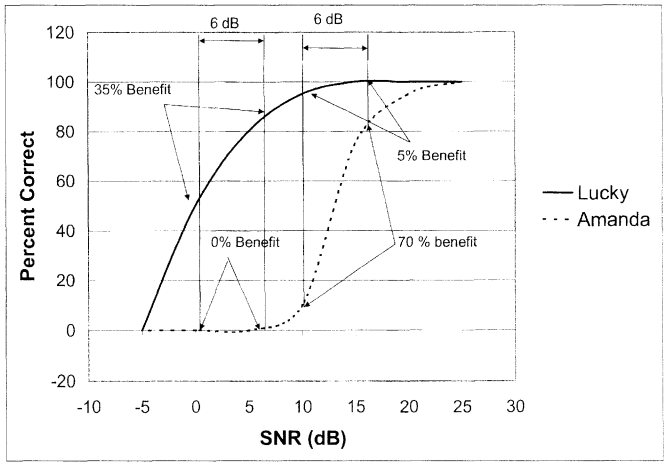

Figure 11. An example of two hypothetical listeners' speech recognition performance as a function of SNRs. Both patients are assumed to receive 6 dB of directional benefit.

Contrasting results are found when examining Amanda's performance across SNRs. Specifically, when listening in the mall she will score approximately 0% in both directional and omnidirectional modes. That is, directional benefit is not measured because the directional microphone is not able to raise the SNR to a value that will allow her to understand speech. In the case of the card game, however, she will score approximately 80% in directional mode and 10% in omnidirectional mode, revealing a directional benefit of 70%! So both Lucky and Amanda receive significant directional benefit but the environments that they benefit most in differ.

The previous example appears to support the choice of multiple or variable SNRs over a single fixed SNR when evaluating directional benefit. Variable SNR tests however, are open to some bias as well, because they are not limited to real-world SNRs. An extreme example would be the case of an individual that reveals 10 dB of SNR benefit, but their omnidirectional performance occurs at –15 dB, a level that rarely, if ever, occurs in the real world. While this example is unrealistic, it is provided to highlight the potential for bias using variable SNR tests. While using a

variable or adaptive SNR method has the advantage of eliminating the selection of the optimal ratio for testing, improvement as measured by SNR may not be as salient for counseling our listeners with hearing loss.

2. Test Environment

There are several test environment factors that are known to impact the magnitude of directional benefit. These factors include number and placement of competing noise sources, reverberation, room size and distance from listener to talker. Any one of these factors can be manipulated to increase or decrease directional benefit relative to a real world average. In fact, directional benefit can be reduced to near zero, simply through manipulation of these factors, over the range that occurs in the real world. There are a variety of real world environments for which no directional benefit would be measured, or would be expected.

It is well known that increasing reverberation can reduce both speech recognition in general and the magnitude of directional benefit (Hawkins and Yacullo, 1984; Hawkins, 1986; Leeuw and Dreschler, 1991; Madison and Hawkins, 1983; Moncur and Dirks, 1967; Ricketts and Dhar, 1999; Ricketts, 2000b). It has been shown that the degradation in speech understanding with increased reverberation is more pronounced in children (Hawkins, 1986), adults (Payton *et al.,* 1994), and elderly (Divenyi and Haupt, 1997) persons with hearing loss than age-matched listeners with normal hearing. Unfortunately, directional benefit is often quantified in sound treated rooms. The average reverberation time measured in such settings is approximately 100–300 ms, in contrast with the 600 to 1500 ms often measured in average rooms (Moncour and Dirks, 1967; Nabelek and Mason, 1981).

Reverberation time is defined as the duration required for a sound to decrease in intensity by 60 dB after the sound has been terminated. When a speaker communicates with a hearing aid wearer, some of the speech signal reaches the listener's amplification system directly and within a few milliseconds. The remainder of the signal strikes surrounding areas and thse reflections reach the listener's ear a few milliseconds after the initial signal. That is, the angle of arrival for the majority of sound sources in reverberant, real-world environments cannot be represented as a single point source. Sound sources presented in enclosed rooms (indoor environments) can be described as being comprised of a direct sound that arrives first from the azimuth of the source, followed by a relatively diffuse sound made up of reflections that may arrive from a variety of directions (Berenek, 1954). As distance between the source and the listener increases, the proportion of reflected versus direct energy also increases. This increase continues through the point, referred to as the *critical distance,* at which the proportion of direct to reflected sound energy is equal. The magnitude of critical distance generally increases with increasing room size and sound source directivity while it is inversely related to the magnitude of the reverberation time. Since directional hearing aids must be able to distinguish between the signal of interest and the competing signal based on their relative positions, it seems likely that little or no directional benefit will be measured in noisy environments when there are high levels of reverberation and the speaker to listener distance is great. Hawkins and Yacullo (1984) examined the magnitude of directional benefit in environments that differed in their reverberation time and in which the speech signal of interest was placed at 0 degrees azimuth at critical distance. Depending on the specific environment, critical distances were between 2.2 and 3.3 m. A single competing noise was presented at 180 degrees azimuth. An adaptive SNR presentation method using NU-6 words was used as the test material. These authors reported a decrease in directional benefit of approximately 4 dB as reverberation was increased from 600 to 1200 ms. The impact of reverberation on performance was even more evident as performance with directional hearing aids was reduced by as much as 10 dB with increasing reverberation.

More recently Leeuw and Dreschler (1991) examined the issue of critical distance as related to directional and omnidirectional performance in a series of three experiments. In the first experiment the change in hearing aid frequency response and the corresponding speech reception thresholds for patients were measured in two environments which differed in terms of their reverberation times. These measurements were made for a single speech source fixed at 0 degrees azimuth and a single competing noise fixed at the angles of 0, 45, 90, 135, or 180 degrees. Speaker

to listener distance was fixed at 1.4 M. This distance placed loudspeakers within the critical distance in the less reverberant environment and outside the critical distance in the more reverberant environment. As expected, in the non-reverberant environment, results revealed that directional benefit was greatest when competing noise sources originated in the rear hemisphere. Significant directional benefit was present in the reverberant environment; however, the amount of directional benefit was relatively independent of the origination angle of the competing noise source. These somewhat surprising results were attributed to the attenuation of reflected energy in the rear hemisphere across all conditions. It is not clear, however, how this argument could result in directional benefit for the condition in which the competing noise source and the speech source originated from the same loudspeaker.

A second experiment was performed by Leeuw and Dreschler in order to more systematically examine the impact of distance on speech reception thresholds in a reverberant room. In this second experiment a single speech source fixed at 0 degrees azimuth and a single competing noise fixed at 180 degrees azimuth were used. Two speaker to listener distances 0.5 M (within critical distance) and 1.4 M (beyond critical distance) were evaluated. Results revealed that SRTs were significantly reduced with increasing distance for both the omnidirectional and directional microphone conditions; however, there was not an interaction between microphone type and distance. That is, directional benefit was not impacted by distance. These results are in obvious opposition to previous findings (Hawkins and Yacullo, 1984; Madison and Hawkins, 1983).

These studies suggest that aided speech recognition performance in noisy, reverberant, environments generally decreases with increasing listener to source distance. This decrement occurs even when the source level is held constant at the listener's ear. The impact that increasing distance has on directional benefit is less clear, however, and further research that varies source to listener distance and critical distance in an independent manner is still needed.

Despite the data of Leeuw and Dreschler (1991), clinical experience and the data of Hawkins and Yacullo (1984) suggest that little or no directional benefit is expected in reverberant far-field listening conditions (those in which speaker to listener distance is significantly greater than the critical distance). Specific listening environments for which little or no directional benefit is expected include:

1. Listening when near the back of a moderate size theatre, church or concert hall, when the sound source of interest is located near the front;
2. Listening when not near the front of a large theatre or concert hall, when the sound source of interest is located near the front, and;
3. Listening at or beyond critical distance in a highly reverberant room such as a hard-walled classroom or restaurant.

In contrast with listening environments with large speaker to listener distance, significant directional benefit is expected, even when moderate reverberation is present in near-field listening conditions (those which speaker to listener distance is small) (Ricketts and Dhar, 1999; Ricketts, 2000b).

The number and placement of competing noise sources is also known to affect the measured directional benefit (Ricketts, 2000b; Valente *et al.,* 2000b). Ricketts (2000b) measured the directional benefit of 25 subjects with symmetrical, sloping, sensorineural hearing loss using a modified version of the HINT. Directional benefit was measured for four different configurations of competing noise source(s) in two different reverberant rooms. Three pairs of hearing aids representing three commercial models were selected for evaluation. The four noise source configurations included placement of competing noise speaker(s) as follows.

1. A single competing noise placed directly behind the listener (0/180);
2. Five competing noise speakers placed at 90, 135, 180, 225, and 270 degrees azimuth (5/B);
3. Five competing noise speakers placed at 30, 105, 180, 255, and 330 degrees (5/S); and,
4. Five competing noise speakers placed at 30, 105, 180, 255, and 330 degrees, with the speakers at 30 and 330 degrees turned to face perpendicular to the listener (m5/s).

