In his book *Out of the Crisis*, W. Edwards Deming (2000) reports that satisfied customers tell eight of their friends, but dissatisfied customers tell their troubles on average to 16 persons. Think of our “customers” as our hearing aid patients. Research by Kochkin (2000) reveals that 16.2% of hearing aid owners (which amounts to approximately one million persons in the United States) never use their hearing aids. Why aren’t these one million persons satisfied, and who are the 16 million that they are telling their troubles to? Their neighbors? Their hearing-impaired friends? Their family physician? There are of course many reasons why people are dissatisfied with the performance of their hearing aids, and we have little or no control over many of these factors. There is one important factor, however, where we have complete control: the selection and adjustment of the hearing aid’s maximum output.

What if a patient is fitted with hearing aids that have a maximum output that is too high (the most probable mistake)? Back in 1984, David Hawkins pointed out that if the hearing aid output exceeds the patient’s loudness discomfort level (LDL), then during the first week or two of hearing aid use the user will normally adopt one of four maladaptive strategies. These four strategies still occur today, at least for those patients whose aids are fitted with volume control wheels (VCWs):

- The patient will constantly rotate the VCW to adjust for different input levels. Patients quickly tire of this approach and try the next strategy.
- The patient only uses the hearing aids in quiet listening situations, where input levels are low enough so that loudness discomfort does not occur. Patients also quickly tire of this strategy, as they prefer to use their hearing aids in a variety of situations. So, they move on to the next maladaptive strategy.
- The patient always uses a low VCW setting so that hearing aid gain plus the input level does not exceed their LDL. Unfortunately, when this strategy is employed, average conversational speech is not loud enough, and the high-frequency components of speech might not be audible at all. Consequently, the hearing aids provide little benefit, which leads to the final strategy.
- *The patient stops using the hearing aid!*

As we mentioned, the above four-step progression would only apply to patients using hearing aids equipped with VCWs. If a VCW
is not available, which is true for most completely-in-the-canal (CIC) products, and the hearing aid output is set too high, the patient’s only two choices are the second and fourth strategies listed on the previous page.

Do patients actually reject hearing aids because the output is too high? Franks and Beckmann (1985) found that in a group of geriatric patients who rejected their hearing aids, the leading complaint was that the hearing aids “made sounds too loud.” This finding, in fact, was reported by an overwhelming 88% of the patients. The Franks and Beckmann research, however, was conducted 20 years ago, when hearing aid compression was not commonly used, and when hearing aids were not as adjustable as they are today. Have things gotten better? Probably, but excessive output still appears to be a major contributing factor to hearing aid rejection. Kochkin (2000), in a study of 348 hearing aid owners who never use their hearing aids, noted that one of the leading reasons for nonuse was that the hearing aid made sounds too loud. It is also possible that an inappropriate loudness level contributed to the leading cause of hearing aid rejection; “the hearing aid provides little or no benefit.” As we discussed earlier, the patients might be turning down hearing aid gain.

Even more relevant research, also reported by Kochkin (2000), examines the satisfaction levels for people using their hearing aids (the survey was conducted in 1997, \( n = 1779 \)). For the item “comfort with loud sounds,” overall patient satisfaction was only 38%! Satisfaction was somewhat higher than average for those people fitted binaurally, with CICs, and with programmable instruments (although not stated, it is probable that the majority of programmable instruments employed some type of compression). Of the 25 categories on the survey related to hearing aid performance and different listening environments, only two items received a lower satisfaction rating: use in noisy situations (30%), and listening in a large group (25%). In a separate analysis, Kochkin (2000) examined the satisfaction levels for those people using hearing aids that were only 3 to 12 months old, that is, hearing aids fitted in 1996 or 1997. For this group, when responding to the category “comfort for loud sounds,” only 44% were satisfied, 29% were neutral, and 27% were dissatisfied. Clearly, the selection and adjustment of the hearing aid’s maximum output remains a critical component of the hearing aid fitting process, and these poor user-satisfaction ratings suggest that there is significant room for improvement in this area.

When the maximum output of the hearing aid is selected correctly, at least four different benefits are realized by the patient (Hawkins et al, 1992; Mueller and Bright, 1994). We believe that these benefits will lead to increased patient satisfaction:

- Physical discomfort from auditory signals is minimized (or even eliminated) for speech, noise, and environmental sounds in everyday listening situations.
- Perceptual discomfort from auditory signals, including one’s own voice, is minimized.
- The dynamic range available to the patient is sufficiently wide and maximized whenever possible.
- The hearing aid output is limited below the level that will cause additional hearing loss.

**Review of Terminology**

There are several terms that have been used to describe loudness ratings, loudness sensations, and the patient’s uncomfortable loudness levels. In this chapter, we will use the term **loudness discomfort level (LDL)**, when referring to the uncomfortable loudness level, but it’s important to briefly review other terms that often appear in the literature (adapted from Mueller and Hall, 1998):

- **Uncomfortable level (UCL)**: Probably the most popular term, this is the dB level [dB hearing level (HL) or dB sound pressure level (SPL)] at which a given sound is uncomfortable for the listener.
• **Uncomfortable loudness level (ULL):** Same as the UCL.
• **Loudness discomfort level (LDL):** Same as the UCL.
• **Threshold of discomfort (TD):** May be the same as UCL, but used to indicate that the discomfort experience can be related to factors other than “loudness.”
• **Output sound pressure level (OSPL90):** The maximum output (dB SPL) for a hearing aid with a 90-dB SPL input measured in the 2-cc coupler.
• **Real-ear saturation response (RESR):** The maximum output of a hearing aid measured in real-ear SPL.
• **Upper level of comfort (ULC):** Highest loudness level (dB HL or dB SPL) that is loud, but not uncomfortably loud.
• **Upper level of comfortable listening (ULCL):** Same as the ULC.
• **Highest comfortable level (HCL):** Same as the ULC.
• **Dynamic range (DR):** The dB range between the patient’s threshold and LDL.
• **Maximum comfort level or most comfortable level (MCL):** A specific dB level where a given sound (usually a speech signal) is the most comfortable (dB HL or dB SPL).
• **Range of comfortable loudness (RCL):** The dB range between the lowest and highest level that a sound is comfortable.
• **Recruitment:** Abnormal growth of loudness as a result of cochlear pathology; if a person has cochlear pathology, and their dynamic range is reduced, they have recruitment.
• **Hyperacusis:** Abnormal reactions to moderately loud sounds.

**Methods for Selecting the Maximum Output of Hearing Aids**

It’s another Monday morning and you are scheduled to see Mr. Jones. You have seen him before, and you know that he’s here today to order a pair of new hearing aids. In past visits, he’s told you about the problem with his old mail-order hearing aids—they often made loud sounds too loud. Now is your chance to show him what a properly trained audiologist can really do. Right? Well, what are your options for determining the correct maximum output for Mr. Jones’s new hearing aids? One choice would be to do nothing and simply mail an audiogram to your favorite manufacturer. That seems rather negligent, so that’s an easy one to eliminate. A second choice would be to order hearing aids that have highly adjustable output controls and assume that the output can be adjusted correctly on the day of the fitting. We’ll talk about that approach later in this chapter, but for now, let’s assume that you would like an estimate of Mr. Jones’s LDLs to assist you in the selection of the hearing aid’s OSPL90. You then have at least five more choices:

• Conduct LDL measurements using earphones (supra-aural or insert).
• Conduct complete loudness scaling using earphones (supra-aural or insert).
• Conduct loudness measurements using a hearing aid (referred to as in-situ testing).
• Assess loudness perceptions using patient history or self-assessment inventories.
• Predict the LDL from the patient’s threshold.

There are advantages and disadvantages to each of these methods, and there is no clear consensus regarding which approach is best. From a clinical standpoint, there is not even a clear definition of what we mean by “best.” For instance, if a method provides improved accuracy for about one-tenth (only) of the fittings, but adds about 30 minutes of testing time, is it really the best method? In the next section, we review each method, and provide some examples of how they can be used in the clinic.

**Earphone LDL Measurements**

This is the procedure that you probably are the most familiar with. In surveys of audiology procedures, it is usually reported that approximately 70% of audiologists conduct LDLs (in most cases, however, this is an LDL for speech) (Mueller and Strouse, 1995).
Using speech material for unaided LDL assessment is popular, but is of little value. In this section, therefore, we will be referring only to LDL testing using narrow-band signals (but not referring to complete loudness scaling, which will be reviewed in a separate section). Usually this LDL procedure involves starting with a presentation level that is at or around the patient’s MCL, and then ascending in 2- or 5-dB steps until the LDL is reached. The ascending run is repeated two or three times and a value believed to be a valid measure of LDL is recorded.

The obvious reason for routinely conducting LDL measurements is that the clinician believes that LDLs cannot be predicted accurately from the patient’s pure-tone thresholds. Although threshold-based LDL prediction procedures will often be acceptable, most clinicians have experienced patients whose amplification needs are not met by “average” prescriptive formulas, as there is a significant variance above and below the average predictive values. For example, Elberling (1999) estimated that the slope of the measured loudness function can be predicted from the hearing loss with an accuracy corresponding to a ±5-dB fine-tuning of the gain for 70% of hearing-impaired individuals. He labeled the remaining 30% as either “sound sensitive” (12%) or “sound addicts” (17%). Is it worth the time investment to identify this portion of your clinical caseload?

To understand the potential mistake that can be made by predicting the LDL, it is important to review large-scale studies that have compared hearing thresholds to LDLs on an individual basis. Although a general relationship between degree of hearing loss and LDL exists, multiple studies have demonstrated that, given the same degree of hearing loss, individual judgments of LDLs can vary dramatically (e.g., Kamm et al, 1978; Shapiro, 1979; Pascoe, 1988; Bentler and Cooley, 2001). When viewing the range of individual results from these studies, the percent of outliers appears even larger than the 30% value suggested by Elberling. We have included the results of two of these studies to illustrate this point.

Figure 2–1, from Kamm et al (1978), shows the range of LDLs for 500 Hz and 2000 Hz pure-tone stimuli. For example, viewing this
figure, let's take someone with a 50-dB hearing loss at 2000 Hz, a common finding in the fitting of hearing aids. Observe that measured LDLs vary from 82- to 124-dB SPL, a range that greatly exceeds the output adjustment capability of a hearing aid. Figure 2–2, taken from the research of Bentler and Cooley (2001), also shows the large range of LDLs (reported as TDs) observed across subjects with similar degrees of hearing loss. The range of LDLs obtained in this study is even greater than that shown in Figure 2–1; note that LDLs for most hearing losses less than 60 dB have a range of over 50 dB. If we return to our example of someone with a 50-dB hearing loss, the LDLs now range from 72- to 132-dB SPL (re: 2-cc coupler).

Given the results shown in Figures 2–1 and 2–2, it is not surprising that several researchers have recommended that, whenever possible, LDLs should be measured, not predicted from hearing thresholds (e.g., Hawkins et al, 1987; Mueller and Bentler, 1994; Valente and Van Vliet, 1997; Valente et al, 1997). In the 1990s, two different consensus statements were published regarding the selection and fitting of hearing aids for adults (Hawkins et al, 1991; Valente et al, 1998). In both protocols, the measurement rather than the prediction of LDLs is recommended.

One reason why many researchers and clinicians advocate a threshold-based LDL prediction technique rather than behaviorally assessing each patient's LDL is that multiple methodologic factors can influence LDL measures. Variables such as instructions, procedures, learning, and training all can influence a subject's perception of loudness discomfort. For example, Elberling (1999) compared normative data from seven different loudness-scaling procedures. The

![Figure 2-2](image-url)

**Figure 2-2.** Loudness discomfort levels (LDLs) as a function of hearing threshold level. Each data point represents an LDL for one of five test frequencies (500, 1000, 2000, 3000, and 4000 Hz). (From Bentler R, Cooley L. An examination of several characteristics that affect the prediction of OSPL90 in hearing aids. *Ear Hear* 2001;22:58–64. Reprinted with permission from Lippincott Williams & Wilkins.)
sound level that subjects with normal hearing rated as “too loud” ranged from 92.5 dB HL to 125.3 dB HL, depending on the procedure. Although controlling for these variables can increase reliability, it is not obvious which LDL is the “right” LDL. In other words, there is no “gold standard” for LDL measures. What is desired, of course, is an audiometric measure that can be used to program the hearing aid, which results in a hearing aid maximum output setting that will provide the patient with the appropriate loudness perceptions in the real world. We will discuss making the connection between measured LDL values and hearing aid adjustments later in this chapter. First, it’s important to consider the variables related to the LDL measurement itself.

