

Distortion product otoacoustic emission input/output functions in normal-hearing and hearing-impaired human ears

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(Received 2 May 2001; revised 8 September 2001; accepted 17 September 2001)

DPOAE input/output (I/O) functions were measured at $7f_2$ frequencies (1 to 8 kHz; $f_2/f_1 = 1.22$) over a range of levels (-5 to 95 dB SPL) in normal-hearing and hearing-impaired human ears. L_1-L_2 was level dependent in order to produce the largest $2f_1-f_2$ responses in normal ears. System distortion was determined by collecting DP data in six different acoustic cavities. These data were used to derive a multiple linear regression model to predict system distortion levels. The model was tested on cochlear-implant users and used to estimate system distortion in all other ears. At most but not all f_2 's, measurements in cochlear implant ears were consistent with model predictions. At all f_2 frequencies, the ears with normal auditory thresholds produced I/O functions characterized by compressive nonlinear regions at moderate levels, with more rapid growth at low and high stimulus levels. As auditory threshold increased, DPOAE threshold increased, accompanied by DPOAE amplitude reductions, notably over the range of levels where normal ears showed compression. The slope of the I/O function was steeper in impaired ears. The data from normal-hearing ears resembled direct measurements of basilar membrane displacement in lower animals. Data from ears with hearing loss showed that the compressive region was affected by cochlear damage; however, responses at high levels of stimulation resembled those observed in normal ears. © 2001 Acoustical Society of America. [DOI: 10.1121/1.1417524]

PACS numbers: 43.64.Ha, 43.64.Jb [BLM]

I. INTRODUCTION

Distortion product otoacoustic emissions (DPOAEs) are produced by nonlinear mechanisms within the cochlea that are tied to outer hair cell (OHC) function. Normal OHC function is necessary for the auditory sensitivity, sharp frequency resolution, and wide dynamic range that are hallmarks of normal auditory function. A literature exists describing normal patterns of DPOAEs [see Probst *et al.* (1991) or Lonsbury-Martin *et al.* (2001) for reviews]. It is well known that damage to the OHCs results in reduced auditory sensitivity (e.g., Dallos *et al.*, 1978; Liberman and Dodds, 1984). As a consequence, one would expect that some relation would exist between DPOAEs and auditory threshold. Indeed, many studies have shown that DPOAEs are reduced or absent in ears with hearing loss (e.g., Martin *et al.*, 1990; Bonfils and Avan, 1992; Avan and Bonfils, 1993; Gorga *et al.*, 1993, 1996, 1997, 2000; Stover *et al.*, 1996; Kim *et al.*, 1996). DPOAEs are now in common use for the purposes of identifying normal or impaired auditory function, as defined by threshold sensitivity. In these applications, eliciting stimuli are typically presented at a single moderate level, DPOAE level (or signal-to-noise ratio, SNR) is measured, and a determination is made as to whether the response would be expected from an ear with normal hearing or an ear with hearing loss. The clinical utility of these measurements is based on the theory that OHC function is important in determining both DPOAE level and auditory sensitivity.

Furthermore, damage to the OHCs might be expected to affect the way cochlear responses grow with level. For example, normal input/output (I/O) functions derived from direct basilar membrane (BM) measurements (Ruggero and Rich, 1991; Ruggero *et al.*, 1997) and from ear-canal recordings (Norton and Rubel, 1990; Whitehead *et al.*, 1992a, b; Mills *et al.*, 1993; Mills and Rubel, 1994) in lower animals show a similar pattern of response. When a place on the BM is driven at its best or characteristic frequency (CF), there is linear growth in response to low stimulus levels, nonlinear growth at moderate levels, and a linear response to stimuli presented at high levels (e.g., Ruggero and Rich, 1991). However, when this same place is driven by a tone whose frequency is well below CF, the level at which motion is first detected is elevated, there is little or no evidence of compression, and the slope is steeper, compared to the slope for CF tones. Following administration of furosemide, an agent known to affect the stria vascularis (which maintains the endocochlear potential that serves as the power supply for OHC motility), the lowest level at which BM motion was detected was elevated, compression was reduced, and the slope of the I/O function was steepened when the stimulus was at CF. Thus, disabling the OHCs resulted in a response to a tone at CF that was reminiscent of the response to a tone lower in frequency relative to CF. Interestingly, the administration of furosemide had no influence on the response to the tone below CF.