The data from this experiment revealed that the configuration of the competing noise source(s)

significantly impacted both directional benefit and the rank order of benefit across hearing aid brands (from best to worst). That is, there was a statistically significant interaction between the benefit provided by the specific hearing aid models and the competing noise configuration. These results were interpreted as strong support that directional benefit assessed in the traditional test environment of a single noise source placed directly behind the listener could not be used to accurately predict directional benefit in more real-world, multinoise source environments. On average, directional benefit was significantly poorer when multiple noise sources, as opposed to a single competing noise, were used. Based on the results of this experiment, it seems likely that data collected using the commonly used 0/180 (a source speaker placed directly in front of the listener [0 degrees azimuth] and a single competing noise placed directly behind [180 degrees azimuth]) will overestimate the magnitude of directional benefit in many diffuse noisy environments. Data from this experiment were also used to argue that using the 0/180 speaker configuration in an attempt to rank order the directional benefit provided across directional hearing aid models in diffuse, real-world environments may lead to error. This certainly is worthy of note because the vast majority of studies that have examined directional benefit have used a single competing noise source placed directly behind the listener (Agnew and Block, 1997; Frank and Gooden, 1973; Gravel *et al.,* 1999; Hawkins and Yacullo, 1984; Hawkins and Yacullo, 1984; Lentz, 1972; Lurquin and Rafhay, 1996; Mueller and Johnson, 1979; Madison and Hawkins, 1983; Ricketts, 2000b; Valente *et al.,* 1995), although other positions have been suggested (Mueller and Sweetow, 1978; Sung *et al.,* 1975; Wouters *et al.,* 1999).

The problem with using a single loudspeaker location for competing noise is that any placement for which the competing noise source is placed at a null of the polar pattern will overestimate the benefit provided in a real-world environment with multiple noise sources at varying azimuths. Wouters and associates (1999) used an interesting method in which the loudspeaker was placed at an angle for which the attenuation provided by the directional hearing aid was equivalent to the DI of the instrument. In this way, the attenuation provided to the competing source should have been approximately equivalent to the average attenuation provided in a diffuse field (as quantified by the DI). While this method certainly is appropriate for some investigations, it could not be applied when comparing across directional hearing aid models that significantly varied in their polar attenuation patterns. Furthermore, it seems likely that reverberation might also interact with this design in novel ways in that the intensity of the reflected sound will be dependent on the angle of incidence in combination with the angular attenuation as defined by the instrument's spherical directional pattern. Finally, this method requires that both DI and directional pattern data of the test instrument are known.

In response to potential limitations of using a single competing noise source, several investigators have advocated the use of multiple competing noise sources to simulate real-word listening more accurately (Nielsen 1973; Preves *et al.,* 1999; Pumford *et al.,* 2000; Revit *et al.,* in review; Ricketts and Dhar, 1999; Ricketts, 2000b; Valente *et al.,* 2000a; Voss 1997). There is, however, been some discussion as to the most appropriate competing noise configuration for assessing directional benefit. One question that has been of recent interest is the use of correlated versus uncorrelated noise sources.[3] Several investigations have used multiple correlated-noise maskers in their investigations of directional benefit (eg, Pumford *et al.,* 2000; Valente *et al.,* 2000a; Voss, 1997); while others have argued that uncorrelated maskers are more appropriate (Preves *et al.,* 1999; Ricketts and Dhar, 1999; Ricketts, 2000b). The choice of correlated versus uncorrelated noise in speech-in-noise testing is an issue of practicality versus realism. From a practical perspective, correlated noise is simpler to implement, in that the same noise can be electrically split into any number of channels and delivered to the listener. The use of uncorrelated noise, while more logistically involved, is more representative of real-world listening situations. In real-world environments, such as a restaurant or party, not only does competing noise arrive at the listener's ears from multiple azimuths, but also, the noise

[3]Multiple noise sources can either be correlated, uncorrelated or psuedo- (partially) correlated at their source. In the case of uncorrelated competing noise, all loudspeakers output different signals. These signals may have the same long term spectral shape and intensity level, but at a minimum, are not in phase with each other. In the case of correlated competing noise, a single noise is routed to multiple loudspeakers.

originates from different speakers and sources. Thus, in such an environment, the competing noise is always uncorrelated at the source and may be partially correlated or uncorrelated at the listener's ear.

Recently, Gnewikow (2001) revealed that the choice of correlated versus uncorrelated competing signals did impact the speech recognition abilities of ten listeners with normal hearing, as measured by the HINT. The speech stimuli were delivered to subjects from a single loudspeaker at a 0-degree azimuth, with noise from four loudspeakers, equidistant from the listener's head and placed at 45, 135, 225, and 315 degrees. For correlated noise, the same cafeteria noise was presented simultaneously from all noise speakers. For the uncorrelated condition, four noncoherent sections of cafeteria noise were presented from the four loudspeakers. All testing was done in both an anechoic chamber and a moderately reverberant test room with average reverberation time (Rt 60) of 450 ms. All loudspeakers were placed at a distance of 1.25 m from the subjects' heads. Subject performance across listening conditions is shown in Figure 12. These data revealed a large significant difference in average subject HINT performance between the uncorrelated and correlated anechoic conditions (4.2 dB). A smaller, but still significant difference of 2 dB was noted between the uncorrelated and correlated conditions presented in the reverberant environment. Even though these data support a significant difference between speech recognition measured in the presence of correlated versus uncorrelated noise, further investigation is needed in order to determine if the correlation of the competing noise interacts with directional benefit as well as performance.

Unfortunately, there is significant variability across the environments that occur in the real world, and even identification and selection of an average environment can be quite challenging. One approach that has been used to measure real-world directional benefit is to actually quantify it in real-world environments (Killion *et al.,* 1998). These authors examined the directional benefit provided by a prototype ITE using a D-Mic in both indoor and outdoor environments. The nature of this experimental design, while providing excellent face validity, has not been standardized. This limits control over stimulus parameters (such as number and position of noise sources), making it difficult to compare these results with past and future investigations of other directional hearing aids. Consequently, the need for a test environment that is easily replicable and approximates real-world reverberation and competing noise source placement seems apparent.

In an effort to simplify the question of an appropriate test environment for evaluating directional benefit, Etymotic Research and Revitronix have jointly developed a multiple-loudspeaker listening system intended to simulate life-like adverse listening conditions (Revit *et al.,* in review). This system, referred to as R-Space, was developed to provide simulated environments that sound real and allow hearing aids and the hearing mechanism to perform as they do in the real world. A smaller, less expensive system is also being developed by these investigators for use by clinicians. The R-Space system uses an array of eight loudspeakers and a multichannel audio recording with eight discrete signals to simulate realistic acoustic environments at the listening position in the center of the array. Initial investigations with this device in both anechoic and an acoustically treated conference room revealed that listeners judged the R-Space environment to provide a similar acoustic experience to that recorded live through the KEMAR. The similarity between the R-Space and the live recording was

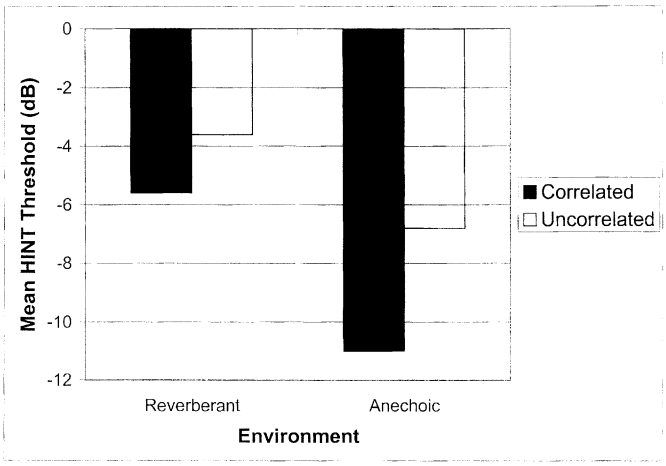

Figure 12. Average speech recognition performance as measured by the HINT of ten normal hearing listeners for both correlated and uncorrelated competing noise stimuli. Data were collected both in an anechoic chamber and a moderately reverberant test room (adapted from Gnewikow, 2001, with permission).

slightly better when presented in the anechoic environment. While additional evaluation is obviously needed, the R-Space system appears to have potential for use as a viable research and/or clinical method for evaluation of real-world performance of advanced hearing instruments in noisy environments.