**Descriptive Anchors**

As with many auditory psychophysical procedures, it is helpful to provide the patient with descriptive anchors prior to conducting the test. This helps the patient understand the test boundaries and provides the patient with a set of terminology to use to describe the listening experience. Over the years there have been several different lists of descriptive anchors that have been used for LDL measurements. We recommend using the following list, adapted from the Cox Contour Test (Cox, 1995):

- 7 Uncomfortably loud
- 6 Loud, but okay
- 5 Comfortable, but slightly loud
- 4 Comfortable
- 3 Comfortable, but slightly soft
- 2 Soft
- 1 Very soft
- 0 Cannot hear at all

The Cox Contour Test is a component of the Independent Hearing Aid Fitting Forum (IHAFF), and the only modification we made to the original Cox anchors was to add the category “Cannot hear at all.” There has been considerable research conducted using these anchors, both unaided and aided, and normative data are available (Cox et al, 1997; Cox and Gray, 2001). These anchors are the same as those published by Hawkins (1984), except that Hawkins had two additional loudness categories at the top end: “Extremely uncomfortable” and “Painfully loud.” Usually, the goal of LDL testing is to determine the point where sound is uncomfortable, not extremely uncomfortable or painful, and therefore Cox did not use these two categories from Hawkins. Moreover, when some patients see the term “Painfully loud,” they become anxious and alter their responses to ensure that this level is never presented. It’s interesting to note that the Pascoe (1988) LDL data, the values that many manufacturers use in their predictive fitting algorithms, were collected using “Very loud” and “Too loud” as the top two categories.

We suggest that the Cox (1995) descriptive loudness anchors be presented to the patient on a large easy-to-read board. The patient simply can call out the number of the loudness rating. Some clinicians report that they have the patient point to the response, as this often provides additional information regarding the certainty of the judgment. These anchors, however, are meaningful only if they are used with the appropriate instructions (see the following section).

To this point, we primarily have discussed LDL measurements for adults, but it is also useful to obtain loudness judgments with children. Examples of descriptive anchors that can be used with children are shown in Figure 2–3. The faces shown on the left side of the panel are from Kawell et al (1988). These authors report using these pictorial anchors and obtaining reliable aided LDL measures from hearing-impaired children between ages 7 and 14 years. Recommendations for testing children also have been made by Skinner (1988). The right side of Figure 2–3 shows two sets of pictures developed at Central Institute for the Deaf for use with children. The three-level balloon series of pictorial anchors can easily be drawn to give school-age children a reference for defining loudness levels.
Instructions

Because the determination of LDLs is a behavioral task that requires a subjective response, the instructions given to the listener play an important role. How the instructions are phrased can influence the listener to accept extremely uncomfortable signals, or to reject sounds that are only slightly above MCL. The instructions that are used also can influence the test-retest reliability of the LDL measurement (Bornstein and Musiek, 1993). In general, the instructions that have been reported in the literature to elicit LDLs can be classified into three categories (Hawkins, 1980a):

- Initial discomfort: This category of instruction suggests that the first point at which discomfort is experienced is the level being sought. For example, individuals may be instructed to respond at “the point where sound first becomes annoying.”
- Definite discomfort: A more pronounced level of discomfort is implied in this category of instructions. Individuals are instructed to respond “when the sound is so loud that you would choose not to listen to it for any length of time.”
- Extreme discomfort: Instructions in this category suggest pain or other physiologic signs of discomfort such as dizziness or tactile sensation.

Although this last category might sound like an improbable clinical test protocol, a hearing aid fitting guide published by Wallenfels in 1967 proposed this LDL procedure: “As soon as you see the muscles around his eyes start twitching, you have your measurement.” Must have been a 1960s thing!

As you might guess, the LDL test instructions can influence significantly the outcome of the test. Beattie et al (1979) compiled LDL results for normal-hearing listeners from a number of studies. Their compilation revealed that as the instructions changed, so did the LDLs. For example, the LDLs for speech stimuli obtained using different instruction sets ranged from 90.5 (initial discomfort) to 137.9-dB SPL (sharp pain). We often use the term “tolerance” when we refer to the hearing aid’s maximum output setting. The term “tolerate,” however, when used in the context of LDL measurements, often will provide an LDL much higher than the patient really will “tolerate” when wearing hearing aids in the real world.

We suggest using the instructions from Cox (1995), which correspond to the loudness anchors previously displayed. This will allow you to compare your findings with the published norms using these anchors. The instructions are as follows:

“The purpose of this test is to find your judgments of the loudness of different sounds. You will hear sounds that increase and decrease in volume. You must make a judgment about how loud the sounds are. Pretend that you are listening to the radio at that volume. How loud would it be? After each sound, tell me which of these categories best describes the loudness. Keep in mind that an uncomfortably loud sound is louder than you would ever choose on your radio no matter what mood your are in. When responding to each sound, it is okay to skip a category, or to repeat a category.”

Notice that these instructions are probably geared more toward initial discomfort than definite discomfort. We would expect, therefore, that the patients would provide a fairly conservative estimate of LDL. When using hearing aids, however, patients often respond immediately to initial discomfort (by turning down gain), so indeed this may be the best level to use.

As discussed earlier, LDL testing often is conducted with children, and the instructions must be adjusted for this population. Listed below are the instructions from Kawell et al (1988), developed to correspond with the descriptive loudness faces shown on the left portion of Figure 2–3:

“We’re going to see how loud this hearing aid makes sounds. You will hear some whistles and I want you to tell me how loud the whistle is. [Go over the descriptor list, explaining each choice, starting with “Too soft.”] When the sounds are too loud, this is where you want the hearing aid to stop and you do not want the sounds to get any louder. Now, for every whistle, tell me how loud it sounds.”

Psychophysical Procedures

Several different types of psychophysical procedures have been used to measure LDLs, such as magnitude estimation, magnitude production, and the most common, categorical scaling. Psychoacoustic approaches that have been employed include both clinician-controlled and listener-controlled procedures, frequently using some type of adaptive technique. The clinician-controlled approach can be either the simple up-down procedure or the ascending approach, sometimes referred to as a bracketing technique. The listener-controlled method usually has been the method of adjustment, often using Bekesy audiology. Research comparing different procedures has revealed that somewhat higher LDLs are obtained using the patient-controlled Bekesy tracking technique (Stephens and Anderson, 1971).

In clinical practice, the most commonly used procedure is the ascending approach. A sample protocol is listed below, modified from the procedures of the Cox Contour Test of the IHAFF protocol (Cox, 1995; Valente and Van Vliet, 1997):

- Frequency-specific stimuli should be used, with a preference for warble or pulsed tones over narrow-band noise for patients with precipitous losses.
- Stimulus duration may be gated manually or pulsed (200 msec on/off) through the audiometer.
- A minimum of two frequencies should be used for assessment of loudness.
growth; the recommended frequencies for the approach are 500 Hz and 3000 Hz, or more appropriate frequencies depending on the configuration of the hearing loss (e.g., if the patient has normal hearing at 500 Hz, a higher frequency would be selected; if the hearing loss is so severe at 3000 Hz that it’s not worth “chasing,” then a lower frequency would be selected).

- Present the stimuli via insert earphones.
- Prior to the actual test, a practice run should be completed for a selected frequency. The patient’s task is to respond to varied presentation levels of the loudness scale.
- Use an ascending approach in 5-dB steps (or 2-dB steps for listeners with narrow dynamic range) until an “uncomfortably loud” response is obtained. Return to the starting intensity level and repeat for a minimum of three runs.
- The patient’s LDL is the median value of the “uncomfortably loud” ratings from the three runs.

With children, it is often necessary to modify the psychophysical procedures used with adults. As we mentioned earlier, Kawell et al (1988) report on the successful use of categorical scaling (using five different faces paired with descriptive titles), for children 7 to 14 years old, although Ellis and Wynne (1999) report poor reliability for this procedure. An alternative approach that has been used with children is cross-modality matching (CMM). Hellman (1999) has reported on the clinical utility of using CMM for loudness and line length, and limited research has shown that this technique is reliable for children as young as 4 years of age (Collins and Gescheider, 1989). In recent research with hearing-impaired young children, Serpanos and Gravel (2000) used CMM for loudness and line lengths—eight different smiling caterpillars varying in length from 0.52 to 65 cm (a ratio of 125:1). These authors report that the CMM method is a valid and reliable technique to assess loudness growth in children with sensorineural hearing loss as young as 6 years of age.

**Effects of Signal Type**

Hawkins (1980b) evaluated the influence of several types of signals on LDLs of normal-hearing listeners. Although the LDLs for 250-Hz stimuli were somewhat higher than for other frequencies, he found no significant difference in LDLs for a group of 19 subjects who listened to pure tones, narrow-band noise, filtered speech, wide-band noise, sentences, and spondees (Table 2–1). Although Hawkins reported no significant difference between stimulus types for the group data, he did note substantial intraindividual differences.

Bentler and Pavlovic (1989) did find some differences in LDLs related to signal type, and differences between normal hearing and hearing impaired subjects. In further study, Bentler (1993) reported that LDLs obtained with 10 different environmental stimuli

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S.D., standard deviation; S.E., standard error.

*Adapted from Hawkins 1980b.
were not equal and were not directly related to the crest factor of the stimuli (crest factor: difference between the peak and the root mean square of a signal). In related research, Warner and Bentler (2001) identified several factors that affect the unaided perception of loudness discomfort other than the overall root mean square (RMS) level of the signal. These authors measured LDLs for subjects with normal hearing using multitone complexes and environmental sounds (e.g., baby crying, glass shattering). They found that the overall RMS level required to elicit a perception of loudness discomfort varied between stimuli. In other words, level alone did not determine the perception of loudness discomfort. Stimuli were described in terms of objective acoustic factors (e.g., peakiness, high- and low-frequency cutoff, center frequency) as well as subjective sound quality factors such as ratings of “annoyance,” “harshness,” “loudness,” “noisiness,” and “tinniness.” The authors used multiple regression analysis to identify factors that could be used to predict differences in LDLs between stimuli. Acoustic factors associated with reduced LDLs included increased peakiness of the frequency response and increased high-frequency content. In addition, the subjective perception of tinniness also resulted in reduced LDLs.

These results lead us to ask two important questions: (1) How do the different stimuli interact with the processing of hearing aids? (2) What clinical stimuli, if any, are reliable predictors of real-world LDLs? We will address both of these questions later in this chapter.

**Earphone Style**

Although insert earphones normally are recommended for clinical use, the older supraaural earphones (e.g., TDH-39, -49, or -50) seem to be maintaining their popularity in many offices and clinics. It is important to consider, therefore, if earphone style has an effect on the measured LDL. This was examined by Valente et al (1997), who compared LDLs for 31 hearing-impaired ears referenced to both HL and ear canal SPL for the two earphone styles. The good news is that the results of this study suggest that these earphones can be used interchangeably and the LDL will not be significantly altered [assuming appropriate reference equivalent threshold in SPLs (RETSPLs) have been used]. As would be expected, the largest mean variance was observed using the HL reference, but even then, most differences between earphones were only 1 to 2 dB with a 3-dB difference at 4000 Hz. When referenced to ear-canal SPL, the largest mean difference observed at any frequency (1500 Hz) was 0.9 dB.

**Test-Retest Reliability**

First, as with any behavioral measure some test-retest variability is observed when determining subject-specific LDLs. Depending on the methodology this test-retest variability may be quite small or quite large (e.g., Bornstein and Musiek, 1993; Robinson and Gatehouse, 1996). In several studies, the test-retest for LDL has been shown to be about the same as for pure-tone thresholds. For example, Ricketts and Bentler (1996), testing a group of normal hearing individuals, reported an average difference of 3 to 4 dB. Hawkins et al (1987), using subjects with hearing impairment, also reported that LDL test-retest variability was small (maximum deviations of 3 to 4 dB over a 4-day period) if certain criteria were met. Hawkins et al suggested that valid repeatable LDL measures could be obtained if (1) subjects understood the purpose of the LDL task; (2) the loudness criterion was clear (e.g., is the audiologist looking for annoyingly loud or painfully loud?); and (3) descriptive anchors were available for loudness categories both above and below the target category. Using 10 subjects with hearing loss, Cox et al (1997) reported slightly higher test-retest variability than did Hawkins et al (1987) for uncomfortably loud stimuli when using the contour test. Measures were made over two test sessions ranging from 3 to 15 days apart. Test-retest differences were typically less than 6
I dB (standard deviation of 5.9 dB), although a few subjects (2%) showed differences greater than 10 dB.

Palmer and Lindley (1998) also examined the test-retest reliability of the contour test using subjects with hearing loss and found that test-retest reliability varied across frequency and loudness category. These authors report a mean test-retest difference of 2.6 dB across five test frequencies for the no. 7 rating (uncomfortably loud). Standard deviations ranged from 5.7 dB (3000 Hz) to 8.1 dB (500 and 1000 Hz). Similar to the Cox et al. (1997) study, 94% of test-retest differences were less than 10 dB. Although the studies cited above used accepted clinical test protocols for determining LDL, all were completed under controlled laboratory conditions. It is possible that test-retest differences would be somewhat larger when these measures are performed clinically.