Similar patterns have been observed in indirect measurements of response growth to tonal stimuli from lower animals with normal and abnormal cochlear function. Specifically, DPOAE I/O functions were nearly linear at levels close

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to threshold, demonstrated compression for moderate-level stimuli, and showed a more linear pattern at high levels. (Whitehead *et al.*, 1992b; Norton and Rubel, 1990; Mills and Rubel, 1994; Ruggero and Rich, 1991). Systematic changes in these DPOAE I/O functions were observed following the administration of ototoxic drugs or soon after the animal was sacrificed. DPOAE threshold was elevated, the region of compression was reduced, and the slope steepened. Just as the normal DPOAE I/O function resembled direct BM measurements from lower animals, so did the DPOAE I/O function following cochlear insult.

DPOAE I/O data have been described in terms of two distinct sources (Norton and Rubel, 1990; Whitehead *et al.*, 1992a, b; Mills and Rubel, 1994). One is a low-level source that is active, sharply tuned, and governed by interactions between the OHCs and mechanical motion of the BM. This source is physiologically vulnerable, as evidenced by large response changes at low and moderate levels when the cochlea is damaged. In the two-source model, there also is a high-level source that is thought to be passive, underlies the active components at low-levels, and reflects BM vibration without acting upon it. This source is more resistant to cochlear trauma, shows little reduction in amplitude between pre- and post-trauma conditions, and exhibits response growth that is more linear, compared to the low- and moderate-level portion of the I/O function. Whitehead *et al.* (1992a, b) further note that the low-level and high-level portions of the function are differentially altered with explorations into the frequency (regions from 1 to 10 kHz) and level (45 to 75 dB SPL) space. In addition to the two-source model to describe DPOAE I/O functions, a single-source model has also been presented (Lukashkin and Russell, 1999) to account for observed DPOAE I/O function patterns. They suggest that any system having a saturating input-output function, such as the mechano-electrical transduction of the OHCs, can produce these patterns.

In summary, data from lower animals reveal that direct measures of BM motion and indirect DPOAE measures of cochlear-response properties are similar, at least in form, for a wide range of levels in ears with normal hearing, and undergo similar changes in ears with induced cochlear lesions. While direct measurements of BM motion are impossible in humans, DPOAE measurements are feasible. Currently, DPOAEs are used mainly to detect threshold hearing loss. Given the similarities between direct and indirect measures of cochlear responses for a wide range of conditions in lower animals, however, it is possible that DPOAE measurements will provide information regarding suprathreshold processing in humans. Indeed, our own preliminary findings in three subjects with normal hearing and one subject with mild hearing loss suggested that DPOAEs can provide indirect measures that are at least qualitatively similar to what has been observed in lower animals (Neely *et al.*, 2000). It is useful to consider responses from normal and impaired human ears in light of direct and indirect measurements in healthy and traumatized ears of lower animals, thus affording an opportunity to compare outcomes from human and lower animal research. This may help determine the extent to which indirect DPOAE measurements in humans can be used to provide

insights into cochlear function over a wide range of levels that are similar to estimates based on direct measurements in lower animals.

The present study was designed to measure DPOAE I/O functions in humans over a wide range of input levels, with a goal of determining the extent to which these data can be used as indirect measures of cochlear-response properties for both normal and impaired ears. The responses obtained in humans with normal hearing are compared to the responses seen in humans with hearing loss. To the extent that DPOAE I/O functions are measures of cochlear-response growth, these comparisons help to describe the changes in response growth that occur as a consequence of hearing loss in humans.