3. Clinical Tips and Hints

Nearly all speech in noise tests include speech material on one channel and competing noise stimuli on the other channel. Consequently, adapting nearly any speech in noise test to the assessment of directional benefit simply requires separation of the test and competing signals through the use of two (or more) calibrated loudspeakers. All of the experimental factors described above can impact the measured directional benefit including reverberation, number and placement of competing noise sources, etc. Consequently, designing a clinical test environment which emulates those found in the real world in order to better approximate actual directional benefit is challenging to say the least. However, if the goal of measuring directional benefit in the clinic is to ascertain whether a particular patient with a specific hearing aid is performing similarly or differently from the average patient in the same environment, or if the goal is to obtain a measure of directional benefit for counseling purposes, the use of the commonly available 0/180 test configuration in a sound-treated room seems quite appropriate. Moving the testing to an available room with more reverberation could also be considered if a slightly more accurate picture of benefit in at least one real-world environment is desired.

While it has been my experience that most clinics are more likely to possess fixed rather than variable SNR tests, variable SNR methods have the added advantage of being faster to administer. For example, the HINT test can be administered in approximately three to six minutes per condition depending on whether it is desirable to use one, or two, 10-sentence blocks, to improve reliability. A total of 6 minutes appears to be very time efficient; however, Etymotic Research (2001) recently developed the QuickSIN that only requires 1 minute of test time per condition. The QuickSIN is made up of 12 lists of six sentences with five key words per sentence presented in four-talker babble noise. The sentences are presented at pre-recorded signal-to-noise ratios, which decrease in 5dB steps from 25 dB (very easy) to 0 dB (extremely difficult).

One concern with using a variable SNR methodology relates to explaining this data to patients. Simply stating that there is 3.5 dB of directional benefit is unacceptable for the average patient. Fortunately, there are some data relating SNR improvement to improvement in percent correct score. For instance, Soli and Nilsson (1994) reported that 1 dB SNR improvement on the HINT corresponded to an 8.5% improvement in speech recognition scores. In addition, Killion and coworkers (1998) have addressed this issue using IEEE sentences spoken by live speakers. More data are needed in this area to address a wide range of speech materials and listening conditions. While not necessarily scientifically accurate, I certainly think that using a rule of thumb, such as 8% per dB of change, is appropriate for the purposes of counseling patients.

In contrast to the variable SNR methods, the fixed SNR methods have the advantage of providing straightforward percent correct information about improvement that is easy to explain to our patients. Choosing the appropriate test SNR, however, can be difficult. While the most defensible position is to select a SNR that corresponds to that which a listener will most often encounter, real-world SNRs vary greatly, and the clinician does not typically know the SNRs experienced by their patients. The work of Pearsons and associates (1976) provides the reader with some estimates of average real-world SNR conditions. Their research illustrated that in face-to-face communication, talkers do not raise the intensity of their voice at the same rate that background noise increases. This study revealed that SNR decreased from +6 dB when background noise levels were 55 dB SPL to –1 dB when background noise levels were 75 dB SPL. These findings could be used as real-world guidelines for establishing a fixed SNR speech testing protocol for the laboratory or clinic. We often use +2 to +4 dB in our laboratory, although these SNRs occasionally need to be modified for certain patients.

Subjective Evaluation of Directional Benefit

In comparison to the large number of investigations that have examined directional hearing aids

through objective methods, relatively few studies have examined subjective directional benefit or satisfaction. Nielsen and colleagues (Nielsen 1973; Nielsen and Ludvigsen, 1978) investigated user preference for directional versus omnidirectional hearing aids and found that subjects preferred directional hearing aids in a sound-treated room and a cafeteria environment, but showed no preferences in other everyday communication environments. Mueller and associates (1983) studied the preferences of a group of listeners with hearing loss. The subjects rated omnidirectional and directional hearing aids as strongly superior, superior, mildly superior or no preference in different listening conditions encountered while wearing the hearing aids during a trial period. A majority of subjects reported no preference; however, when a preference was present, the subjects preferred the directional aid. While these studies indicate either no preference or a directional preference, the findings are limited in that the subjective measures used are not standardized and the reliability of these measures is not known. Furthermore, these studies were conducted with first generation directional hearing aids and the results may not be able to be generalized to currently available directional hearing aids.

Since the development of second-generation directional hearing aids, there have been relatively few published studies that have systematically examined subjective ratings of directional benefit using standardized methodology (Preves *et al.,* 1999; Valente *et al.,* 1995; Walden *et al.,* 2000). Valente and associates (1995) used the Profile of Hearing Aid Benefit (PHAB, Cox and Rivera, 1992) and the Abbreviated Profile of Hearing Aid Benefit (APHAB , Cox and Alexander, 1995) to determine if subjects with hearing loss received significantly more benefit with directional hearing aids than the average user of linear amplification. These data were collected at two different sites. The authors reported better PHAB scores for the directional hearing aids on the background noise (BN) and reduced cues (RC) subscales at one site, and better APHAB scores on the BN and aversiveness (AV) subscales at the other site. Additionally, the authors reported a general preference for the directional hearing aids in comparison to the subjects' current aids at one of the two experimental sites. These data provide some support for the perceived benefit provided by directional hearing aids over their omnidirectional counterparts. The magnitude of this preference remains unclear, however, since the directional hearing aids were compared to the subjects' own hearing aids and the amount of benefit due strictly to the directional component cannot be independently assessed.

Preves and associates (1999) examined subjective differences across the directional and omnidirectional modes for subjects fit bilaterally with a single model of ITE hearing aid using the APHAB, paired comparison judgments, and interview data. Results indicated that the equalized directional mode was ranked significantly better on the reverberation (RV) and background noise (BN) subscales of the APHAB when compared to the omnidirectional fitting. In addition, when asked to choose a single mode that they would be required to listen with all of the time, six of the ten subjects chose the directional mode over omnidirectional. Finally, paired comparison testing revealed that the majority of subjects preferred the equalized directional mode for clarity, quality, and reduced annoyance, when listening in noise over the omnidirectional mode.

Most recently Walden and coworkers (2000) examined the performance of 40 adults with hearing loss fit with: 1) low-threshold compression DSP instruments; 2) linear hearing aids with input compression limiting (AGO-I), and 3) two-channel analog wide dynamic range compression (WDRC) instruments. The DSP instruments were evaluated with an omnidirectional microphone, dual-microphone directionality, and a noise reduction circuit in combination with dual-microphone directionality. Each of the hearing aid conditions was assessed following a two week trial using the Connected Speech Test (CST), the Profile of Hearing Aid Benefit (PHAB), and subjective ratings of speech understanding, listening comfort, and sound quality. Significant directional benefit, as measured by the CST, was reported. Concomitant directional benefit in everyday listening situations, as measured by the PHAB, however, was not found. These results are in sharp contrast of those reported by Valente and associates (1999) and Preves and associates (1999), and are especially surprising because all three studies reported large and significant directional benefit measured using objective measures. Walden and coworkers (2000) have suggested several factors that may contribute to lack of subjective benefit observed, even in the presence of objective benefit. These factors include the possibility that objective laboratory measures may over-

estimate directional benefit in the real world due to environmental factors (reverberation, number of competing noise sources, etc.), the fact that the PHAB was not independently administered for each hearing aid condition, the lack of appropriate acclimatization, and a possible lack of real-world experience of some subjects with the difficult SNR conditions of the test environment.

Hearing aid satisfaction is another subjective measure that can be assessed with directional hearing aids. A recent survey by Kochkin (2000a) indicated that 16.2% of individuals who own hearing aids never wear them, and 62.3% of those people cite dissatisfaction when listening in noise as the cause for their failure to use the aids. Interestingly, previous data from Kochkin (1996), which examined whether advanced hearing aid features such as programmability, multi-memory, directionality, and so on impacted listeners' satisfaction with hearing aids revealed that the hearing aid receiving the highest satisfaction rating was a dual-microphone BTE. That same year Kuk (1996b) demonstrated improved hearing aid satisfaction in a group of multiple microphone BTE directional hearing aid users relative to the average hearing aid user. Similarly, Schuchman and associates (1999) showed better satisfaction with the use of a directional ITE hearing aid than is reported for the average hearing aid wearer. More recently, Kochkin (2000b) reported MarketTrak survey results that revealed 78% of hearing aid wearers fit with directional microphone digital instruments were satisfied with their hearing aids. This was substantially greater than the 64% that were satisfied users of digital hearing aids with omnidirectional microphones. This 14% increase in overall satisfaction is rather noteworthy when compared to results revealing that users of digital hearing aids with omnidirectional microphones were only 3% more satisfied than the 61% satisfaction reported by all hearing-aid wearers as a group.

The majority of these data demonstrate subjective directional benefit. These results are not nearly as convincing as objective directional benefit data. While these differences may be explainable by differences in test instruments and experimental methodology, they certainly highlight the importance of measuring not only objective, but also subjective directional benefit. Jerger (2000) has further suggested that identifying the exact characteristics of everyday listening environments in which directional hearing aids will be helpful to the user may also be necessary to improve our understanding of the benefits of directional amplification.