Training Effect

Related to the topic of test-retest reliability is the consideration of a training or practice effect for the LDL measurement. Behavioral tasks like LDL assessment can be influenced by procedural, stimulus, or task learning (see Robinson and Summerfield, 1996, for review). For example, Morgan and Dirks (1974) report a 6- to 8-dB increase in LDLs over the first four of six test sessions for normal listeners. Walker et al. (1984) and Sammeth et al. (1989) reported an increase of a similar degree for hearing-impaired listeners across test sessions a few days apart. In contrast, Cox et al. (1997) reported no significant change for hearing-impaired subjects' overall loudness judgments tested two different times at intervals separated by 3 to 15 days. Palmer and Lindley (1998), who also conducted research with the Cox Contour Test, report that slightly less intensity was needed to attain the same loudness rating on retest (test sessions separated by 2 weeks). It appears that the effects of learning are relatively small, are most notable when the session is repeated within a short period of time, and do not appear to be a concern when OSPL90 is selected.

Acclimatization

When a patient’s LDL is measured and this value is used to select or adjust the OSPL90 of the hearing aids, the general assumption is that the patient’s LDL will remain stable following hearing aid use. If, however, the patient experiences “acclimatization” for loudness, then it is possible that the LDL would increase, and the OSPL90 would then need to be readjusted. Does acclimatization for loudness occur?

In the classic hearing aid paper on the Harvard Report (Davis et al., 1946), the hearing aid with the least maximum output was dubbed “Beginners,” suggesting that the hearing aid users could move on to more powerful hearing aids after some experience. In 1947, Silverman encouraged “tolerance training” by exposing patients to sounds just below their LDL for several minutes a day over 3- to 4-week intervals. Tolerance training is not a common clinical practice today, at least not conducted intentionally. There is some research to suggest, however, that a patient’s LDLs do increase after tolerance training, although in some research this simply could be reflecting a practice effect (see Byrne and Dirks, 1996 for review). Certainly, acclimatization for high-level sounds does not occur to a large degree, or the satisfaction rate would be higher than 38% when hearing aid users are asked to rate “comfort with loud sounds” (Kochkin, 2000).

Even if acclimatization occurred frequently for the patient’s upper level of comfort, it is not practical to purposely fit individuals with output that exceeds their LDL, as excessive output is one of the leading causes of hearing aid rejection, or at the least will prompt the patient to reduce the gain of the hearing aids. In general, when this approach has been used in research, adaptation to the loud sounds did not occur for many of the patients (Lindley, 1999; Lindley et al., 2001). When hearing aids are initially
programmed so that the maximum output corresponds with the patient’s LDL, the unaided LDL does not appear to increase significantly after prolonged hearing aid use (e.g., Bentler et al, 1993; Lindley et al, 2001; Mueller and Powers, 2001). Acclimatization for high-level outputs, therefore, does not appear to be a major area of concern when the OSPL90 of the hearing aid is selected. Certainly, on postfitting visits, a quick check of the patient’s loudness rating for loud speech (e.g., 85-dB SPL) would provide an indication if the hearing aid’s maximum output was set too low.

**Validity of Measures**

The primary purpose of conducting LDLs is to obtain values that can be used to determine appropriate gain for loud inputs and to set the hearing aid’s maximum output (although indirectly, many fitting algorithms also use LDLs to determine gain for soft and average inputs too). Because LDL testing usually is conducted with an audiometer, the results typically are expressed in HL, which must be converted to 2-cc coupler. It is important to consider, however, that the value that would be the most useful for hearing aid selection would be the LDL expressed in ear canal SPL. This leads us to question the relationship between LDLs expressed in HL and ear canal SPL. Commonly, HL values are converted to ear canal SPL values using an average correction referred to as the real-ear dial difference (REDD). The REDD is the total of two other correction factors: the real-ear coupler difference (RECD) and the reference equivalent threshold in SPL (RETSPL).

For example, if a patient’s LDL at 3000 Hz was 102 dB HL, and it is assumed that the average RECD for 3000 Hz is 9 dB, and the RETSPL for 3000 Hz (insert earphones) is 3 dB, then the predicted LDL, referenced to ear canal SPL, would be 114 dB (102 + 9 + 3 = 114). Because the RECD (or REDD) is seldom measured when fitting adults with hearing aids, it is important to consider individual variation from average REDD values.

Valente et al (1994) assessed the REDDs for 50 adult ears for both supra-aural and insert earphones. The results are shown in Table 2-2. For example, let’s assume that the frequency of interest is 3000 Hz, and that supra-aural earphones are used for the LDL measures. Observe that the standard deviation is 5.6 dB and the range is 30 dB. Stated another way, this means that at 3000 Hz, two people with identical LDLs (re: ear canal SPL) could have audiometric LDLs that differed by as much as 30 dB, which, if average corrections were used, would lead to very different prescriptions of gain and output. This illustrates how individual real-ear correction factors will improve the validity of

| Table 2-2. Real-Ear Dial Difference (REDD) for 50 Adults* |
|---------------------------|---------------------------|---------------------------|---------------------------|---------------------------|---------------------------|---------------------------|
| Earphone                  | Frequency (Hz)            |                           |                           |                           |                           |                           |
|                           | 500                       | 1000                      | 1500                      | 2000                      | 3000                      | 4000                      |
| TDH-39P                   |                           |                           |                           |                           |                           |                           |
| Mean                     | 99.3                      | 99.0                      | 98.7                      | 103.1                     | 101.1                     | 95.2                      |
| SD                       | 4.9                       | 2.4                       | 3.5                       | 4.6                       | 5.6                       | 7.3                       |
| Range                    | 20.0                      | 9.0                       | 16.0                      | 23.0                      | 30.0                      | 36.0                      |
| ER-3A                    |                           |                           |                           |                           |                           |                           |
| Mean                     | 88.9                      | 92.9                      | 96.3                      | 99.5                      | 92.6                      | 88.7                      |
| SD                       | 5.8                       | 2.9                       | 4.3                       | 5.6                       | 4.0                       | 5.6                       |
| Range                    | 23.0                      | 12.0                      | 21.0                      | 29.0                      | 20.0                      | 25.0                      |
| Mean difference          | 10.4                      | 6.1                       | 2.4                       | 3.6                       | 8.5                       | 6.5                       |

*Real-ear SPL values shown obtained using 90 dB HL input signal (Adapted from Valente et al, 1994).
the LDL measurement and increase the precision of the hearing aid fitting.

Obtaining LDLs Using Loudness Scaling

Over the years there has been an interest in using loudness scaling as a method to assist in prescribing hearing aid gain and output. Various manufacturers have developed automated procedures, and loudness scaling is an integral part of the IHAFF protocol (Valente and Van Vliet, 1997). Ricketts (1997) discusses some of the advantages and disadvantages of loudness scaling, and reviews the six most common methods used in the United States.

Elberling (1999) mentions several factors that argue against the general use of loudness scaling techniques for the fitting of hearing aids. For example, he points out that using different methodologies results in different perceptions of a given loudness category. The question then becomes, Which LDL is “right”? Another factor arguing against the routine use of loudness scaling is the fact that the slopes of loudness growth functions can be predicted from threshold for about 70% of persons with hearing loss (within about 5 dB). Finally, for a given procedure, the normative data obtained from subjects with normal hearing show a large variance in the sound level required to elicit a given categorical response (e.g., over 30 dB). Elberling points out that because of this variance many normal-hearing subjects would need some type of nonlinear processing to “restore” their loudness perception to normal. This combination of variance inherent in the loudness growth function for a person with hearing loss and the normative reference implies that even if individual loudness growth functions are obtained for the 30% of people who do not fall within the predictable range (the “sound sensitive” and “sound addicts”), substantial fine-tuning still may be required. And further, from a practical standpoint, even if you learned that your patient had a “curvilinear” loudness growth pattern, if you were fitting a hearing aid that employed linear compression, you would have limited adjustments that you could make. And, of course, there is the clinical time investment—complete loudness scaling across several frequencies can be a lengthy procedure.

For the purposes of this chapter, we are not so much interested in the slope of the loudness growth function, but primarily are concerned with the LDL that would result from conducting loudness scaling. Clearly, loudness scaling requires more time than simply using an ascending LDL procedure that has its starting point at or around the patient’s MCL. This additional time investment could only be justified if the scaling procedure resulted in a more accurate LDL assessment. As discussed, when it relates to a patient’s LDL, “accuracy” is a difficult term to define. What we can address, however, is if loudness scaling results in a “different” LDL than when a more abbreviated LDL procedure is used.

Bentler and Cooley (2001) compared the LDLs that were obtained using an ascending method of limits, and an ascending method of limits with descriptive anchors (the Cox Contour Test). Lower LDLs (i.e., 4 to 6 dB) were obtained using the contour test for three of the frequencies tested (500, 1000, and 2000 Hz), but no significant difference was observed between the two methods for 3000 and 4000 Hz. In clinical practice, it is possible to use descriptive anchors but yet not complete the entire loudness scale (e.g., focus on the area above the MCL).

If the primary goal of LDL testing is to separate the “sound sensitive” and “sound addicts” from the patients with relatively average loudness judgments, then complete loudness scaling (e.g., starting at the patient’s threshold and increasing in 2- or 5-dB steps) probably isn’t necessary. As mentioned earlier, if the wide dynamic range compression (WDRC) of the hearing aid being fitted has only one kneepoint and employs linear compression, the gain, compression kneepoint, and compression ratio within a given channel probably will be programmed the same
whether you enter a complete loudness scaling function or only a single LDL value.

**Using the Hearing Aid to Select Maximum Output**

Based on the previous sections we can conclude that clinically measured LDLs provide at least an estimate of real-world levels of discomfort and as such can be used to select the OSPL90 of hearing aids. The previous sections, however, also point out that substantial variability may exist between unaided and aided LDLs. One likely source of measurement error that may contribute to this variability is the use of average correction factors to convert the unaided LDL value in dB HL into the prescribed dB SPL value measured in a 2-cc coupler (Hawkins et al, 1987). We discussed the variation of individual REDDs, which is illustrated in Table 2–2 (Valente et al, 1994). One method of reducing this measurement error is to utilize in-situ techniques when determining the requirements for maximum hearing aid output.

In this discussion, in-situ techniques refer to the use of the patient’s actual hearing aid as a sound generator for determining threshold levels and/or loudness judgments for sounds presented at various levels and then using this information to set the gain, output, and compression parameters of hearing aids. With the advent of digital hearing aid technology, several manufacturers have written fitting software that enables the digital hearing aid to present programmed test signals (Mueller, 2000). When in-situ measures of thresholds or loudness judgments are made using custom hearing aids, or behind-the-ear (BTE) hearing aids with custom earmolds, the typical conversion factors for the RECD are not needed, as the residual volume with the hearing aids in place is part of the test condition. [Because the input is produced by the hearing aids themselves, this procedure, however, does not account for the microphone location effects (MLEs) that will occur in the real world.] Because some of the conversion errors resulting from the use of average correction factors are removed when using in-situ methods, we might expect that variability between the prescribed maximum outputs across hearing aids also would be reduced.

We examined five commercially available digital hearing aids that are capable of utilizing in-situ techniques to prescribe gain and output. Table 2–3 describes the methods recommended by each manufacturer to determine the maximum output of the hearing aid. All manufacturers utilize the information obtained through these in-situ measures in conjunction with audiometric threshold data to determine appropriate gain, output, and compression parameters. We questioned if these different in-situ procedures would result in different predictions of maximum output for the same patient.

We recruited two subjects with hearing loss to examine how the prescribed maximum output, across hearing aids, varied when in-situ testing was included in the fitting process. Subject B.Y. (75 years old) had a mild-to-severe symmetrical hearing loss and had never worn hearing aids. In contrast, subject J.A. (33 years old) had a long-standing asymmetrical hearing loss, with a normal, sloping to severe sensorineural loss for the right ear, and mild, sloping to profound mixed loss on the left. Subject J.A. had worn hearing aids, although infrequently, for several years. Audiometric thresholds for the ear tested (right ear for subject B.Y. and left ear for subject J.A.) are shown in Figure 2–4.

Both subjects completed the manufacturers recommended in-situ testing procedures, as described in Table 2–3, with each device. Once each hearing aid was programmed, probe-microphone testing was performed to determine the real-ear saturation response (RESR) for each instrument. Testing was conducted using both a 90-dB pure-tone sweep (100 to 8000 Hz in 100-Hz intervals), and individual 90-dB pure tones (500, 1000, 2000, 3000, and 4000 Hz) presented in isolation with several seconds of silence between each signal presentation. Shown in Figure 2–5 is the highest value obtained, regardless of test procedure.