II. METHODS

A. System distortion

The level of distortion produced by the measurement system was estimated by measuring distortion products in six different cavities: (1) an IEC711 coupler with the ER-10C microphone at the standard distance from the coupler's microphone, (2) an IEC711 coupler with the ER-10C within 1 mm of the microphone, (3) a standard 2-cm³ coupler, (4) a small brass cavity (0.1 cm³), (5) a 10-m PVC tube (internal diameter=8 mm), and (6) a 65-cm³ syringe. The distance between the tip of the ER-10C probe and the end of the cavity ranged from less than 1 mm (the close position in the IEC711 coupler and in the 0.1-cm³ brass tube) up to 10 m (the 10-m tube). This set of cavity measurements was used in efforts to determine the source of system distortion. The cavities with small volumes (the IEC711 coupler with the probe unit close to the microphone and the small brass cavity) were chosen because targeted levels could be achieved with less voltage delivered to the loudspeakers. Thus, these conditions should isolate distortion due mainly to the ER-10C probe's microphone and amplifier. Measurements in the large cavity (65-cm³ syringe) were included because this represented a condition in which larger voltages were needed to achieve targeted SPLs. These measurements were included to help determine if system distortion was generated by the loudspeakers. Measurements in standard cavities (IEC711 coupler with the probe in the standard position, the 2-cm³ cavity) were included to represent the standard conditions for the "average" human ear. The measurements in the long tube were chosen because the tube presents an impedance similar to that seen in an anechoic termination roughly corresponding to a normal average ear canal, without standing waves. Such standing waves are present in standard cavities and in human ear canals. The distortion in each of these cavities was similar, and did not allow us (on the basis of these measurements) to isolate the source exclusively to the stimulation or recording side of the measurement system. However, the availability of measurements in all six cavities allowed us to derive a predictive model of system distortion.

Signal levels were calibrated in each cavity, so that all measurements were made for equivalent SPLs at the probe microphone. Cavity measurements were made with primary levels of 75 to 85 dB SPL (f_2 frequencies at 6 and 8 kHz) or

75 to 95 dB SPL (f_2 frequencies at 1, 1.5, 2, 3, and 4 kHz), determined by the output limitations of the loudspeakers in the ER-10C. The ER-10C probe system was modified in order to bypass the 20-dB attenuator in the probe driver-amp and, thus, enabled us to achieve these output levels. In addition to the developed SPLs, the voltages (dBV) needed to drive the two loudspeakers to achieve these SPLs and the primary frequencies were noted. These values were used as input variables into a multiple linear regression to provide an estimate of system distortion. For each cavity, data were input to the linear model if the measured stimulus levels (L_1 and L_2) were within 1 dB of the target value. Because L_2 had linear growth, the predictive linear model for system distortion could be extended down to lower L_2 levels (<75 dB SPL). The following predictive linear model for system distortion (D), in dB SPL, was derived:

$$D = 1.57L_2 + 1.89V_1 + -2.51V_2 + 1.48f_2 - 106.90, \quad (1)$$

where L_2 represents levels from 75 to 85 or 95 dB SPL in 5-dB steps. V_1 and V_2 are the voltages (in dBV) needed to drive the loudspeakers that produced f_1 and f_2 , respectively. The f_2 frequencies were 1, 1.5, 2, 3, 4, 6, and 8 kHz.

To obtain an overall estimate of the level below which reliable data could not be obtained, the energies derived from these system-distortion estimates were added to the measured noise-floor estimates in each ear to determine a distortion-plus-noise level (D+N). The D+N estimates were dominated by noise at levels below 70 dB SPL and by distortion at levels of 70 dB SPL and above. It should be noted that in the development of the predictive linear model previously presented, more complicated models involving more input variables were evaluated, such as interaction and squared value terms. However, the derived estimates of system distortion from these more complex models were similar to those obtained with the simpler model. The distortion estimate (D) was first applied to cochlear-implant users to test the model. Cochlear-implant users were selected because it was assumed that any distortion measured from their implanted ears (device turned off) would not be of cochlear origin, due to the magnitude of their hearing losses and the implant surgical procedure.

B. Subjects

DPOAE I/O functions were measured in 27 ears from 27 subjects (aged 14 to 40 years) having thresholds of 20 dB HL or better (*re*: ANSI, 1996) at octave and half-octave frequencies from 0.250 to 8 kHz. These measures were also made in 84 ears from 50 subjects (aged 13 to 83 years) with hearing loss ranging from 25 to >70 dB HL. In addition, four subjects with cochlear implants were used (implanted ear with device turned off) in efforts to describe the levels at which system distortion occurred (as described in Sec. II A). All subjects had normal middle-ear function, based on 226-Hz tympanometry, at the time of the DPOAE test.

C. Stimuli

DPOAE I/O functions were measured in response to pairs of primary tones, labeled f_1 and f_2 , at a fixed f_2/f_1

= 1.22, with f_2 ranging from 1 to 8 kHz in half-octave steps. The DPOAE I/O functions were measured at L_2 levels ranging from -5 to 85 dB SPL (6 and 8 kHz) or from -5 to 95 dB SPL (1, 1.5, 2, 3, and 4 kHz). The levels of the stimuli were determined in the ear canal at the plane of the probe. Primary tones were presented such that as L_2 decreased below 65 dB SPL, the separation between L_1 and L_2 increased. This level difference paradigm has been shown to maximize the measured level of the $2f_1 - f_2$ DPOAE in normal-hearing ears (Whitehead *et al.*, 1995; Kummer *et al.*, 1998; Janssen *et al.*, 1998).