Summary of the Verification and Validation of Directional Hearing Aids

There certainly are a number of issues related to the testing of directional hearing aids that must be considered. Electroacoustic methods include the common laboratory based methods of directional patterns and DI calculations. There are several methods for calculating frequency specific DI, and while most methods yield similar values, differences do exist. These differences make it difficult to compare hearing aid directivity across instruments based on measurements made in different laboratories. Multifrequency DI calculations are sometimes simplified to a single value (AI-DI) using an Articulation Index weighted average. The AI-DI has been advocated as a simple way to compare directivity across different instruments in terms of their potential impact on speech recognition in noise.

In contrast to laboratory methods, FBR represents a quick and simple method to examine directivity in the clinic using commonly available probe microphone equipment. Due to possible interactions between nulls in the directional pattern and loudspeaker placement, however, the use of either FBR or behavioral measures with a single competing noise source for comparison across hearing aid models is not advised.

Behavioral measures of directional hearing aids include quantification of performance and directional benefit. Directional benefit can be measured using speech recognition testing at variable, adaptive, or fixed SNRs. SNR varies greatly in real-world environments, however, data suggest that difficult listening environments have SNRs ranging from approximately +6 to –1 dB. Increasing the number of competing noise sources, reverberation, room size and distance from listener-to-talker will all negatively impact directional benefit. These factors may combine to have a significant interactive effect on the directional benefit measured across various hearing aid models. Traditional 0/180 directional benefit measures continue to be advocated, however, when the goal of testing is to determine general benefit compared to average patients or for counseling purposes. It is suggested that it is impor-

tant to examine directional benefit not only objectively, but also using subjective measures to obtain a more accurate picture of how much directional amplification may benefit a user in everyday listening situations.

Fitting Factors Impacting Directional Benefit and Directivity

Our interest in directional hearing aids has resulted in the testing of more than 150 adult patients with these devices in our laboratory. These patients have been evaluated using many different hearing aid models; however, nearly all have listened in at least one environment that was intended to simulate a difficult, near field, real-world listening condition. That is, levels of reverberation were at least moderate (> 400 ms), and at least four, uncorrelated competing noise sources were used. These data reveal that the average listener fit with directional hearing aids in these difficult listening environments receives approximately 3 to 4 dB of directional benefit as measured by a variable SNR test, and 20% to 35% as measured using a fixed SNR. In addition to these data, significant directional benefit has also been reported in children[4] (Gravel *et al.,* 1999; Hawkins, 1984; Kuk *et al.,* 1999). While it certainly appears that there is a significant advantage for directional hearing aids over their omnidirectional counterparts when listening in noise, it is important for us to consider fitting factors that may impact the magnitude of directional benefit. From these data we can better determine when directional hearing aids may be appropriate (or not). Knowledge of the impact of fitting factors on directional benefit and directivity is necessary in order to best decide how to weigh changes in directivity against other fitting decisions. For example, if a particular patient required a fitting parameter, or combination of parameters, that were known to eliminate the difference in directivity between directional and omnidirectional hearing aids, it would be foolish to also order a directional microphone.

[4]It is my opinion that directional hearing aids are not appropriate for very young children, but may be appropriate for children who are old enough to orient their heads toward the sound source of interest.

Examining the impact of all potential fitting factors on directional benefit would be a momentous task indeed. Fortunately, data suggest that the magnitude of directional benefit is, at least relatively, predictable from DI (Ricketts and Dittberner, 2002). Consequently, it is argued that it is appropriate to examine the impact of fitting factors on DI, and then draw conclusions concerning directional benefit from these data. Before doing this, however, it is useful to examine the directivity of modern directional hearing aids. As noted previously, both the measurement method and calculation method can impact directivity results. Consequently, in an attempt to provide an accurate picture of the relative directivity of a large number of instruments, the DI of several instruments all measured in our laboratory using the same methodology are provided in the following.

Specifically, the frequency specific range of DI across ten models of commercial directional hearing aids that are currently on the market are shown in Figure 13 (data are from Ricketts, 2000a; Ricketts *et al.,* 2001; Ricketts, unpublished data). These hearing aids were places in categories based on hearing aid style and design. First, it is important to note that these DI values are generally much greater than those reported for first generation directional hearing aids. In

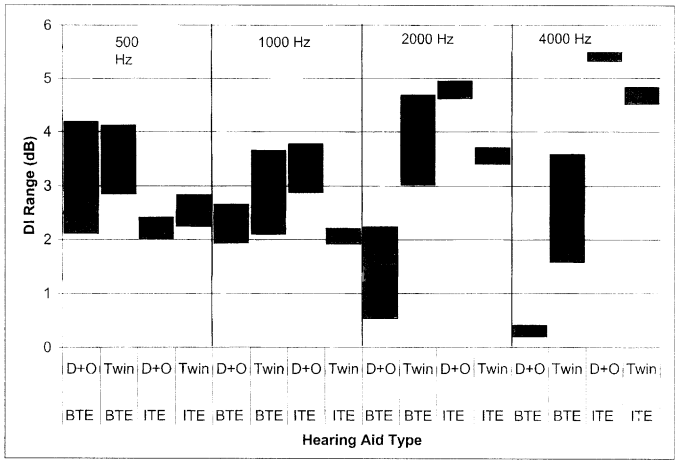

Figure 13. The range of calculated DI values across ten models of commercially available directional hearing aids. All DI calculations were made from single plane directional pattern data. The individual hearing aids are differentiated by style (BTE versus ITE) and design (directional + omni, "D+O" and twin microphone, twin).

fact seven of the ten instruments had DI values that exceeded the highest of those reported for the first generation directional instruments by at least 1 dB (Killion *et al.,* 1998). Another obvious conclusion from these data is that not all modern directional instruments reveal similar directivity, either on average, or at specific frequencies. For example, DI values range from 2 to more than 4 dB at 500 Hz and from 0.2 to 5.5 dB at 4000 Hz. It is further evident that it is not possible to predict the magnitude of DI values based on overt design differences. For example, instruments using the directional + omni design revealed AI-DI values that were the highest measured for ITE instruments (4.1 dB), and the lowest measured for BTE instruments (1.3 to 2.1 dB). In addition, the directivity of all ITE instruments was higher than the BTE instruments on average; however, the AI-DI value of the twin microphone BTE 1 was greater than both of the twin microphone ITE models evaluated. These data support the conclusion that the specific microphone design, rather than the general design principles have the greatest impact on the achieved directivity. Consequently, relying on general assumptions such as dual microphone designs are superior to directional + omni, or directivity is predictable based on hearing aid style can obviously lead to error.

While it is evident that the directivity of modern directional hearing aids varies across models, there is also evidence that there is significant variance within models. Ricketts (2000a) reported the within model AI-DI variability of three commercial, directional, BTE hearing models. AI-DI values were calculated for three hearing aids of each model. The results of this study indicated AI-DI values varied by between approximately 0.5 and 2 dB depending on the model. This variation was significantly greater than the 0.1 dB test-retest reliability obtained using single instruments in this same investigation. It is assumed that variability in the directivity provided by the same hearing aid model reflects quality control problems since all models were evaluated on arrival (before fitting them to patients). In the case of twin microphone instruments, it is assumed that this variability was most likely due to differences in how closely the microphones were matched, problems with the electronic components responsible for internal delay and/or signal summation, or measurement error. In the case of the directional + omni instruments it is assumed that the variability must result from physical differences in the acoustic phase shifting pathways, or measurement error. If the assumption is made that similar measurement error was present across all instruments, these data support that the within model directivity of some models is significantly more variable than others.

To date, investigators have reported data related to the effect of a number of fitting factors on directivity (eg, Beck, 1983; Mueller and Wesselkamp, 1999; Ricketts and Mueller, 1999; Ricketts, 2000a). In the following the impact of microphone port orientation, venting, compression, low-frequency gain equalization, monaural versus binaural fitting, head turn and hearing aid style on directional benefit will be considered. The use of adaptive and automatic directional microphones will also be considered.

Microphone Port Orientation

Many manufacturers have stressed the importance of adjusting the microphone ports of directional hearing aids so that they are in the horizontal plane. This certainly appears to be valid since microphone ports in the horizontal plane are necessary to maintain the desired external delay. For example, if the sound source of interest is directly in front of the listener, and the microphone ports are placed perpendicular to the horizontal plane, the resulting external delay will be reduced to zero. The question is, however, how close to the horizontal plane must the microphone ports be before directivity is affected? Deviation from the horizontal plane can occur with ITE hearing aid fittings if the ear impression is not appropriately marked, or if the manufacturer is not able to orient the microphone ports appropriately due to the constraints of an individual's ear geometry. Port deviation from the horizontal plane can occur in BTE hearing aids because microphone port angle is impacted by the length of the earmold tubing.