In response to our initial question about whether the prescribed maximum output for a given subject varies when using different in-
I
situ methods, the answer appears to be yes. These data demonstrate clear differences in the maximum output, as a function of frequency, between hearing aids. Differences varied with frequency, and similar to the trend observed using a threshold based predictive formula, the largest range of values (up to 17 dB for subject J.A.) occurred at the frequency extremes, 4000 Hz and 500 Hz, for both subjects. The smallest range of differences occurred at 2000 Hz, where maximum output differences between the hearing aids were no greater than 3 to 6 dB for subjects B.Y. and J.A., respectively. Although differences between hearing aids varied somewhat between subjects and across frequencies, some general trends were apparent. First, across all frequencies, hearing aid 5 (which prescribes gain and output based solely on in-situ feedback and threshold measures) provided outputs that were higher or approximately equivalent to that provided by most other aids. Second, aids 1 and 5 consistently provide greater output in the high frequencies (3000 to 4000 Hz) than aids 2 and 3. In addition to the observed differences between aids as a function of frequency, peak and overall RMS levels between some aids were also substantially different.

Figure 2-6 shows the peak output and overall RMS levels for each aid and for both

<table>
<thead>
<tr>
<th>Manufacturer</th>
<th>Rationale</th>
<th>Technique</th>
<th>Presentation Mode</th>
<th>Stimuli</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Measure entire loudness growth function at multiple frequencies</td>
<td>Subjects push a button corresponding to the perceived signal loudness</td>
<td>Random stimulus presentation across level within a frequency</td>
<td>NBN (selectable frequencies; tested 0.5, 1, 2, and 4 kHz for these subjects)</td>
</tr>
<tr>
<td>2</td>
<td>Measure entire loudness growth function at multiple frequencies</td>
<td>Subjects push a button corresponding to the perceived signal loudness</td>
<td>Random stimulus presentation across level and frequency</td>
<td>NBN (0.5, 1, 2, and 4 kHz)</td>
</tr>
<tr>
<td>3</td>
<td>Define the upper and lower limits of the &quot;comfortable&quot; range at two frequencies</td>
<td>Subjects rate stimulus as either comfortable or too soft/loud</td>
<td>Frequency is fixed; level varies to track lower, or upper, edge of &quot;comfortable&quot; range (5-dB steps)</td>
<td>500- and 3000-Hz pure tones</td>
</tr>
<tr>
<td>4</td>
<td>Equalize loudness across entire frequency range for soft, comfortable, and loud signals</td>
<td>Subjects report if presented tones are equally loud</td>
<td>Individual or swept tones; subjects compare loudness across frequency. Adjust levels in 1- or 5-dB steps</td>
<td>Nine pure tones from 0.5 to 8 kHz</td>
</tr>
<tr>
<td>5</td>
<td>Threshold-based formula in conjunction with maximum allowable gain without feedback</td>
<td>Measure in-situ thresholds for a low, middle, and high band</td>
<td>Fixed frequency, levels vary as with audiometric threshold testing</td>
<td>Composite tones, number and frequency of tones vary with desired filter cutoffs</td>
</tr>
</tbody>
</table>

NBN, narrow band noise.
Figure 2–4. Hearing thresholds of the test ear for subject B.Y. (right ear) and J.A. (left ear).

Figure 2–5. Frequency-specific output levels [dB sound pressure level (SPL) at the eardrum], for subjects B.Y. and J.A., when using five different hearing aids. Hearing aid outputs were set following manufacturer-specific suggested in-situ procedures.
subjects. Recall that the overall RMS level refers to the total signal power across all frequencies, and a strong relationship between RMS level and loudness exists (e.g., Bentler et al, 1990). Peak outputs were obtained by taking the larger of (1) the maximum output off of the graph resulting from the pure-tone sweep or (2) the output of an individually presented pure tone. As expected, peak outputs for all hearing aids and both subjects occurred in the 2000- to 3000-Hz frequency regions. When plotted in this fashion it is apparent that aid 5 provides the greatest peak output (11 and 8 dB more than aids 3 and 4, for subjects B.Y. and J.A., respectively). The remaining differences across aids are smaller and vary to some degree between subjects.

The amount of time required to complete the in-situ procedure varied across procedures and subjects. Most procedures, however, took at least 5 minutes to complete (monaurally), and some took over 10 minutes. We were curious to see how this investment in time affected the prescribed maximum output. Figure 2–7 shows the difference between the frequency-specific prescribed maximum outputs (as estimated from each manufacturer’s fitting screen) based on custom in-situ measures and the default maximum output based solely on the audiogram.

The upper and lower solid lines represent the ±3-dB range. A positive number means the maximum output based on custom in-situ measures was greater than the output predicted based on the audiogram alone.

Figure 2–6. Peak output and overall root mean square (RMS) levels (dB SPL at the eardrum), for subjects B.Y. and J.A., when using five different hearing aids. Hearing aid outputs were based on manufacturer-specific suggested in-situ procedures.
whereas a negative number indicates the output was greater based on the audiogram alone measures. We have plotted the results from both subjects on the one graph to demonstrate that differences in the prescribed output based on the audiogram alone and those based on custom in-situ measures are, for the most part, small. Of the 50 data points (five hearing aids × five frequencies × two subjects) shown, only five data points fall outside the ±3-dB range. In other words, at least for these two subjects, the additional information provided by in-situ measures has only a minimal impact on manufacturers’ prescriptions for maximum output.

There are a couple of probable reasons for the good agreement between the manufacturers’ default prescription for maximum output and the output based on in-situ measures. First, our subjects did not appear to be “sound sensitive” or “sound addicts.” Despite the wide range of RESRs obtained with the various hearing aids, neither subject complained that the RESR measures were “too loud” with any hearing aid. Our subjects, therefore, probably fit into the 70% of the population for which threshold-based techniques provide an adequate estimate of LDL.

Assessing Loudness Using Assessment Scales
To this point, we have discussed various methods of assessing loudness discomfort. In all cases, signals were presented to the patient and a loudness rating was obtained. An alternative method of identifying a patient who might be either “sound sensitive” or a “sound addict” is to conduct a self-assessment inventory, which contains questions regarding loudness. Patients provide ratings for different sounds based on their memory of the sound, not based on actual sound samples. Although this approach does not provide specific dB levels for discomfort, it can be useful in identifying individuals who might require more extensive adjustment—particularly the sound sensitive patient. One could argue that these inventories would provide a more realistic view of the patient’s loudness perceptions than clinically measured LDLs, as the items relate to sounds that the patient hears in the real world. There are two different self-assessment scales that easily can be used for this testing, the Abbreviated Profile of Hearing Aid Benefit (APHAB) (Cox and Alexander, 1995) and the Profile of Aided Loudness (PAL) (Mueller and Palmer, 1998; Palmer et al, 1999).

APHAB
The APHAB consists of four separate subscales with six questions for each scale. One of the four subscales is termed “aversiveness,” which relates how bothersome the patient finds various environment sounds. Individuals rate these six questions on a
seven-point scale ranging from always (99%) to never (1%). Examples of the questions are as follows:

- Unexpected sounds, like a smoke detector or alarm bell, are uncomfortable.
- Traffic noises are too loud.
- The sounds of running water, such as a toilet or shower, are uncomfortably loud.
- The sounds of construction work are uncomfortably loud.
- The sound of a fire engine siren close by is so loud that I need to cover my ears.
- The sound of screeching tires is uncomfortably loud.

Cox (1997) has published norms for the APHAB for elderly people with few or no hearing problems. Column AV in Figure 2–8 is the aversiveness subscale. Because average loudness perceptions for loud sounds do not change significantly until hearing loss exceeds 55 to 60 dB, these norms can be used as general guidelines to identify patients who might be sound sensitive. For example, the unaided APHAB results for a sample patient with a mild-to-moderate hearing loss are plotted on Figure 2–8 (darkened circles). Observe that this patient’s "problems" for the aversiveness scale are near the 95th percentile for elderly individuals, and exceed the 95th percentile for young people with normal hearing. These findings would suggest that this patient is "sound sensitive."

Palm

A second self-assessment scale that can be used to obtain general information regarding a patient’s loudness perceptions is the PAL. The PAL is a 12-question inventory designed to assess the patient’s loudness perceptions of everyday sounds (four questions each for different soft, average, and loud sounds). Loudness perceptions are rated based on the Cox Contour Test (Cox 1995), which was discussed earlier in this chapter. Normative data for the PAL is shown in Figure 2–9, from Palmer and Mueller (2000). Observe that for the items in the "loud sounds" category, the majority of ratings are in the no. 6 category ("loud, but okay"). If a patient considered all or most of these sounds to be "uncomfortably loud" (no. 7), more attention should be focused on the output limiting of the hearing aids.

Predicting LDL from the Auditory Threshold

The final option we will discuss is not to measure LDLs or loudness ratings at all, but rather to predict loudness levels from the pure-tone threshold. Despite the variability displayed in Figures 2–1 and 2–2, a long-standing debate exists regarding the clinical efficiency of measuring subject-specific LDLs (e.g., Hawkins and Schum, 1991). You might think it a bit strange that anyone would question the need to measure subject-specific LDLs after seeing the large range of LDLs for subjects with the same hearing loss. There are a few things to consider, how-
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Figure 2-9. Normative data for the 12 items of the Profile of Aided Loudness (PAL).
ever, before deciding that threshold-based techniques are not a viable means of estimating loudness discomfort. As was discussed earlier, subject-specific LDLs vary based on test-retest differences and based on the type of measurement procedure. It is clear, then, that the LDL for a subject is not a single value. Like the estimate provided by threshold prediction methods, subject-specific LDL measurements obtained during a clinical visit provide only an estimate of perceived discomfort levels. Verification that amplified signals remain below LDL is required at the hearing aid fitting and at follow-up visits, whether the original estimate is determined using subject-specific LDL measures or a threshold-based procedure.

If we accept the notion that an individual’s LDL is actually a range, as opposed to a specific level, then the idea of using a threshold-based technique to place us in “the ballpark” seems more reasonable. Hawkins and Schum (1991) stated that, for many subjects, measuring LDLs prior to the hearing aid fitting was not required given the following: (1) aided loudness measurements were made at the hearing aid fitting to allow for setting the maximum output; and (2) the prescribed hearing aid allowed for adjustment of maximum output over a relatively wide range (e.g. 15 dB). In the Hawkins and Schum article (1991), Schum proposed that mean functions relating LDL to degree of hearing loss, in other words threshold-based formulas, could be used to determine an approximate maximum output. Then aided loudness judgments could be used to verify and fine-tune the maximum output at the hearing aid fitting. Hawkins and Schum stressed the importance of having considerable control over the hearing aid maximum output. If this control is not available, then measurements of subject-specific LDLs may provide a better estimate of real-world discomfort levels.

There certainly are obvious advantages to using a threshold-based technique to predict the hearing aid’s maximum output. The foremost advantage is the reduction in test time spent at the prefitting. Clinical techniques for determining subject-specific LDLs generally require between 10 and 20 minutes to complete (e.g., Cox et al, 1997). In addition, some subjects may have difficulty completing the task reliably. Two important questions, however, immediately come to mind when considering the use of threshold-based techniques for determining the maximum output of hearing aids. First, can an individual’s LDL be accurately predicted using threshold-based procedures? Second, what types of procedures are currently available for predicting OSPL90 based on threshold? And, related to the second question, because there are several prescriptive methods available, how do these threshold-based procedures differ in their predictions of maximum output?

**Auditory Threshold/LDL Relationship**

An obvious first question to ask when considering using a threshold-based technique to determine maximum hearing aid output is, Does a relationship exist between the degree of hearing loss and the patient’s LDL? In general, research suggests that the answer is yes. Although large between-subject variability in LDLs as a function of hearing loss is common, group data do show a relationship between LDLs and hearing loss (Kamm et al, 1978; Shapiro, 1979; Pascoe, 1988; Dillon and Storey, 1998; Bentler and Cooley, 2001). Hearing loss appears to have little impact on the LDL for hearing loss less than 40 to 60 dB HL. For hearing loss above 60 dB, however, LDL tends to increase with increasing hearing loss (Kamm et al, 1978; Shapiro, 1979; Pascoe, 1988; Dillon and Storey, 1998; Bentler and Cooley, 2001). This general trend is shown in both Figure 2–1 and Figure 2–2. The largest data sample is shown in Figure 2–2, which is based on data from over 600 ears, showing LDLs collapsed for several frequencies (500, 1000, 2000, 3000, and 4000 Hz) as a function of hearing loss.