In some subjects, due to equipment limitations, it was difficult to attain the target stimulus level in the ear canal at higher levels. As a consequence, the only data points retained for further analyses were those for which the average ear-canal level $(L_1 + L_2)/2$, was within ± 3 dB of the target level. This resulted in the removal of 213 data points, with the majority (141) occurring at 95 dB SPL and mainly at f_2 frequencies from 1.5 to 4 kHz. In contrast, the number of data points meeting this inclusion criterion were 4967, meaning that, for the majority of ears and stimulus conditions, targeted SPLs were achieved.

D. Procedures

DPOAE measures were obtained with the subject comfortably seated in a sound-treated booth. The primary tones were generated using a Pinnacle (Turtle Beach) sound card and delivered to the ear canal by means of an ER-10C (Etymotic Research) probe-microphone system, in which the probe driver-preamp was modified in order to recover 20 dB of attenuation. The ER-10C houses two loudspeakers (receivers) and a microphone that were coupled to the ear canal by means of a foam probe tip. The two primary tones were generated by separate channels of the sound card, delivered individually to each loudspeaker, and mixed acoustically in the ear canal. Testing was performed using a PC running custom-designed software (EMAV, Neely and Liu, 1993) that enables the use of measurement-based stopping rules. Data collection at each point on the I/O function continued until the noise floor during individual measurements was equivalent to the estimated level of system distortion based on coupler measurements, or after 32 s of artifact-free averaging, whichever occurred first. A stopping rule based on coupler measures of system distortion was used because these estimates were level dependent, increasing as primary level increased above 70 dB SPL (see Fig. 2). For L_2 levels ≤ 70 dB SPL, this rule resulted in a stopping criterion (D+N) of between -20 and -30 dB SPL, because the criterion was dominated by noise at these moderate and low primary levels. Thus, a stopping rule based on noise level could have been used. However, there would have been little point in stopping at a fixed noise level (for example, -20 or -30 dB SPL) for primary levels ≥ 75 dB SPL because "response" levels at $2f_1 - f_2$ equivalent to system distortion levels would not have been interpretable as biological responses, regardless of the noise level measured at adjacent frequencies. A stopping criterion based on the sum of system distortion and noise took this into account.

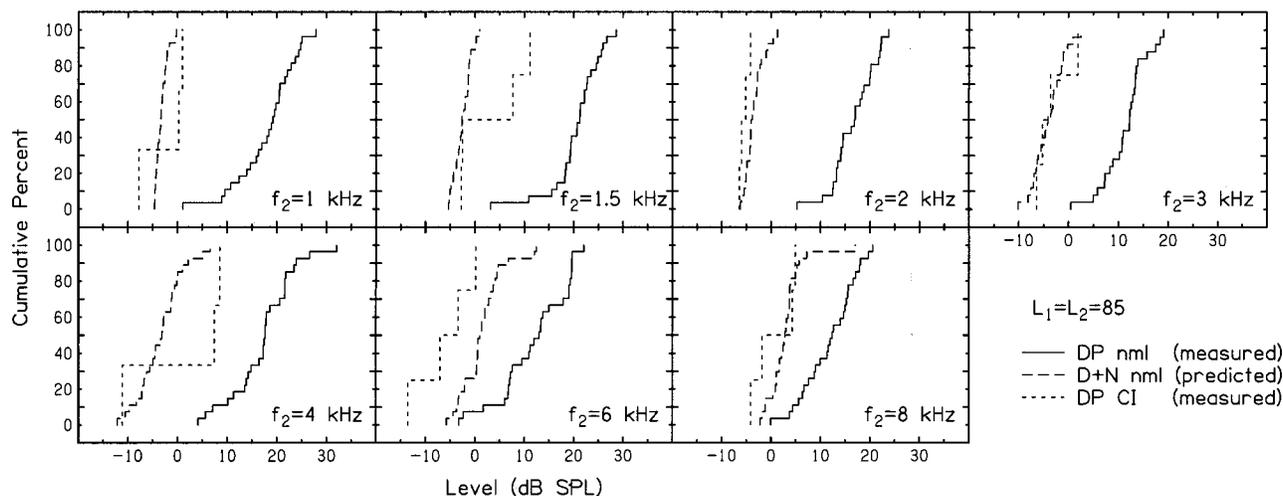


FIG. 1. Cumulative distributions (in percent) of DPOAE response levels at seven f_2 frequencies in normal-hearing subjects (solid line) and in cochlear-implant subjects (short dashed line) when $L_1=L_2=85$ dB SPL. An overall estimate of system distortion and noise (D+N, long dashed line) is displayed for the normal-hearing group.