Ricketts (2000a) reported that placement of the case of specific BTE hearing aids so that it is in contact with the back of the pinna over the entire length of the hearing aid, resulted in a microphone port orientation that deviates from the horizontal plane by as much as 24 degrees. The reduction in frequency specific DI across port orientations for ITE and BTE hearing aids reported in two previous investigations are reproduced in Figure 14 (adapted from Mueller and Wesselkamp, 1999; Ricketts, 2000a). These authors highlighted three conclusions:

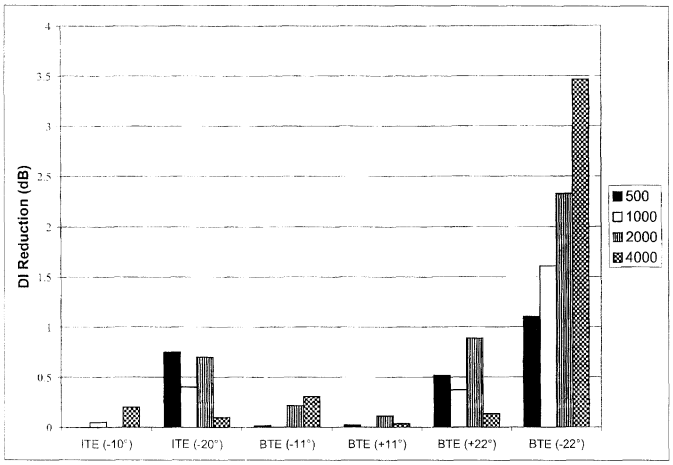

Figure 14. Frequency specific reduction in DI (relative to an optimal [horizontal] orientation) averaged across three BTE or single ITE hearing aid model. Nonoptimal hearing aid angles included +11, –11, +22, –22 degrees for the BTE instruments and –10 and –20 degrees for the ITE instrument. Angles for which the rear microphone port was lower (relative to a horizontal plane) than the front were considered negative, and angles for which the rear opening was higher were considered positive. These data represent the average DI values across at least two directional hearing aids of each model. (Adapted from Mueller and Wesselkamp, 1999; Ricketts, 2000a).

1. Small deviations from the horizontal plane (approximately ±10 degrees) do not significantly impact DI for either BTE or ITE style instruments.
2. Deviations of approximately ±20 degrees from horizontal significantly reduce low-frequency directivity; however, the DI even in the worst condition was still significantly greater than that obtained by omnidirectional fittings.
3. In the case of BTE instruments, the reduction in directivity due to a non-optimal microphone port orientation can be further exacerbated by "sound shadow" due to close proximity of the microphone ports to the helix of the pinna. This additional reduction is quite evident when comparing the directivity for the –22-degree and +22-degree conditions (BTE only) shown in Figure 14. Positive orientations are used to denote conditions for which the front microphone is lower than the rear, while the opposite is assumed for negative orientations (relative to the horizontal plane).

These findings suggest that audiologists must be especially careful when fitting directional BTE instruments to insure that instruments are not fit tightly against the pinna, if maximum directivity is the goal. However, it might not always be possible to obtain a perfectly horizontal microphone port orientation on specific patients, even if care and precision are used in all phases of the hearing aid selection and fitting. In the case of ITE instruments, this failure may result from the previously mentioned inability to orient the microphone ports appropriately during the manufacturing process due an individual's ear geometry. When this problem arises it may be necessary to send the hearing aid back for remake until a satisfactory compromise is reached. In the case of a BTE, the patient simply may not tolerate the positioning of the BTE that is necessary to provide the horizontal orientation. A number of patients in the laboratory that are past wearers of BTE instruments complain of discomfort when the instrument is placed at its optimal angle. In other cases the optimal angle results in less than optimal retention (the hearing aid is floppy). Consequently, it is fortunate that data suggest that small deviations from a horizontal microphone port orientation do not significantly impact directivity. In some cases, however, it may be necessary to reach some compromise between listener comfort and directivity, as it is quite obvious that most listeners with hearing loss will not wear hearing aids that are uncomfortable. Clinical assessment of directivity for individual fittings seems advisable, given the potential trade-off between optimal microphone port orientations for directivity versus comfort.

Venting

Venting also has the potential for affecting the directivity of a hearing aid. If a vent is present in the earmold, low-frequency sounds (< approximately 1000 Hz), originating from behind the hearing aid, may simply pass through the vent without attenuation, and may approach the intensity level of the amplified signal originating from in front, significantly reducing directivity. Studies have shown that directivity is generally reduced with increasing vent size (Beck, 1983; Mueller and Wesselkamp, 1999; Ricketts, 2000a). Consequently, it appears that if maximum directivity of these hearing aids is the goal, no venting should be used. Frequency specific and average

AI-DI values across four earmold venting configurations are shown in Figure 15 (derived from Ricketts, 2000a). The earmold configurations include three with a full shell, acrylic earmold and #13 tubing (1 mm venting, 2 mm venting, no venting); and a single loose fitting open earmold condition. Prior to testing, the hearing aids were programmed with linear gain and the gain was adjusted until no feedback was present with the open earmold coupling. As can be seen the reduction of directivity with increasing venting is concentrated in the low frequencies. These data revealed that the effect was only significant at the lowest test frequencies (500 and 1000 Hz) and no decreases were noted in the high frequencies. The decrease in AI-DI was relatively small (approximately 0.4 dB) as vent size was sequentially increased from closed to 1 mm, and then again from 1 mm to 2 mm. In contrast, AI-DI values were reduced 0.8 dB when changing from a 2 mm vent to an open earmold. It should be noted that the open earmold condition still resulted in average AI-DI values approximately 4 dB greater than measured for the omnidirectional condition. It appears then, that listeners will receive significant directivity from many modern directional hearing aids, regardless of venting. Therefore, unless evidence to the contrary emerges, it appears that recommendations for directional hearing aid use are viable regardless of venting needs.

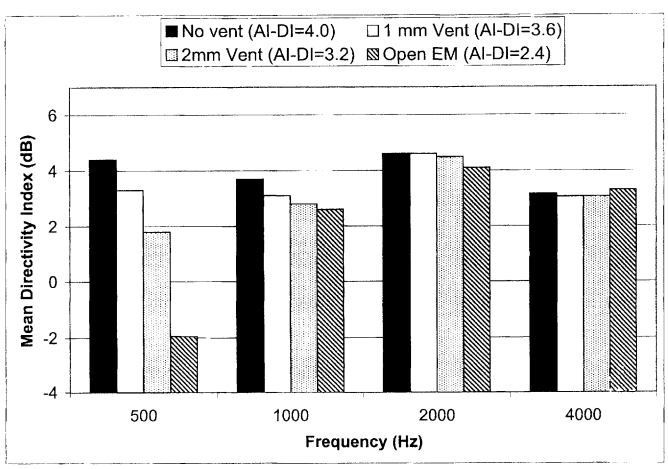

Figure 15. The average frequency specific DI and AI-DI for eight hearing aids (four each of two different models) calculated for closed earmold (closed), 1 mm, 2 mm, and open earmold (OM) venting conditions (data from Ricketts, 2000a).

Caution may be warranted, however, when using an open earmold with some modern directional hearing aids that exhibit directivity mainly in the low frequencies. While somewhat rare, such directional hearing aids do exist (Ricketts, 2000a). It is evident that applying a large vent to a directional hearing aid that has positive directivity only in the low frequencies (when compared to an omnidirectional instrument) is likely to result in a cancellation of the directional effect and may result in no directional benefit. It is therefore important for audiologists fitting directional hearing aids to know the frequency specific directivity characteristics of the instruments that are being fit. It also seems prudent to quantify the impact of venting on directivity in the clinic. Quantification of the relative effect of factors such as venting and microphone opening azimuth on directivity can be easily completed in the clinic using FBR measurements.

Compression

Directional hearing aids vary in their use of dynamic amplitude processing from linear through multi-channel, low-threshold compression. The potential for interaction between low-threshold compression and directivity has led some investigators to explore this topic (Mueller and Wesselkamp, 1999; Ricketts, 2000a). An interaction between low-threshold compression and directivity is possible because the purpose of a directional microphone is to change the intensity of sounds based on their angle of arrival, and low-threshold compression hearing aids will vary the amount of gain applied to signals based on the intensity level of the signal at the output of the microphone. That is, a compression hearing aid will generally provide more gain for low intensity sounds than for high intensity sounds. Consequently, low-threshold compression hearing aids will provide more gain (less compression) for signals arriving from azimuths for which amplitude is reduced by the directional microphone (primarily the rear hemisphere) than for those signals arriving from azimuths for which there is little or no amplitude reduction (primarily the front hemisphere). Not surprisingly, this interaction results in a reduction in the magnitude of traditionally, single source at a time, measured directivity (ie, directional patterns and FBR) for low-threshold compression hearing aids in comparison to their linear counterparts (Mueller

and Wesselkamp, 1999; Ricketts, 2000c). By definition, the amount of gain provided by a linear hearing aid will be constant regardless of the output of the hearing aid microphone until a saturation level is reached. Not surprisingly, the presence or absence of compression does not effect diffuse field DI measures because, as in the real world, the multiple sources are on at the same time.