Dillon and Storey (1998) pooled LDL data from several studies in an attempt to derive a threshold-based formula for predicting maximum output for hearing aids. Using this pooled data they reported that the rela-
tionship between LDL and hearing loss was described well using two linear functions, with a breakpoint at hearing loss of 60 dB HL. The relationship they describe is illustrated in Figure 2–10, which shows the three-frequency average (500, 1000, and 2000 Hz) LDL as a function of hearing loss for subjects from several different studies.

Like the data from Bentler and Cooley (2001), the large between-subject variability in LDL measures is easily seen in Figure 2–10. Despite this large variability, in a companion paper designed to validate their predictive formula, Dillon and colleagues reported that using subject-specific LDLs to predict “optimal” output “increased accuracy by such a small amount that it was not considered worthwhile” (Storey et al, 1998). The theoretical method for predicting the “optimal or prescribed” OSPL described by Dillon and Storey (1998) differs from some other threshold-based (and subject-specific) procedures. These authors argue that although hearing aid output should not exceed LDL, it need not be set just at or just below LDL either.

An estimated optimal output is considered midway between the maximum output required to limit loudness discomfort and the minimum output needed to avoid saturation of the hearing aid. Dillon and Storey (1998) defined the onset of saturation “to be when saturation caused the long-term RMS level of the output signal to be 2 dB below the value that would occur in the absence of saturation.” Using this definition the authors derived a formula to predict the minimum output required such that the output of a hearing aid, providing gain based on National Acoustics Laboratories “revised-profound” (NAL-RP) prescription to a 75-dB SPL speech input, would be high enough to cause only minimal saturation of the hearing aid, at least for most subjects with three-frequency average (500, 1000, and 2000 Hz) hearing loss of 60 dB HL or less. Additionally the hearing aid output would be low enough to prevent loudness discomfort.

For subjects with three-frequency averages greater than 70 dB HL, the formula actually predicts that the minimum output required to avoid saturation is greater than the

![Figure 2-10. Three-frequency (500, 1000, and 2000 Hz) average (3FA) LDL as a function of 3FA hearing loss. Data set is based on results from five different studies. (From Dillon H, Storey L. The National Acoustic Laboratories' procedure for selecting the saturation sound pressure level of hearing aids: theoretical derivation. Ear Hear 1998;19:255–266. Reprinted with permission from Lippincott Williams & Wilkins.)](image-url)
maximum output. Despite this, the authors suggest the optimum output should remain at the midpoint between the two levels. They use the following as support for this recommendation. First, given the effect of various factors (e.g., instructional set and psychophysical procedure) on LDL measures, it is not clear that an LDL measured using a specific method will exactly match the optimum hearing aid output. Second, as stated earlier, subject-specific LDLs provide only an estimate of a real-world LDL. Finally, it is possible that setting the OSPL too low, resulting in excessive saturation with conversational inputs, may be more objectionable than exceeding the measured LDL by a small amount. Finally, the authors make a good point when they suggest that at the time of the hearing aid fitting, it is easier for a clinician to determine when hearing aid output is too high than when it is too low.

In addition to the method proposed by Dillon and Storey (1998), clinicians today can choose from several threshold-based formulas for predicting LDL and maximum hearing aid output. Many of the formulas used for prescribing hearing aid gain also prescribe maximum output. Modern prescriptive methods that allow for predicting maximum output include NAL nonlinear version 1 (NL1) (Dillon et al., 1999; Byrne et al., 2001), visual input-output locator algorithm (VIOLA) for Windows version 1.0 (Cox and Flamme, 1998; Cox 1999), desired sensation level (DSL) version 4.1 (Cornelisse et al., 1995; Seewald et al 1997), and FIG6 (Killion and Fikret-Pasa, 1993; Gitles and Niquette, 1995; Etymotic Research, 1997). Given the large number of options currently available for predicting LDL and the hearing aid’s maximum output, a valid question to ask is, Do these formulas differ significantly in their predictions of maximum output?

Hawkins et al (1992) asked this exact question concerning earlier methods for predicting maximum output. These authors examined six different procedures for predicting maximum output. Two procedures were threshold-based whereas the other four involved subject-specific measurements of LDL. The authors found substantial differences in the predicted maximum outputs for some subjects and techniques. The largest difference occurred at 2000 Hz with mean prescribed saturation SPL (SSPL90) values ranging from 115-dB SPL for the DSL procedure (Seewald and Ross, 1988) to 102-dB SPL for the prescription of gain and output (POGO) procedure (McCandless and Lyregaard, 1983). When averaged across three frequencies (500, 1000, and 2000 Hz) the largest difference between methods was 8 dB. In this study, the techniques for measuring LDL varied between procedures; therefore, some differences were likely due to procedural effects, such as psychophysical method, instruction sets, and stimulus parameters.

The procedural effects described above are absent when using threshold-based procedures to predict maximum output. Therefore, we might expect that the variance between modern threshold-based procedures would be smaller than that observed in the Hawkins et al (1992) study. To examine this, we used the software distributed by the developers of four prescriptive formulas:

- DSL version 4.1 (Seewald et al, 1997)
- FIG6 for Windows version 1.0 (Etymotic Research, 1997)
- NAL-NL1 version 1.01 (Dillon et al, 1999)
- VIOLA for Windows version 1.0 (1999)

Using the above software, we determined the maximum output for three hypothetical adult hearing-impaired subjects. In addition, the formula provided by Dillon and Storey (1998) was used to predict the maximum three-frequency average (500, 1000, and 2000 Hz) output for each hypothetical subject. The three audiograms were a mild-moderate sloping loss, severe sloping loss, and a moderate flat loss.

Differences in prescribed output between procedures did not vary substantially as a function of degree or slope of hearing loss. Therefore, results for the prescribed outputs were averaged. Figure 2–11 shows the average prescribed outputs for each of the four
procedures that provide frequency-specific targets.

The largest differences between procedures occurred at the frequency extremes (e.g., 250 and 4000 Hz). A 16-dB difference between the VIOLA target and the FIG6 target occurred at 250 Hz. At the other frequency extreme, a 13-dB difference was observed at 4000 Hz between the predicted outputs when comparing the DSL and NAL-NL1 procedures. Between these extremes (e.g., 500 to 2000 Hz), however, differences between procedures do not vary substantially. To simplify comparisons across procedures, and to include the Dillon and Storey method, prescribed maximum outputs were averaged across 500, 1000, and 2000 Hz to create a single three-frequency average (3FA) maximum output for each procedure. These results are shown graphically in Figure 2–12.

For this three-frequency average, FIG6 and VIOLA consistently prescribed the least output whereas NAL-NL1, DSL, and the Dillon and Storey procedures consistently prescribed the highest output. Differences in the mean prescribed maximum output were small for the NAL-NL1, DSL, and Dillon and Storey procedures. The largest mean difference in prescribed output (almost 8 dB) occurred when comparing maximum output targets derived using the FIG6 and NAL-NL1 prescriptive methods. This large difference, however, must be viewed with some caution. FIG6 differs somewhat from the other prescriptive procedures used in this comparison. The NAL-NL1, VIOLA, DSL, and Dillon and Storey (1998) procedures all provide prescriptions for the maximum hearing aid output regardless of input level. In contrast, FIG6 provides 2-cc coupler gain values assuming a 95-dB SPL input. The maximum output predictions based on FIG6 calculations assume that these gain plus input values will result in the aid reaching its maximum output. For the hearing losses described here, the compression parameters prescribed by the NAL-NL1, DSL,
and VIOLA procedures do not consistently result in reaching maximum output with a 95-dB SPL input (as used by FIG6). A more reasonable comparison between procedures is seen in Figure 2–13. This figure shows the prescribed output for each procedure for a 95-dB SPL input. The Dillon and Storey (1998) procedure does not provide input/output (I/O) functions; therefore, the value for this procedure is unchanged from that in Figure 2–12.

The average differences in these threshold-based procedures are surprisingly similar to those observed by Hawkins et al (1991). Although most differences between the current procedures are small, we observed frequency-specific differences of up to 16 dB. Unfortunately, little research has been conducted to determine which one of these methods, if any, provides the most appropriate output setting for hearing aid use in the real world.

In an attempt to validate their prescriptive technique, Storey et al (1998) determined a range of acceptable outputs for 34 hearing-impaired subjects and compared theoretical predictions using the Dillon and Storey method to the midpoint of this measured range. In the laboratory setting they reported 86% of subjects’ theoretical predictions fell within their acceptable range. In addition, a field experiment was performed in which subjects wore multimemory hearing aids with different output settings saved in the memory. Multiple output settings were compared over a number of weeks until a range of acceptable outputs as used in the real world was determined. The authors reported that the predicted optimal output was within 2 dB of the acceptable range for only about 63% of their subjects. A somewhat surprising finding of this study was that including frequency-specific mea-

**Figure 2–12.** 3FA (500, 1000, and 2000 Hz) prescribed maximum output for five threshold-based methods. Outputs are averaged across three hypothetical hearing losses described in the text.
asures of individual LDLs in the formula for predicting optimal output did not significantly improve predictive accuracy. In other words, using the measured LDL, rather than the predicted LDL, as the top end of the range didn’t change the range enough to substantially improve their predictions. This reinforces the need for clinician control of maximum output and aided verification of LDL during the hearing aid fitting. It appears likely that a substantial proportion of hearing-impaired persons will require some modification to the initial setting of maximum output whether it is selected based on individual LDLs or by a threshold-based procedure. Clinical verification techniques are discussed later in this chapter.

When predicting the LDL from the hearing threshold there also are other factors that must be considered. We will review some of the most important variables.

Differences between Young Adults and the Elderly

Although limited work has been done examining LDL age effects between children and adults, even less is known about the impact of aging on LDL in the adult and elderly populations. We know that the aging process can impact some auditory processes, such as speech understanding in noise and temporal processing (e.g., Studebaker et al, 1997). The impact of the adult aging process on LDL, however, has received only scant research attention. Recently, Bentler and Cooley (2001) examined several factors, includ-

Figure 2-13. 3FA (500, 1000, and 2000 Hz) prescribed output for a 95-dB input for five threshold-based methods. Outputs are averaged across three hypothetical hearing losses as in Figure 2-11. The Dillon and Storey method only calculates optimal prescribed output; therefore, the value is the same as in Figure 2-12.
I

ing age, in an attempt to identify factors that may help predict LDL. Combining the results from five previous investigations they pooled LDL data from over 400 subjects (710 ears) ranging in age from 11 to 97 years old. Multiple regression analysis using almost 2000 LDL responses, across multiple frequencies, was performed to determine if LDLs varied significantly based on age. The authors believed that the use of such a large data set should show obvious trends, in terms of age effects, if they truly existed. Results of the multiple regression analysis, however, indicated that only auditory threshold and measurement method played a role in determining an individual’s LDL. The LDL was not significantly affected by age.

In addition to the behavioral measure of LDL, researchers also have compared subjects’ ratings for loud sounds for different age groups in the real world. Cox (1997) reported on the percent of problems for the aversiveness scale (annoying loud noises) of the APHAB for two different groups: “young normal hearing” and “elderly with few or no hearing problems” (Figure 2–8; AV = Aversiveness subscale). Her results, described in percentiles, and illustrated in Figure 2–8, show that for younger subjects, the 20th percentile was 10% of problems, the 50th percentile was 20% of problems, and the 80th percentile was 42% of problems. For the elderly group, the corresponding percent of problems for the same percentiles were 2%, 10%, and 40% respectively—somewhat fewer problems than the younger group, but in general, similar findings.

When the normative data for the PAL was established, the effects of age were also studied (Mueller and Palmer, 1998; Palmer et al., 1999). Using the descriptive anchors of the Cox Contour Test (Cox, 1995), groups of subjects from seven age decades ranging from the 20s to the 80s rated their perceived loudness levels of several loud environmental sounds. No significant difference in loudness ratings for the different age groups was observed. Based on the results of these subjective real-world loudness ratings and on the findings of Bentler and Cooley (2001), we can conclude that corrections to adult LDLs based on age are not needed.

Differences between Children and Adults

The amplification needs of young children with hearing loss who are learning speech and language may be very different from those of postlingual adults experiencing late-onset hearing loss (Nittrouer and Boothroyd, 1990). In addition, the well-known differences in RECDs between adults and children can have significant effects on the desired 2-cc coupler amplification requirements for children (Pediatric Working Group, 1996). What is not clear, however, is whether subjectively measured LDLs, when differences in RECD values are accounted for, are substantially different in adults and children.

Surprisingly, very little work has been done investigating the differences in suprathreshold measures of loudness between children and adults. One reason for the limited work is that reliable measures of LDL are notoriously difficult to obtain from young children (e.g., Macpherson et al., 1991; Ellis and Wynne, 1999). Some authors, however, have reported modifications to LDL measures that improve the reliability with children and thus allow for comparisons of LDLs between adults and children (e.g., Kawell et al., 1988; Stuart et al., 1991; Serpanos and Gravel, 2000). For example, shown in Figure 2–14 from Kawell et al. (1988) are the LDLs of a group of 20 children aged 7 to 14 years with hearing impairment. As shown, the average LDLs of these children are essentially identical to those of adults with similar hearing loss.