III. RESULTS

A. Estimates of system distortion

Estimates of system distortion, as determined through the cavity measurements and analyses described earlier, were used to help ascertain if measures obtained at high stimulus levels were of biological origin or due to system distortion generated by the hardware. The estimate of D+N in normal ears was viewed in relation to DPOAE levels from normal ears ($N=27$). The same measurements were made in ears with cochlear implants ($N=3$ at 1 and 4 kHz or $N=4$ at 1.5, 2, 3, 6, and 8 kHz) to determine whether distortion products (DPs) measured under these circumstances were similar to estimates of D+N. If so, these apparent DPs may be interpreted as arising from system distortion. Recall that cochlear-implant subjects were included to provide a model of a biological system in which cochlear-generated distortion should not occur. Although data will be presented for a single high-level condition, similar data were acquired for all high-level primaries (≥ 75 dB SPL).

Figure 1 displays estimates of D+N and DPOAEs in subjects with normal hearing at each f_2 in the form of cumulative distributions (in percent). Cumulative distributions of the DP also are shown for the group of subjects with cochlear implants. The results shown in this figure are for the stimulus condition in which $L_1=L_2=85$ dB SPL. This primary level was chosen because it was the highest targeted SPL that could be achieved at all seven test frequencies.

DPOAEs in normal-hearing subjects were separated from estimates of D+N at all seven frequencies. In addition, the DPs from subjects with cochlear implants were similar to or less than D+N estimates at five of seven frequencies (1, 2, 3, 6, and 8 kHz), consistent with model predictions, but were higher than D+N at 1.5 and 4 kHz. Even for these two frequencies, however, DPOAEs from ears with normal hearing were separated from DPs in ears with cochlear implants. Thus, we would conclude that at this L_2 , measured DPOAE levels were biological in origin, at least in ears with normal hearing. Unfortunately, the differences between D+N and

cochlear-implant cumulative distributions increased (with higher distortion in cochlear-implant ears) at the higher L_2 levels of 90 and 95 dB SPL, even though distributions of DPs in ears with cochlear implants remained separated from distributions of DPOAEs in ears with normal hearing. The distributions of D+N and DPOAEs from normal-hearing subjects also remained separated at these higher L_2 levels.

Given that the DPOAEs in normal ears were highest in level, that estimates of D+N were the lowest in level, and that DPs from ears with cochlear implants were greater than D+N, but less than those observed in ears with normal hearing, several hypotheses may be suggested. The responses from normal ears included all possible sources of distortion, including distortion generated by biological sources as well as distortion generated by the hardware used to elicit and record these responses. However, it was observed that in normal-hearing ears, DPOAEs exceeded the level produced by cochlear-implant subjects and estimates of system distortion (assuming that the D+N measured in the couplers estimates the D+N levels occurring when testing in ears). This observation suggests that the distortion measured in normal ears was of biological origin.

The more complicated interpretation is associated with the data from subjects with cochlear implants. These are ears in which cochlear-generated distortion was not expected. The combination of pre-implant hearing loss (which would be consistent with at least severe loss of both OHCs and inner hair cells) and the changes to any residual nonlinear cochlear mechanical response as a result of implantation would likely eliminate all cochlear sources of distortion. Yet, the distortion measured in the ear canals of these implant subjects sometimes exceeded estimates of D+N (see, for example, Fig. 1, $f_2=4$ kHz). If we accurately estimated D+N, then it may not be entirely appropriate to associate the distortion measured in their ear canals with system distortion, meaning that an alternative, as yet undetermined, biological source might exist.