The potential impact of compression on directional benefit and performance has also recently been investigated on 47 listeners bilaterally fit with five different hearing aid models (one BTE and four ITEs) in a listening environment intended to simulate a noisy restaurant (Ricketts *et al.,* 2001). Speech recognition performance was measured using the connected speech test (CST) presented at a fixed SNR (+1 dB or +4 dB depending on the test site) and the HINT presented using an adaptive SNR. Four of the five hearing aids were capable of both linear and low-threshold compression (WDRC) processing, while the fifth model was a linear peak clipping ITE. Results revealed that compression versus linear processing had no impact on the magnitude of listeners' performance or directional benefit. The directional benefit results were not surprising given directivity data and the fact that the signal of interest and the competing signals were present at the same time. The lack of significant performance differences between linear and compression fittings was also not that surprising, given that gain for input levels approximating those used in this study was matched across linear and compression conditions. The combination of matched gain and relatively high presentation levels likely resulted in similar audibility of speech across the linear and compression conditions. A number of investigations have reported similar speech recognition performance across linear and compression hearing aid fittings when no audibility advantage is provided by the compression condition (ie, Dillon, 1996; Souza and Turner, 1998).

Low Frequency Gain Equalization

Despite the potential for an increase in internal microphone noise, some compensation for the change in frequency response that results from activating a directional microphone is sometimes recommended to offset the potential loss of audibility (Christensen, 2000; Ricketts, 2000b; Wolf *et al.,* 1999). This compensation is commonly referred to as a directional-equalized frequency response. Currently, several hearing aid manufacturers offer fitting options that compensate, to some degree, for the change in frequency response that occurs with the use of directional microphones. Frequency response compensation can be accomplished at either the amplifier, or microphone preamplifier stage, and can be implemented in either twin microphone or directional + omni designs.

How, and when, to best apply frequency response compensation is still questionable. The magnitude of the negative impact of low-frequency roll-off on audibility for speech recognition would appear to be somewhat dependent on the configuration of hearing loss. For listeners with significant low-frequency hearing loss, there is little doubt that even with the greatest port spacing (resulting in approximately 15 dB of attenuation at 500 Hz) the reduction in low-frequency speech output could significantly reduce audibility, resulting in reduced speech recognition. For listeners with little or no hearing loss in the low frequencies, however, it is likely that the fitting will include venting so that the primary, low-frequency, sound reception pathway bypasses the hearing aid altogether. Consequently, switching to directional mode may not have as great of an effect on audibility for those listeners with normal or near normal low-frequency hearing.

Data from Preves and associates (1999) support the hypothesis that listeners with little low-frequency hearing loss may be relatively unaffected by the reduction of low-frequency output that occurs when switching from omnidirectional to directional mode. In this study, the speech recognition of ten subjects, eight of which had normal hearing through 1000 Hz, was evaluated across two ITE directional hearing aid conditions. These conditions included a standard directional mode (ie, unequalized), and a directional-equalized mode. The results of this study revealed no significant differences in speech recognition abilities of subjects fit with the two microphone conditions.

Currently, we are examining the question of frequency equalization in directional mode using three groups of ten adult listeners with hearing-impairment (a total of 30). These three groups were differentiated by degree of low frequency hearing loss. All groups exhibited hearing thresholds at 3000 Hz between 35 and 75 dB HL. Group 1 exhibited hearing thresholds at 500 Hz of less than or equal to 25 dB HL. Group 2 exhibited hearing thresholds at 500 Hz of between 30 and

45 dB HL. Group 3 exhibited hearing thresholds at 500 of 50 dB HL or greater. An appropriate SAV vent size was selected for all subjects. Speech intelligibility and sound quality in quiet and noise were measured for adult subjects fitted bilaterally with four hearing aid conditions. These hearing aid conditions consisted of a commercial hearing aid (Bernafon Smile) set to directional mode with: 1) no gain compensation, 2) full gain compensation, 3) hearing loss–dependent gain compensation, and 4) omnidirectional mode. The hearing loss dependent gain compensation fitting is part of the Bernafon fitting software and assigns increasingly more low-frequency equalization with increasing low-frequency hearing loss. Both the omnidirectional and directional with full gain compensation conditions were programmed using the National Acoustics Laboratory Nonlinear (NAL-NL1) procedure (Dillon, 1999) and verified using probe microphone measures (Frye 6500-CX). Testing was performed using a single speech source and five uncorrelated competing noise sources, in a simulated restaurant environment.

Partial directional benefit data as measured by the Connected Speech Test at a + 2 dB SNR (8 listeners in each group) are shown in Figure 16. These results reveal no significant differences in directional benefit for subjects with mild-to-moderate, low-frequency hearing loss (groups 1 and 2) across the three gain configurations. A significant difference in directional benefit was measured, however, for the group with the poorest low frequency thresholds (group 3). Specifically, failure to equalize the directional frequency response resulted in significantly less directional benefit for this group.

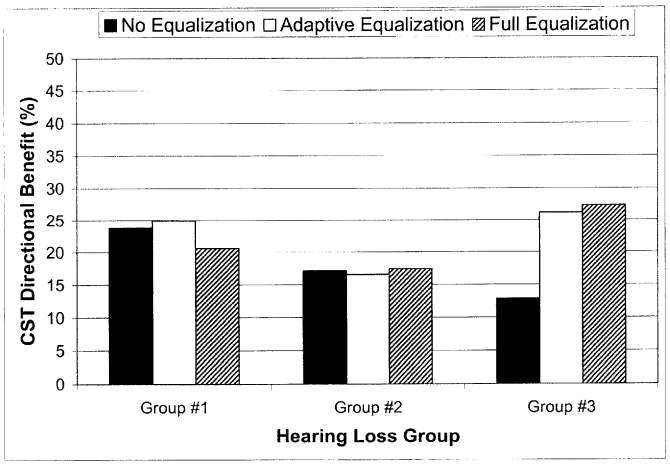

Figure 16. The directional benefit measured by the connected speech test for three groups of 8 listeners (24 total) differentiated by degree of low-frequency hearing loss. Group 1 exhibited 500 Hz hearing thresholds of less than or equal to 25 dB HL. Group 2 exhibited 500 Hz hearing thresholds of between 30 and 45 dB HL. Group 3 exhibited 500 Hz hearing thresholds of 50 dB HL or greater.

Based on these data, the following general recommendations are offered. If significant low-frequency hearing loss is present, an equalized response is recommended to insure audibility. For patients with normal, or near-normal, low-frequency hearing thresholds, no equalization appears to be necessary to provide optimal speech recognition. In addition, using a frequency equalized response when low-frequency thresholds are near normal may reduce sound quality if the level of microphone noise becomes audible.

Monaural Versus Binaural Fitting of Directional Hearing Aids

The binaural advantage for speech recognition in reverberant environments and a background of noise is well documented for both listeners with and without hearing loss (Arsenault and Punch, 1999; Bronkhorst and Plomp, 1989; Bronkhorst and Plomp, 1992; Byrne, 1981; Carhart, 1965; Moncur and Dirks, 1967; Peissig and Kollmeier, 1997; Saberi *et al.,* 1991; Yost, 1997). Three studies have examined the binaural advantage present for listeners fit with directional and omnidirectional hearing aids (eg, Hawkins and Yacullo, 1984; Nabelek and Mason, 1981; Ricketts, 2000c). Aided binaural advantages that range from 1.5 to 3.4 dB have been reported depending on conditions. One interesting finding is that the magnitude of the binaural advantage does not appear to be significantly different for directional versus omnidirectional amplification. That is, directional hearing aids have no impact, either positive or negative, on the measured binaural advantage.

Data from Ricketts (2000c), shown in Figure 17, reveal the magnitude of speech recognition performance in directional and omnidirectional modes, as measured by the HINT, across monaural and binaural fittings. All listeners exhibited symmetrical hearing losses, and the unaided ear remained unoccluded in the monaural condition. These data show that while a binaural advantage is clearly evident, the magnitude of directional

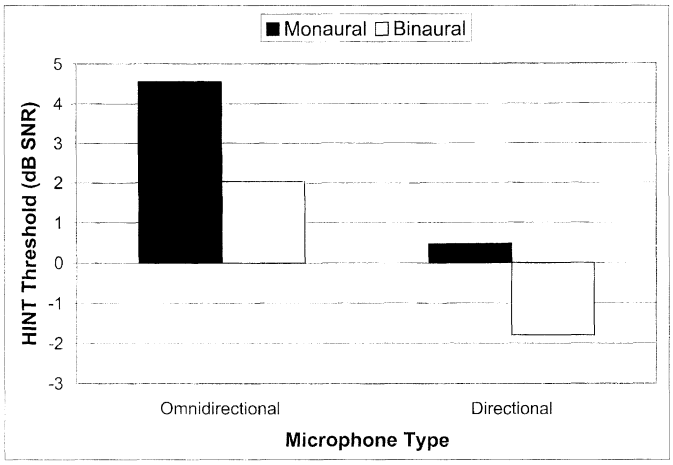

Figure 17. The avearage speech recognition performance of listeners fit monaurally and binaurally in both directional and omnidirectional modes (as measured by the HINT).

benefit was not impacted by the number of hearing aids used. Stated another way, the same amount of directional benefit was present for both monaural and binaural fittings. These data are viewed as support for the use of directional hearing aids for either binaural or monaural fittings.