Serpanos and Gravel (2000) found that the slopes of loudness growth functions (LGFs), as measured using a cross-modality matching technique between line length and loudness, were similar for adults and children (4 to 12 years old) with normal hearing. Individual points on a given LGF, however, tended to be perceived as louder by the children even when differences in RECD values were accounted for. This difference in loudness perception was largest, however, at low
to medium levels. Differences in loudness perception between children and adults at high levels (e.g., LDL) were minimal.

Ellis and Wynne (1999) reported similar findings in a study that examined differences in loudness growth between children and adults when using a modified version of the Loudness Growth in Octave Bands (LGOB) test (Allen et al, 1990). These authors found that children (7 to 12 years old) consistently judged narrow-band noises as “too loud” at levels lower than adults and sounds as “too soft” at levels higher than adults. The findings, however, were referenced to HL rather than ear canal SPL, which could have accounted for this difference, as individual differences in ear canal SPL were not accounted for. In general, it appears that when referenced to ear canal SPL, the LDLs for adults and children are similar.

**Gender Effects**

Multiple studies have shown that some auditory processes are affected by gender. For example, changes in auditory thresholds with age, auditory interhemispheric transfer, and speech understanding in noise and reverberation have all been shown to have gender effects (Gates et al, 1990; Helfer and Wilber, 1990; Dubno et al, 1997; Bellis and Wilber, 2001). A reasonable question to ask then is, Are LDLs also affected by gender?

The previously mentioned study by Bentler and Cooley (2001) also included gender as a predictor variable in their regression analysis of LDL data obtained from over 700 ears. Of the almost 2000 data points analyzed, 1531 were from female subjects and 466 were from male subjects. Based on the results of their regression analysis, however, it appears gender, like the adult aging process, does not significantly affect measures of loudness discomfort. This finding is in agreement with the real-world subjective results reported by Mueller and Palmer (1998), which also showed no significant gender differences for loudness ratings for several different loud environmental sounds.

**Binaural Fittings**

Predictive methods for the LDL are based on data from monaural measurements. Should corrections be made to the OSPL90 for a binaural fitting? In other words, how does loudness summation influence the bilateral LDL versus the monaural one? This is not only an issue when LDLs are predicted, but also applies when LDLs are measured. We know
from loudness summation research that the summation effect for suprathreshold signals can be 6 to 8 dB or more (e.g., Reynolds and Stevens, 1960). It is interesting to observe, however, that when summation is measured in the sound field rather than under earphones the summation effect is much less. For example, Hawkins (1986) and Hawkins et al. (1987) report that there is very little difference between monaural and binaural aided LDLs (a summation of only 0.9 dB). Recently, Cox and Gray (2001) report very similar findings for the #7 rating of the Cox Contour Test. It appears, therefore, that corrections to the desired OSPL90 or RESR are not necessary for binaural fittings. See Mueller and Bentler, 2002, for review. This point also relates to aided loudness judgments, which we will discuss at the end of this chapter.

Conductive Hearing Loss

Nearly all research related to predicting the LDL has been directed toward patients with hearing loss of cochlear etiology. But what about patients with conductive or mixed losses? Does the conductive loss simply act like an earplug and the LDL goes up on a 1:1 basis? What about the absent acoustic reflex? Although several prescriptive methods have gain corrections for a conductive loss, little has been published regarding the effects of a conductive loss on the desired OSPL90 or RESR. An exception is the NAL-SSPL procedure, which calls for adding 87.5% of the conductive loss to the desired output level (Dillon, 2001). (Even Harvey Dillon, an engineer, does admit that using 90% is probably okay)

Using the LDL for the Hearing Aid Fitting

There are two primary reasons for determining a patient’s LDL: (1) to determine an OSPL90 setting to assist with hearing aid selection or prefitting adjustment, and (2) to provide a maximum output target (desired RESR) when the hearing aid is fitted. We have already talked about two different methods of selecting the hearing aid’s 2-cc coupler maximum output: through in-situ methods, and by using a threshold-based approach in conjunction with a prescriptive fitting method (e.g., VIOLA, FIG6, DSL 4.1 or NAL-NL1). But what if you choose to measure the patient’s LDL? How do you now convert these HL levels into 2-cc coupler values so that an appropriate output can be selected? There are two primary methods that can be used.

LDL Plus Average Corrections

As we discussed earlier, the RETSPL is the difference between the audiometer dial setting and the output in a coupler—in the case of insert earphones, this is a 2-cc coupler, the same volume used to express hearing aid output. Hence, by adding the RETSPL to the patient’s audiometric LDL in HL, we have a reasonable estimate of the desired maximum output in the 2-cc coupler. Table 2-4, taken from ANSI 3.6–1996, provides RETSPLs for insert earphones.

If supra-aural earphones are used, the correction requires somewhat more thought, as the RETSPL is for a 6-cc, not a 2-cc, coupler, and therefore the ANSI RETSPLs are not a direct correction for the values of interest. Hawkins (1992) has made the necessary cor-
rections, however, and Table 2–5 shows the values that can be used to convert from dB HL to 2-cc coupler SPL for different types of supra-aural TDH earphones.

It also would be possible to use a prescriptive method like the DSL version 4.1a to establish the desired OSPL90 values. As mentioned earlier, the DSL 4.1a method will predict OSPL90 from pure-tone threshold, but it also is possible to enter the patient’s LDLs (ULCs are recommended) and the type of earphone that was used, and the desired OSPL90 will be calculated. Additionally, desired real-ear output targets (using an average REDD) will be provided.

**Using the RECD**

The RECD can be used to assist with output selection for adults, but it is most important with children. If we know a child’s RECD, we have information that will assist us in selecting the appropriate hearing aid output and predicting the output in the child’s ear (OSPL90 + RECD = Predicted RESR). The RECD is also a component of the REDD (RECD + RETSPL = REDD); therefore, knowing the RECD allows us to more accurately predict the LDL in terms of ear canal SPL. Moreover, if the RECD is known on the day of the fitting, the majority of the patient-specific hearing aid adjustments can be made in the 2-cc coupler, reducing the time and cooperation required from the child. The RECD of a child can be quite different from those of an adult. The RECDs for different ages are shown in Table 2–6.

Although the consensus is that the RECD is an important measure, there does not seem to be a standard test procedure. Also, different procedures lead to different results. For example, the signal can be delivered to the ear using the patient’s own hearing aid, an insert earphone coupled to the patients earmold, or an insert earmold coupled to a foam plug. The 2-cc coupler used for comparison could be either HA-1 or HA-2. In

<table>
<thead>
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<th>Frequency (Hz)</th>
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<th>TDH 49 and 50</th>
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<td>22</td>
</tr>
<tr>
<td>500</td>
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<td>0</td>
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<tr>
<td>6000</td>
<td>0</td>
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</table>


<table>
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<tr>
<th>Frequency (Hz)</th>
<th>HA-1 coupler</th>
<th>HA-2 coupler</th>
</tr>
</thead>
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</tr>
<tr>
<td>5000</td>
<td>22.4</td>
<td>17.8</td>
</tr>
</tbody>
</table>

*Adapted from Seewald et al, 1997.*

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**Table 2–5. Conversion Values from dB HL to dB SPL in a 2-cc Coupler; to Convert from HL to 2-cc Coupler, the Correction Factors Below are Added to the HL Values.**

<table>
<thead>
<tr>
<th>Frequency (Hz)</th>
<th>TDH-39</th>
<th>TDH 49 and 50</th>
</tr>
</thead>
<tbody>
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<tr>
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<td>0</td>
</tr>
<tr>
<td>6000</td>
<td>0</td>
<td>-2</td>
</tr>
</tbody>
</table>

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**Table 2–6. Real-Ear Coupler Differences (RECDs) for Different Ages**

<table>
<thead>
<tr>
<th>Frequency (Hz)</th>
<th>HA-1 coupler</th>
<th>HA-2 coupler</th>
</tr>
</thead>
<tbody>
<tr>
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<td>5.4</td>
<td>5.5</td>
</tr>
<tr>
<td>500</td>
<td>9.8</td>
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<td>1000</td>
<td>13.0</td>
<td>11.9</td>
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<tr>
<td>1500</td>
<td>14.4</td>
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<tr>
<td>2000</td>
<td>14.5</td>
<td>10.5</td>
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<tr>
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<td>17.8</td>
</tr>
<tr>
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<td>22.4</td>
<td>19.7</td>
</tr>
</tbody>
</table>
fact, if you wanted to push the RECD rules, you could deliver the signal to the ear with a TDH-series earphone, compare that to the output in a 6-cc coupler, and still call your findings an RECD. There are many factors that complicate the precise description of the RECD measurement, and some of them are as follows (Mueller, 2001b):

- If the measurement is conducted using the patient’s own hearing aids, MLEs become part of the RECD (assuming the testing is conducted in the sound field; if direct audio input is used, then the MLE is not included). If a given prescriptive fitting approach (such as DSL or NAL-NL1) or a manufacturers’ software already includes average MLEs, and they allow you to enter the patient’s RECD, then it’s possible that the MLE will be added twice. The coupler response for flat insertion gain (CORFIG) is calculated by the formula \( \text{REUG} = \text{RECD} + \text{MLE} \), which assumes that the MLE is not part of the RECD.

- The MLEs for hearing aids tested using a moderate input level are not the same as the MLEs when the hearing aids are in saturation. One of the uses of the RECD is to predict the output of hearing aids when they are in saturation, yet the hearing aid real-ear aided response (REAR) measurement usually is conducted using a 60-dB SPL input.

- If the patient is fitted with a WDRC instrument, then it is necessary to consider the effects of WDRC on the MLE. That is, if a 60-dB SPL signal at 4000 Hz becomes a 70-dB SPL input because of an MLE of 10 dB (CIC fitting), the increase in output will not be proportional if the input signal is compressed (e.g., if the compression ratio is 2:1, the MLE will now only yield a 5-dB change in output). In essence, the MLE itself doesn’t change, but its influence on the resulting ear canal SPL does.

- If an HA-1 coupler is used, the RECD will be different than if an HA-2 coupler is used (see Ricketts and Bentler, 1996 for review).

- If an inset earphone is used, the output impedance effects need to be the same as for the patient’s hearing aid.

- If a foam plug is used, the placement depth in the ear canal should approximate the depth of the patient’s hearing aid or earmold.

Although there does not appear to be a standard protocol for conducting the RECD, we recommend using the procedures from the research group at the National Centre for Audiology, University of Western Ontario (UWO), which has conducted extensive research on the clinical applications of the RECD (see review by Seewald, 2000). This method was first reported in 1994, when Moodie et al (1994) described an RECD procedure that utilized insert earphones, an HA-2 coupler, and the child’s own earmold (Fig. 2–15). In general, the procedure they described is as follows:

- Conduct appropriate calibration and configure the software of the probe-microphone (probe-mic) equipment for RECD measurements (this procedure will vary depending on manufacturer of the probe-mic equipment).

- Connect the insert earphone to the loudspeaker output terminal of the probe-mic equipment using a matching phone plug.

- Thread the probe tube through the calibrator adapter plug so that the tip of the tube extends no more than 3 mm above the surface of the plug.

- Place the calibrator adapter plug into the microphone port of the HA-2 2-cc coupler.

- Connect the probe tube to the probe microphone.

- Connect the insert earphone tip to the tubing of the HA-2 2-cc coupler.

- Deliver a speech-weighted signal of 50-dB SPL, and record the coupler output. Save this as the “unaided” measure (REUG).

- Remove the probe-microphone apparatus from the coupler and place the tube in the patient’s ear with the probe tip at the appropriate depth.
54 Strategies for Selecting and Verifying Hearing Aid Fittings

- Place the patient’s earmold in the ear.
- Attach the insert earphone tip to the tubing of the earmold (the length of the earmold tubing should be appropriate for BTE hearing aid use).
- Deliver a speech-weighted signal of 50-dB SPL, and record the real-ear output. Save this as the “aided” measure (REAR).
- Either automatically or through software command, the probe-mic equipment will calculate a real-ear insertion gain (REIG), which in this case is not an REIG, but an RECD. What will be displayed is the coupler output subtracted from the real-ear output for all test frequencies.
- Positive values are added to the OSPL90 to predict the RESR, and subtracted from ear canal SPL targets to determine appropriate 2-cc coupler output.

The preceding protocol is unique to a specific probe-mic system, and does vary somewhat when other equipment is used. The UWO group has published guidelines for RECD measurement with different types of probe-mic equipment (Seewald et al, 1997). This RECD procedure has been used successfully for several years, and has been found to be a highly accurate predictor of real-ear output. In recent research using this approach, Seewald et al (1999) compared the predicted REAG and RESR to the respective measured values for 14 children fitted with BTE hearing aids (head diffraction and MLE effects were included for the REAG prediction). Their results showed a 95% confidence interval of ±2.3 dB for the REAG and an average error range of 4.4 dB for the RESR.

With children, the RECD is most commonly used to select OSPL90 based on a real-ear target, or to predict the maximum output in the real ear based on the OSPL90 2-cc coupler results. If the goal was to select the appropriate 2-cc output based on the patient’s LDLs, then it would be necessary to measure the LDLs in real-ear SPL (a relatively easy task with probe-mic equipment; see the procedures used by Valente et al, 1997). The RECD would then be subtracted from the ear canal SPL LDL values to obtain desired OSPL90.

Because today’s hearing aids are very adjustable, one approach that is used by some audiologists is to disregard the patient’s pre-

dicted or measured LDL and simply order the hearing aid with the maximum OSPL90. The assumption is that during the fitting process, the maximum output can be adjusted to fit the patient’s needs. There are two potential problems with this approach. Let’s assume that a digital custom in-the-ear (ITE) product from a given manufacturer is available in four different maximum outputs ranging from 108- to 128-dB SPL. Our patient needs the 108-dB SPL model, but the 128-dB SPL model is ordered. On the day of the fitting, the automatic gain control for output (AGCo) is reduced to 108-dB SPL, and the patient does not experience loudness discomfort. But the patient now owns a hearing aid with a larger receiver than necessary, which probably means more battery drain. More importantly, the noise floor of the hearing aid (determined by the dynamic range) is 20 dB louder than necessary, and in fact may be annoying if the dynamic range is not large. Because LDLs change very little with increasing hearing loss, and because there is little or no acclimatization for LDLs, there seems to be no reason to order OSPL90s higher than what is required to maximize headroom.

When selecting the maximum output for hearing aids, or the amount of gain desired for loud inputs, several other factors also must be considered.

**Clinical LDLs and Real-World Levels of Discomfort**

A fundamental assumption is that the clinically measured LDL provides a reliable estimate of aided real-world levels of discomfort (i.e., external validity). Filion and Margolis (1992) were one of the first to address this issue empirically. Somewhat surprisingly, they reported that clinically measured LDLs substantially underestimated real-world judgments of loudness discomfort. To compare real world perception of loudness discomfort to clinically measured LDLs, these authors went into two environments (a nightclub and a manufacturing plant) frequented by study subjects and actually measured the SPLs over a 1-hour period. They also made audio recordings in both these environments that were used later as stimuli for LDL measurements. Subjects also used a questionnaire to estimate the percentage of time that sound levels in the environment they frequented (i.e., the nightclub or the manufacturing plant) were uncomfortably loud. The authors then compared the percent of time that the measured SPLs in an environment actually exceeded a subject’s clinically measured LDL (using recordings from the environment) to the percent of time subjects perceived the environment as too loud.

Results for six of the seven subjects listening in the nightclub setting showed that the actual SPLs in the nightclub would have exceeded their clinically measured LDLs over 95% of the time. Based on questionnaire responses, however, five of the seven subjects reported levels in the nightclub exceeded their LDLs less than 10% of the time. Unfortunately, the length of time between each subject’s exposure to these settings and completion of the questionnaire varied from 5 days to 4 weeks. Despite this delay it seems likely that the clinical judgments of LDL in this study substantially underestimated these subjects’ real-world discomfort levels.

The results of Dillon et al (1984), however, suggest there is a closer relationship between clinical testing and real-world judgments. These authors asked 15 hearing-impaired subjects (22 ears) if wearing their hearing aids caused them to experience sounds they felt were uncomfortably loud. Eleven of the 22 reports noted experiencing discomfort in everyday situations. They then measured the OSPL90 of the subjects’ hearing aids and found that for most subjects (nine of 11 ears) who reported loudness discomfort in everyday situations the aid’s OSPL90 exceeded their clinically measured LDL. In contrast, the measured OSPL90 of most hearing aids (eight of 11) used by subjects who did not report loudness discomfort in everyday situations did not exceed clinically measured LDLs.

Likewise, Munro and Patel (1998) reported a similar relationship between real-world loudness discomfort and the clinically mea-
sured LDLs. They used a probe-microphone technique to measure the SPL at the eardrum while determining frequency-specific LDLs for 20 adult subjects. The RESR of each subject's aid also was estimated by measuring RECDs and adding these values to OSPL90 values measured in a 2-cc coupler. Subjects also completed questionnaires in which they were asked to judge (from memory) the loudness of several short- and long-duration environmental stimuli (e.g., cutlery, door slamming, traffic, and wind noise). The authors found a significant positive correlation between subject ratings of loudness for long-duration sounds (e.g., traffic and wind noise) and differences between the RESR and clinically measured LDL. Subjects were more likely to report experiencing loudness discomfort in the real world when RESR values exceeded clinically measured LDLs. Eighty-three percent of subjects (10 of 12) whose OSPL90 exceeded their clinically measured LDL reported these long-duration sounds as too loud. Only one of eight subjects (12.5%), with an MPO set below their LDL, reported that traffic or wind noise was too loud. In contrast, no significant relationship was observed between loudness judgments of short-duration sounds (cutlery and door slamming) and RESR-LDL differences. Israelsson et al (1995) reported a similar finding for children (11 to 16 years old).

The studies cited above offer some support for the idea that clinical LDLs provide an estimate of real-world aided discomfort levels. There are other factors that can influence these findings, which we will discuss in the next section.

**OSPL90 Considerations Related to the Hearing Aid’s Signal Processing**

In the preceding section, we discussed how the measured LDL could be used to assist in the selection of the OSPL90 of the hearing aid. It is also important to consider the signal processing of the hearing aids that are fitted, as to some extent the processing can influence the maximum output that is selected, and also could influence verification measures.

**Peak Clipping versus Compression**

The maximum output of a hearing aid can be limited by using either peak clipping or compression—either automatic gain control for input (AGCi) or automatic gain control for output (AGCo). Peak clipping will introduce distortion to the signal, which usually is bothersome to the hearing aid user. For this reason, some form of compression limiting usually is recommended, especially for individuals with mild, moderate, and severe loss. When WDRC is employed, the manner in which output is limited becomes less important, as outputs rarely reach the saturation level (more on WDRC later).

As the hearing loss becomes more severe, ranging into the profound category, some research has indicated that peak clipping might be the preferred limiting method for some patients (e.g., Stelmachowicz et al, 1999). For one thing, if the patient is a “sound addict,” the output for a speech signal will typically be 5 dB or more for a peak clipper than for an AGCo instrument with the same OSPL90. The patient may tolerate the additional distortion if the greater loudness sensation is important.

**Saturation-Induced Distortion**

Related to our discussion of peak clipping and compression limiting are the effects that saturation-induced distortion has on the LDL. For example, Fortune and Preves (1992) found that under some conditions aided LDLs could be substantially reduced compared to LDLs obtained under headphones. Aided LDLs were obtained using four linear hearing aids with varying maximum output settings. The hearing aids with lower OSPL90s had more saturation-induced distortion than aids with higher OSPL90s. LDLs obtained with the lower output (and higher distortion) hearing aids were lower than LDLs obtained with the higher output (and lower distortion) aids.
All aided LDLs were lower than clinically measured LDLs obtained under headphones. The finding that LDL may be reduced in the presence of saturation-induced distortion is not surprising. Hawkins and Naidoo (1993) showed a strong subjective preference for output limiting compression over peak clipping, in terms of sound quality and clarity. The preference became stronger as saturation-induced distortion increased. These findings and those of Bentler et al (1990) suggest that clinically measured LDLs may overestimate real-world discomfort levels, particularly when listening through hearing aids that introduce distortion at high levels.

Wide Dynamic Range Compression (WDRC)

Nearly all of today’s hearing aids employ AGCi with a low kneepoint, commonly referred to as WDRC. The primary application of this type of compression is to restore normal loudness perceptions—that is, allowing the hearing-impaired individual to have the same perception for soft, average, and loud sounds as someone with normal hearing (without constant VCW adjustment). The use of WDRC is not typically thought of as a method of output limiting, as most instruments with WDRC also employ a second high-level AGCi, AGCo, or peak clipping. However, when the compression ratio (and/or kneepoint) of the WDRC instrument is selected by the fitting algorithm, the patient’s LDL (measured or predicted) is used to determine gain for loud sounds. Therefore, to some extent, the WDRC becomes the output limiter (although the output can be raised or lowered with the VC, if one is made available to the patient). If gain for loud inputs were indeed programmed correctly, other means of output limiting would seem to be unnecessary.

As hearing aids continue to become available with lower and lower compression kneepoints, the effects of the WDRC sometimes can lead to inappropriately low ear canal SPL for individuals who have average or higher-than-average LDLs. This might lead to the patient complaint that high-level sounds are “muffled.” Consider the following example. A WDRC single-channel hearing aid with a kneepoint of 40-dB SPL is programmed with a 2:1 compression ratio and 35 dB of gain for soft inputs (40-dB SPL) for a person with a moderate hearing loss. His LDLs are around 110 HL, and therefore the AGCo kneepoint is set to 115 SPL (re: 2-cc coupler). After using the new hearing aid for a few weeks, the patient states that loud sounds are not loud enough. It might be tempting to raise the kneepoint of the AGCo, to create more headroom, but first it’s important to consider that with the stated WDRC settings, an 80-dB SPL input is only receiving 15 dB of gain. Hence, it is the WDRC that is restricting the headroom, not the AGCo.

Channels of Compression

Historically, compression limiting has been accomplished using a single channel compressor (either AGCi or more commonly AGCo). Even many of today’s hearing aids with multichannel WDRC still employ a single channel compression limiter. From a “protective” standpoint, a single channel system works quite well. Assuming that the kneepoint of the compression system is adjusted correctly, this type of limiting will assure that loud input signals will not be uncomfortably loud for the patient. A limitation of a single-channel limiting system, however, is that the signal may be unnecessarily limited for some frequency regions. For example, an 80-dB SPL speech signal, amplified to 110-dB SPL, may have a real-ear peak at 1500 Hz. If this peak is at the AGCo kneepoint (set to the person’s LDL for this frequency), then a 90-dB SPL input of the same speech signal only would be 1 dB or so louder (assuming a 10:1 compression ratio). This small change in output for the 80- versus 90-dB SPL speech input would apply to all frequencies, even though at 500 Hz, the
output might be 10 dB below the patients LDL and does not require limiting. The advantage, therefore, of multichannel AGCo is that when the output is limited in one frequency region, it can continue to grow in other regions until the AGCo kneepoint for that specific channel is reached; recent digital hearing aids have employed this technology (Mueller, 2000). The patient can experience changes in loudness, without listening discomfort (Powers and Burton, 2000). By setting the kneepoint for each channel to correspond with the patient’s LDL, we are able to maximize the patient’s aided dynamic range. This is particularly helpful for patients with an unusually small unaided dynamic range.

When fitting hearing aids with multichannel AGCo, however, the effects on the real-ear maximum SPL must be considered, and adjustments made to the desired OSPL90. That is, if each independent channel is set to the patient’s LDL based on a narrow-band input, due to summation effects, and if a loud input was present simultaneously in several channels, the combined output likely would exceed the patient’s LDL. The summation effect cannot be predicted precisely as it will depend on the energy in each channel. Dillon (2001), however, provides correction factors that can be applied (e.g., reduction in the OSPL90 or desired RESR): 5 dB for two channels, 7 dB for three channels, and 9 dB for four channels. Mueller and Bentler (2002) discuss several different aspects of channel (power) summation.

**Time Constants and Signal Duration**

The attack time of the hearing aids compression circuit can have an influence on determining the appropriate output setting. Fortune and Scheller (2000) explored the relationship between signal duration, hearing aid compression parameters, and aided LDLs. They reported that LDLs increased substantially as signal duration decreased (1024 to 32 msec). LDLs for short-duration sounds (i.e., 32 msec) were approximately 10 to 15 dB higher than LDLs for longer duration sounds (1024 msec). Using steady-state, long-duration tones to measure LDLs clinically may result in an underestimate of real-world levels of discomfort for short-duration sounds. This finding is in agreement with that of Munro and Patel (1998). These authors found that subjects wearing hearing aids with OSPL90s that exceeded their clinically measured LDLs reported experiencing discomfort when listening to long-duration sounds such as traffic and wind noise. In contrast, these same subjects did not report discomfort due to short-duration sounds such as a door banging or cutlery. In other words, LDLs as measured clinically, utilizing longer duration stimuli (e.g., >200 msec), may only relate to real-world discomfort for longer duration sounds.