Head and Body Turn

Previously it was suggested that the DI of a hearing aid approximates the effective SNR of a listening environment in which the signal of interest originates directly in front of the listener in the presence of a diffuse noise field. However, the majority of hearing aids, including directional hearing aids, are not most sensitive directly on-axis, but rather, show the greatest sensitivity at angles ranging from approximately 30 to 180 degrees (Beck, 1983; Fortune, 1997; Mueller and Hawkins, 1990; Ricketts, 2000a). Consequently, it is not surprising that it has been demonstrated that DI values could be significantly affected by changing the assumed angle of sound incidence (Fortune, 1997).

Recent data have shown that changes in DI resulting from varying the angle of incidence of the sound source of interest are also reflected in behavioral data (Ricketts, 2000c). Specifically, results indicated that listeners performed significantly better (approximately 2 dB as measured by the HINT) with a 30-degree head angle than when directly facing the primary speaker in an *auditory only* condition (in good agreement with DI predictions). Data from Henry and Ricketts (2001) revealed that for *auditory + visual* input, performance for a 20-degree angle was better than when the listener faced the stimulus loudspeaker. Listener performance, however, did not continue to improve with further increases in head angle. Instead, performance decreased between 20 and 40 degrees. These investigations were based on the assumption that listeners wearing hearing aids may be able to increase speech recognition in noisy environments after being counseled to rotate their heads so that the primary signal originates from the angle corresponding to the hearing aid's greatest sensitivity. Unfortunately, it is not yet known whether listeners already generally turn their heads in an attempt to improve speech intelligibility in noise, in the absence of instruction.

There are also limited data concerning the magnitude of the negative impact that a nonoptimal listening angle may have on speech understanding. Lee and associates (1998) reported a 20% reduction in speech recognition for speech presented directly behind adults fit with a directional hearing aid. Only a single angle (180 degrees) was investigated, however, and only a single noise source was used. In addition, the directional hearing aid chosen is known to have directivity which, unlike the majority of hearing aids on the market, is concentrated only in the low frequencies.

Hearing Aid Style

As noted previously, the directivity provided by directional BTE and ITE instruments are fairly similar. As a result similar performance in noise is expected from listeners fit with these two styles. A number of clinicians and authors have remarked that it appears that their patients get more directional benefit from BTE style instruments. If true, why would this be the case? Somewhat surprisingly the answer becomes apparent when the performance in noise of listeners fit with different styles of *omnidirectional* hearing aids are examined. Specifically, while it has been shown that the directivity of an omnidirectional ITE hearing aid is approximately equal to that of the open ear; directivity of BTE instruments, especially in the high frequencies, is significantly poorer (Beck, 1983; Ricketts 2000a, Ricketts *et al.,* 2001). The reduced high frequency directivity for BTE fittings

in comparison to the open ear are not surprising, in that the microphone placement of the BTE instrument does not take advantage of the natural sound shadow provided by the pinna that is present with a deeper microphone placement (Beck, 1983). Not surprisingly, behavioral evaluation of listeners wearing *omnidirectional BTE* instruments in noisy environments also generally reveals reduced performance when compared to their ITE counterparts (Pumford *et al.,* 2000; Ricketts *et al.,* 2001).

Fortune (1997) has reported that, as omnidirectional microphone depth is increased from the position of an ITE microphone to a completely in the canal (CIC) placement, high frequency DI is enhanced. The magnitude, and/or frequency range of the increased directivity, however, is not generally comparable to that provided by a directional microphone. Specifically, differences in DI values between ITE and CIC hearing aid fittings were only significant at frequencies of 4000 Hz and above.

Given differences in word recognition performance between different styles of omnidirectional instruments, it is not surprising that the directional benefit provided to listeners fit with ITE style instruments is significantly less than that reported for BTEs, even though subjects' performance when fit with these instruments in directional mode may be very similar (Pumford *et al.,* 2000).

As an example of this issue, let's compare the fitting of select ITE and BTE directional hearing aids on a single patient. A likely outcome using a percent correct speech recognition test at a fixed SNR is shown in Figure 18. From this example it is obvious that, while the performance of these two hearing aids in directional mode is equal, they are certainly not equivalent hearing aids in terms of directional benefit or performance in omnidirectional mode. This scenario can be especially troublesome if the performance parameters of these two hearing aids are unknown to the fitting audiologists. Specifically, if these two hearing aids were simply fit to the patient without validation, a much greater wow effect resulting from the large difference between directional and omnidirectional modes would likely occur for the ITE than for the BTE.

Caution must be exercised against misinterpreting these results as a recommendation against fitting BTE style hearing aids. Obviously for many patients, there are a variety of other

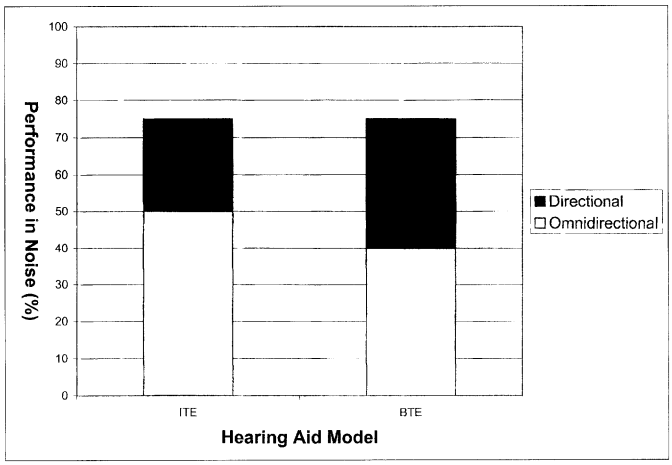

Figure 18. A sample percent correct speech recognition performance as measured at a fixed SNR for a representative listener fit with ITE and BTE style directional hearing aids.

fitting factors that will yield to a generally more successful fit with BTE in comparison to ITE style hearing aids. Potential BTE advantages (in comparison to ITE hearing aids) including, but not limited to, greater power, greater headroom, reduced feedback problems, and reduced cerumen /maintenance issues must of course, also be considered when making decisions as to the appropriate style. In addition, one must remember that differences in performance in noise when in omnidirectional mode are of little consequence, if this mode is primarily used in quiet environments.

Variable and Adaptive Directional Hearing Aids

When examining theoretical directional patterns, it is easy to see how the selection of the best pattern may be difficult for a designer of directional hearing aids. If the assumption is made that the listener is primarily surrounded by noise, a hypercardioid pattern seems the best choice. In contrast, this choice will be inferior to a cardioid pattern when noise is directly behind the listener and inferior to a bidirectional pattern when noise is directly to the listener's side. Because of these potential limitations a few manufacturers have developed directional hearing aids with variable directional patterns. This technology is possible since internal delay is applied electronically in twin-microphone directional hearing aids and

these devices can be designed to allow for varying the duration of the internal delay (resulting in variable directional patterns).

One of the first commercially available circuits of this type was the Gennum FRONTWAVE, which was designed to be used with any twin-microphone directional system. The second method, introduced more recently, is designed to adaptively switch between polar attenuation patterns in response to the listening environment. Adaptive directional hearing aid systems were first introduced commercially in the Phonak Claro and have since been implemented in the GN Resound Canta7 and yet to be released Widex Diva products.

Adaptive directional hearing aids operate by automatically varying the physical directional properties until an attenuation pattern that results in the lowest output intensity from the directional microphone is obtained. The adaptation time in commercial hearing aids (that time over which a change in directional pattern occurs) ranges from a few milliseconds to more than five seconds. While there is no data supporting either longer or shorter adaptation time constants, it is clear that shorter time constants are necessary for the directional pattern to adapt to a moving noise source position. The adaptive directional circuitry is limited in hearing aids so that directional microphone parameters that result in directional patterns with nulls in the front hemisphere are excluded from consideration. In this way important sound information that arrives from the front hemisphere is not inadvertently, and undesirably, attenuated. With the front hemisphere attenuation limitation, the assumption is made that the lowest output from the directional microphone will correspond to the greatest noise attenuation.