**Safety Considerations**

It certainly is possible that an individual can suffer permanent hearing loss through acoustic trauma, or continued exposure while using hearing aids. Although this has not been commonly documented, it may happen more often than we realize and is simply attributed to “progressive hearing loss.” The primary concern regarding safety is for children, who cannot recognize that the sound is too loud, or cannot operate the volume control (or do not have a volume control). But even with adults, as pointed out by Mueller and Bentler (1994), just because a signal is not uncomfortable to the listener does not mean it is not loud enough to cause a permanent noise-induced hearing loss. As shown in Figures 2–1 and 2–2, some adults with hearing loss of only 40 to 50 dB HL have LDLs of about 130-dB SPL (re: 2-cc coupler). Accounting for the RECD, if the hearing aid output was set to the patient’s LDL, the real-ear output easily would be 135-dB SPL or greater.

There are many adults with severe-to-profound hearing loss, however, who fall into the category of the “sound addicts” that we discussed earlier. That is, there seems to be no upper limit to what they consider loud enough. It is tempting to grant their wishes and fit these people with linear hearing aids with OSPL90s of 140-dB SPL (which could be
close to 150-dB SPL in the ear canal if a deep-fitted earmold is used). It is difficult to predict what the safe level is, as the degree of noise exposure is largely dependent on the input signal, the duration of the input signal, and the gain of the hearing aid. If overamplification is a concern, we recommend periodic monitoring for temporary threshold shift following several hours of hearing aid use (after establishing baseline thresholds following no exposure); see Macrae, 1994, 1995, for a complete review of this important topic.

Clinical Verification and Validation of Aided Loudness

Because of the many variables associated with the prediction and measurement of the LDL, and the many variables associated with converting from HL to SPL to 2-cc coupler to the real ear, we recommend that the maximum output of the hearing aid always be evaluated in the real ear. There are three different procedures that can be used, and verification with one does not assure that an acceptable finding will be obtained with the other two. Therefore, it’s probably best to include all three in your loudness verification protocol. The three methods of maximum output verification/validation that we recommend are the following:

- Probe-mic assessment of real-ear output for high-level signals.
- Aided loudness ratings for high-level speech and environmental noises.
- Self-assessment inventories related to real world-loudness judgments using hearing aids.

Probe-Mic Assessment

One method to verify the maximum output of the hearing aids at the time of the fitting is to conduct an RESR. The RESR normally is measured by delivering a signal (narrow-band) to the hearing aid at a 90-dB SPL input. The hearing aid is adjusted to the maximum setting that could be used by the patient; if the patient has a volume control, it is turned to full on, or just below feedback. The targets for the RESR can be obtained in one of three ways:

- At the time that the earphone LDL was conducted, the ear canal SPL was measured for the patient’s rating of “Loud, but okay” (which indirectly was a measurement of the patient’s REDD).
- Average REDD values are added to the measured or predicted LDL to create RESR targets (the measured RESR should fall slightly below these targets). This is included in the software of most probe-mic equipment.
- Target RESR values from prescriptive methods such as the DSL 4.1 or NAL-NL1 can be used. For example, see Table 2–7, from Dillon (2001) for target values.

The goal of the RESR testing is to determine if the output falls at or around the desired

| Table 2–7. Target Values for Real-Ear Saturation Response (RESR)* |
|------------------|----------------|----------------|----------------|----------------|
| Frequency (Hz)   | 250            | 500            | 1k             | 2k             | 4k             |
| HTL              |                |                |                |                |
| 0                | 95             | 96             | 95             | 98             | 100            |
| 5                | 95             | 97             | 96             | 100            | 101            |
| 10               | 96             | 97             | 98             | 101            | 102            |
| 15               | 96             | 98             | 99             | 102            | 103            |
| 20               | 96             | 99             | 101            | 104            | 104            |
| 25               | 97             | 101            | 102            | 105            | 106            |
| 30               | 97             | 102            | 104            | 107            | 107            |
| 35               | 98             | 103            | 105            | 108            | 108            |
| 40               | 99             | 105            | 107            | 109            | 109            |
| 45               | 100            | 106            | 108            | 111            | 110            |
| 50               | 101            | 108            | 110            | 112            | 112            |
| 55               | 103            | 109            | 111            | 113            | 113            |
| 60               | 104            | 110            | 113            | 115            | 114            |
| 65               | 107            | 114            | 115            | 117            | 117            |
| 70               | 111            | 117            | 118            | 120            | 119            |
| 75               | 115            | 120            | 121            | 122            | 122            |
| 80               | 118            | 124            | 123            | 125            | 124            |
| 85               | 122            | 127            | 126            | 128            | 127            |
| 90               | 125            | 131            | 128            | 130            | 129            |
| 95               | 129            | 134            | 131            | 133            | 132            |
| 100              | 132            | 137            | 134            | 135            | 135            |
| 105              | 136            | 141            | 136            | 138            | 137            |
| 110              | 139            | 144            | 139            | 141            | 140            |
| 115              | 143            | 147            | 142            | 143            | 142            |
| 120              | 147            | 151            | 144            | 146            | 145            |

Adapted from Dillon, 2001.
level. If LDL measures are used, the RESR should fall below these targets, as by definition, the target point is “uncomfortable” (this is why the DSL procedure uses ULC values). If the RESR is not appropriate, adjustments to the hearing aids are made (e.g., usually AGCo kneepoints, or WDRC kneepoints or ratios).

When digital hearing instruments are evaluated, the digital noise reduction should be turned off when RESR testing is conducted (see Mueller 2001a,b for review). Also, today it is common to use “speech like” signals when conducting probe-mic testing [e.g., composite speech noise or one of the International Collegium of Rehabilitative Audiologists (ICRA) signals]. In general, however, we do not recommend using broadband signals for real-ear maximum output measures, as when they are used, the RESR will be reduced significantly (Stelmachowicz et al, 1990). Although these signals probably do represent the RESR for many common real-world inputs, especially long-term speech, there also are narrow-band real-world sounds that would drive the hearing aid to a higher (and possibly uncomfortable) output at discrete frequency regions. Additionally, most prescriptive methods provide RESR fitting targets for narrow-band, not broadband, inputs.

Some concern has been expressed that a risk of acoustic trauma exists when conducting the RESR procedure, and for this reason many clinics do not include it in their test protocol. One could argue, however, that if maximum output isn’t measured in the clinic, sooner or later that same output will be reached in the real world, so why not measure it in the clinic so that it can be adjusted appropriately? If conducting the RESR is a concern, it is a simple process to calculate a “predicted RESR,” by adding the RECD to the OSPL90. This is the procedure we recommend for children, as it has been shown to provide values that are very close to the measured RESR (Seewald et al, 1999).

Aided Loudness Judgments

Although probe-mic testing provides an excellent indication if the hearing aid’s maximum output is adjusted correctly, and also provides useful frequency-specific information, it does not assure that the output is adjusted correctly for broadband signals like speech. Factors such as binaural summation, loudness summation, and hearing aid channel (power) summation all can influence the patient’s loudness perceptions.

We recommend, therefore, conducting aided loudness judgments for speech at three input levels: 45-, 65-, and 85-dB SPL (see Valente and Van Vliet, 1997, and Mueller, 1999, for review). Although our primary concern is the setting for maximum output, it also is important to know the patient’s loudness ratings for soft and average inputs. This will provide guidance if hearing aid adjustments are needed. For example, if soft, average and loud were all judged to be too soft, we would simply increase gain. If only average and loud sounds were too soft, we would increase the kneepoint, and if only loud were too soft we would make the WDRC ratio smaller. If loud sounds were still too soft, we would increase the AGCo kneepoint.

Using the Cox Contour Test descriptive anchors, described earlier in this chapter, the following protocol can be used:

- Present the speech signal at 45-dB SPL. The desired rating from the contour chart is no. 2. Acceptable ratings would be a no. 1 or a no. 3.
- Present the speech signal at 65-dB SPL. The desired rating from the contour chart is no. 4. Acceptable ratings would be a no. 3 or a no. 5.
- Present the speech signal at 85-dB SPL. The desired rating from the contour chart is no. 6. Acceptable rating would be a no. 5 (no. 7 is not acceptable).

It is important that the correct SPL values are used. Mueller and Hall (1998; pp. 243–245) provide a step-by-step protocol for calibration. If the hearing aid fitting is conducted in a setting where an audiometer or test booth is not available, the desired input levels can be recorded on separate tracks of a CD and presented to the patient using an inexpensive portable stereo system (e.g. boom-box).
It is also possible to do more extensive loudness scaling with speech to verify the fitting. Figure 2–16, from Palmer and Mueller (2000), illustrates the unaided and aided results for a given patient. Observe the mean (and standard deviation bars) for normal hearing listeners for the seven categories of the Cox Contour Test. These are the targets for the aided testing. Notice that this patient had unaided loudness judgments that were outside the normal range for all categories except no. 7. When aided, all loudness perceptions fell within the normal range. The normative data shown in Figure 2–16 may not be the same for all clinics (due to room and loudspeaker differences), so we recommend that you establish your own sound-field norms for the contour test (Palmer and Mueller, 2000; Cox and Gray, 2001). Additionally, Cox and Gray (2001) have shown the minor procedural variables can influence the sound-field scaling, so it is important to use a structured protocol. As we discussed earlier, there appears to be little binaural summation for aided testing in the sound field, so once norms are established, they can be used for either monaural or binaural measurements (Hawkins et al, 1987; Cox and Gray, 2001).

It also is important to conduct loudness testing using environmental sounds; using loud obnoxious sounds will assure that you have created a worst-case scenario for determining if the maximum output is okay (Mueller and Bentler, 1994). A patient’s LDL often will vary depending on the “quality” or

![Figure 2-16](image_url)

**Figure 2–16.** Illustration of normal hearing listeners’ judgments of loudness for binaural speech. Square symbols represent the mean with standard deviation bars. U, the unaided results for a sample patient; A, the aided results. In this example, normal loudness perception has been returned to this patient with hearing aids. (From Palmer CV, Mueller HG. Hearing aid selection and assessment. In: Alpiner JG, McCarthy PA, eds. Rehabilitative Audiology. New York: Lippincott Williams & Wilkins, 2000, pp. 332–376. Reprinted with permission from Lippincott Williams & Wilkins.)
“pleasantness” of the test signal, and it’s certainly possible that loud speech signals will be rated “okay” but loud obnoxious environmental noises will not. There are several CDs available that include a variety of environmental noises, and several hearing aid manufacturers have included noises as part of their fitting software. Flamme and Cox (1999) have developed instructions for assembling a kit using calibrated noisemakers, which also works well for this type of testing.

Self-Assessment Scales

We already have talked about two self-assessment scales that can be used to assess loudness: the aversiveness subscale of the APHAB and the PAL. Either or both of these scales can be used to help determine if the clinical verification measures resulted in the correct hearing aid maximum output settings for the real world. Although there only is limited research using these scales to assess loudness, they appear to be sensitive to real-world issues. For example, Ebinger et al (1996) compared the aversiveness scores for a group of patients fitted with linear peak-clipping CICs versus a second group fitted with CICs incorporating K-Amp technology; the patients fitted with the CIC K-Amps reported significantly fewer problems for the aversiveness subscale. Mueller and Powers (2001) report than when the hearing aid’s maximum loudness is adjusted at the time of the fitting using speech scaling, both new and experienced users report PAL loudness judgments for “loud” equal to that of people with normal hearing after 1 month of hearing aid use. Both the PAL and the APHAB can be used for individual cases to provide direction concerning the possible need for adjustment of the hearing aid’s maximum output.

Conclusion

There are several methods to measure or predict a patient’s LDL. Each method has its own set of variables that must be considered, different LDLs tend to result from different methods, and we really don’t know what LDL is the “right” LDL. We do know, however, that selecting the correct hearing aid output for a given patient will increase patient satisfaction. Whether the correct output is determined through prefitting testing or through adjustments on the day of the fitting may not be important—just so it’s accomplished at one time or another. Some audiologists might say that because of today’s hearing aid technology, and the widespread use of WDRC, the selection of maximum output is no longer a major concern. Yet when we review surveys of patients who are using hearing aids that were only fitted a few years ago, we find that only 38% are satisfied with the way their hearing aids amplify loud sounds. This is not good.

Today’s hearing aids have a variety of adjustments that allow us to control the maximum output. By including probe-mic measures and loudness judgments for speech and environmental sounds in our verification protocol, we can assure that the maximum output is adjusted correctly. And no fitting protocol is complete without real-world validation. We now have self-assessment tools that allow us to validate our verification protocol, and we can use this information to make changes to the maximum output settings when necessary. When the maximum output is adjusted appropriately, it is reasonable to assume that increased patient benefit and satisfaction will follow.

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