Data to date suggest some possible advantages and some limitations of one existing adaptive directional system (Phonak Claro) in simulated real-world environments (Gross, 2001; Ricketts, 2001). The primary advantage shown for this technology has been that listeners fit with an adaptive mode performed significantly better when there were competing sound sources at the listeners' sides, while performing as well, or better, in a number of other noise configurations (Ricketts, 2001). These data support the hypothesis that the adaptive circuitry switches to a more appropriate bidirectional type attenuation pattern when sound sources are present at the listener's side(s).

In contrast, data obtained by Gross (2001) revealed no difference in speech recognition performance for subjects fit with the adaptive and fixed directional modes using the HINT and Connected Speech test. The test environment in this study included a panning noise source delivered from five loudspeakers situated from +90 to –90 degrees around the listeners' head. The overall intensity level was fixed at 65 dBA; however, all loudspeakers were active at the same time. A single loudspeaker was randomly selected and the level of this loudspeaker was increased by 8 dB while the other loudspeakers were decreased accordingly. The chosen methodology of an 8 dB increase of a single speaker in a background of noise was based on real-world measurements in noisy restaurants.

In addition to the lack of difference between adaptive and fixed directional modes across speech recognition measures, these authors also reported no subjective preference as measured across scales of hearing aid benefit and satisfaction.

In general, the data from these experiments suggest that current, commercially available adaptive directional microphones provide increased speech recognition in noise in comparison to their fixed counterparts in some specific listening environments, and equivalent performance in others.

Automatic Switching Directional Hearing Aids

Despite advances in remote control and switch technology relative to ease of use, the fact remains that a number of hearing aid wearers are unable to switch between settings due to physical and/or mental limitations. This switching may be especially difficult, or impossible, for very young children, and elderly adults. In response to these concerns, at least three manufacturers have introduced instruments that automatically switch between directional and omnidirectional modes depending on the acoustic environment. The first such system, the Directions sound processor by Audio D, was introduced in 1999. This circuitry is combined with a D-Mic (Etymotic Research) to provide a system that can be programmed to automatically switch between directional and omnidirectional modes without user input. Instead of invoking directional and omnidirectional modes only, this system switches between omnidirectional, quasidirectional (45% omnidirectional and 55% directional), and directional modes,

depending on the intensity level of the input. While the Directions sound processor was designed to switch modes based on sound intensity level only, other systems switch based on a combination of environmental factors including intensity level of the signal, amplitude modulation rate and depth, and spectral shape. While having intuitive appeal, little is known about patient satisfaction with automatic directional systems. Research has shown that patients report large differences in the sound quality of omnidirectional versus nonequalized directional modes. It is assumed that this difference is due both to directivity and the change in frequency response. One might hypothesize that patients may be bothered by an automatic system in a particular environment in which the system repeatedly changes modes. Whether such problems actually exist, and whether they would be alleviated through the use of fully or partially equalized directional modes, is unknown. Personal experience in fitting these instruments leads me to recommend consideration of automatic directional hearing aids, primarily for patients who are unable to appropriately switch between modes on their own.

Summary of Fitting Factors Impacting Directional Benefit and Directivity

Average directional benefit provided by modern directional hearing aids in difficult (near-field) environments is about 20% to 35% as measured by fixed SNR methods and 3 to 4 dB using variable SNR methods. Data to date also reveal that microphone port orientation, venting, low-frequency gain equalization, head turn and hearing aid style can all impact the magnitude of directional benefit obtained by patients. In contrast, compression and monaural versus binaural fittings appear to have little or no impact on the measured directional benefit. Given the potential interaction between these fitting factors and directivity, it may often be desirable to assess the directivity of instruments fit to individual patients in the clinic. I think this is especially important given the fact that directional microphones, like all other parts of a hearing aid, are susceptible to damage. Unfortunately however, this damage may not be revealed by traditional electroacoustic evaluation. Finally, the use of adaptive and automatic directional microphones appears warranted for some patients/listening environments.

Directional Hearing Aids: When Won't They Work?

Directional hearing aids certainly are not expected to provide benefit in all environments, so it is important that we provide appropriate expectations regarding these devices during counseling. In terms of noisy environments, we must remember that increasing reverberation and speaker to listener distance can both negatively impact directional benefit. Simply stated, directional amplification does not reduce the need for appropriate and preferential seat selection by, and for, listeners with hearing impairment. We must also keep in mind that the range of SNR over which benefit is expected may vary from patient to patient (as demonstrated by the previous Lucky and Amanda example). Consequently, it is important that we are careful not to overstate the general effectiveness of these devices for specific patients.

It has long been known that directional benefit is not present, or expected, in quiet listening situations (ie, Frank and Gooden, 1973). A number of investigators have examined whether this lack of directional benefit is revealed in subjective measures. For example, Mueller and associates (1983) examined the subjective preferences between omnidirectional and directional microphones using a questionnaire. Subjects in this investigation were fit monaurally with a hearing aid capable of being switched between omnidirectional and directional modes. While there was a clear preference for the directional microphone in noisy environments, results revealed no clear preference for either directional or omnidirectional modes when listening in quiet.

More recently, Kuk (1996a) surveyed 100 users of a hearing instrument incorporating multiple microphone technology. In contrast to the data presented by Mueller and associates (1983), not only did listeners not show a preference for directional amplification in quiet listening environments, many listeners reported a preference for omnidirectional amplification. Sixty-five percent of respondents stated a preference for the omnidirectional mode in quiet environments, while 25% preferred the directional mode, and 10% reported no preference. Other interesting findings included a stated preference for omnidirectional amplification when listening in a car, as well as when listening to nature sounds, warning sounds, music, your own voice, and while eating, but preferred the dual microphones when listen-

ing to annoying sounds. In addition, 67% of the users stated they regularly switched between directional and omnidirectional modes, with only 15% reporting constant use of the directional mode.

The preference of hearing aid wearers' for switchable as opposed to full-time directional hearing aid configurations was also reported by Wolf and associates (1999). These authors reported results from a survey mailed to 125 users of a dual microphone directional hearing aid. Of these respondents, 73% reported switching between directional and omnidirectional mode two or more times a day, while 35% reported switching five or more times a day.

Preves and coworkers (1999) reported results from paired comparison testing indicating that approximately 80% of listeners stated an overall preference for an equalized directional mode in noise over an omnidirectional mode. Most subjects, however, also indicated that they would prefer to have the ability to switch between directional and omnidirectional modes. The authors stated that this preference was related to the fact that more subjects preferred the omnidirectional mode when listening in quiet. These data revealed a preference for the omnidirectional mode in quiet even when the frequency response in the directional mode was equalized. Consequently, this preference could not have been due to differences in frequency response. Potential reasons for the preference for omnidirectional amplification in quiet include, but are not limited to, differences in sound quality due to the increased audibility of microphone noise and/or reduced audibility for sounds that were not in front of the listener.

Despite evidence that hearing aid wearers desire the ability to switch between directional and omnidirectional modes, one recent study suggests that this desire does not outweigh the preference for directional amplification. Valente and associates (1999) examined the speech intelligibility and user preference of 40 listeners with hearing loss across two hearing aids. These two hearing aids were identical in all processing features except that one was omnidirectional and the second instrument was directional only. That is, patients fit with this second hearing did not have the option of switching to an omnidirectional mode. Results revealed that subjects not only demonstrated better speech recognition in noise with the directional hearing aid, but also stated a significant preference for this hearing aid over its omnidirectional counterpart, or their own hearing aids. These results suggest that the performance increase achieved by these patients in noise may have outweighed problems associated with the use of a directional hearing aid in quiet. It should be noted that the directional hearing aid used in this study has been reported to have significantly lower directivity, especially in the high frequencies, when compared to the majority of current, commercially available, hearing aids (Ricketts, 2000a). It might be speculated that this reduced directivity may have acted to reduce the negative impact of directivity in quiet settings. While also likely reducing directional benefit in noise, it appears that significant directional benefit was still achieved by wearers of this instrument. Further study is necessary to determine if the magnitude of directivity affects the negative impact of directional hearing aid use in quiet environments, however.

General Summary

Directional amplification represents one of only a handful of methods that have been shown to consistently improve SNR for listeners across a wide range of noisy environments. Data supporting the use of these devices to aid the speech understanding of listeners with hearing loss in noisy situations are overwhelmingly positive. It is equally clear, however, that a number of factors can impact directivity and directional benefit. Due to individual differences, some patients may not achieve significant directional benefit. Venting and the orientation of the microphone ports are known to reduce directivity. Little or no directional benefit is expected in specific listening environments especially those with high reverberation and large speaker-to-listener distances. In some situations, such as listening in quiet and listening to talkers which are not in front of the hearing aid wearer, directional amplification may be undesirable or even detrimental. Despite these limitations, it seems clear that the use of directional amplification, combined with appropriate counseling and expectations, can lead to increased speech understanding in noise and increased hearing aid satisfaction for many listeners with hearing impairment.

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