In summary, the above observations led us to conclude

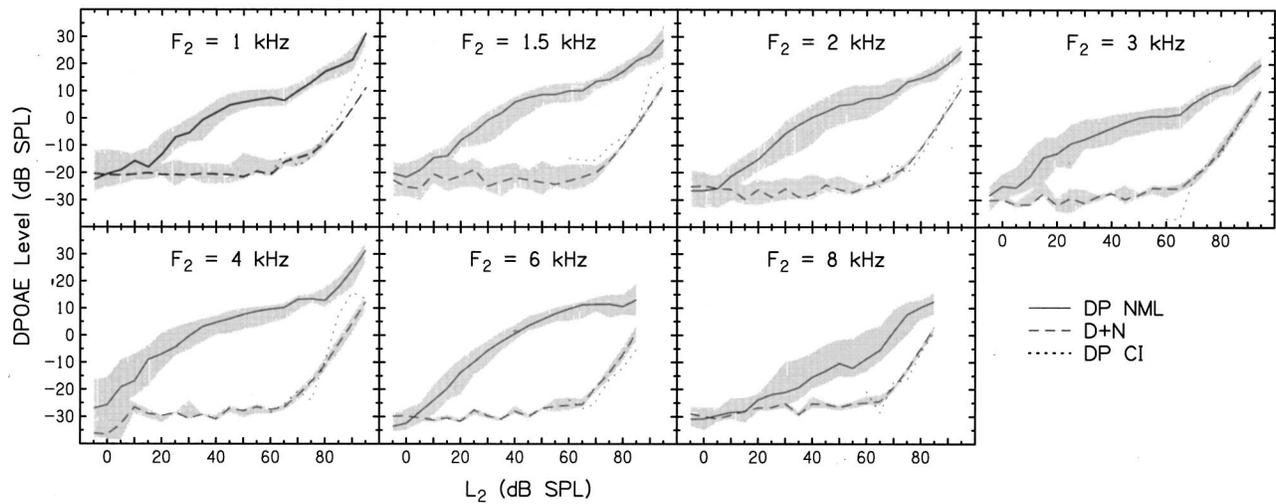


FIG. 2. DPOAE I/O functions for normal-hearing subjects at seven f_2 frequencies. Median DPOAE levels (solid line) and median D+N levels (dashed line) are shown, along with their associated interquartile ranges (25th and 75th percentiles, shaded regions). Median data from cochlear-implant subjects are shown as dotted lines.

that our measurements of DPOAEs were biological (cochlear) in origin, at least for subjects with normal hearing. As will be shown subsequently (Fig. 3), similar conclusions may apply for some subjects with hearing loss, but we are less confident of the source of the DPOAE as the magnitude of hearing loss increases into the moderate-to-severe range. As a consequence, caution must be exercised when considering data points at those higher levels in ears with hearing loss, due to the fact that, for some of these subjects, their responses were not well separated from those observed in cochlear-implant ears.

B. DPOAE I/O functions in subjects with normal hearing

Figure 2 displays DPOAE I/O functions for normal-hearing subjects. The median DPOAE levels and median D+N levels are shown, along with their associated interquartile ranges (25th to 75th percentile, shaded regions). The median DPs from subjects with cochlear implants are also shown in each panel. Medians were chosen to be consistent with other representations in this figure. At all f_2 frequencies except 8 kHz, DPOAEs from ears with normal hearing increased above the overall noise estimate (D+N) with a relatively steep slope, followed by a compressive region for moderate input levels. Starting at between 60 and 75 dB SPL, the slope of the I/O function steepened. The steeper high-level segment was not obvious at 6 and 8 kHz, compared to lower frequencies, which may be partially due to stimulus output limitations at these two frequencies. The overall pattern was reminiscent of what has been reported from direct measurements of basilar-membrane motion in lower animals (Ruggero *et al.*, 1997; Ruggero and Rich, 1991). This finding was not surprising, given that DPOAEs are tied to BM mechanics. At 8 kHz, the D+N function was similar to that of other f_2 frequencies; however, DPOAE level at low stimulus levels had a slope that was less than was observed at other frequencies. In addition, the range of levels over which compression was obvious was reduced at 8

kHz, in comparison to all other f_2 frequencies. Finally, greater variability was observed at 8 kHz, although it was unclear how this would contribute to the differences in the shape of I/O functions. Even with these differences across f_2 , separation between DPOAE and D+N functions was observed for the entire range of stimulus levels, including high L_2 levels (all f_2 frequencies) once DPOAE threshold was exceeded. Furthermore, DPOAE levels in ears with normal hearing exceeded the levels observed in subjects with cochlear implants. At three f_2 frequencies (1, 1.5, and 4 kHz), the median DPs from implanted ears exceeded D+N at the highest L_2 levels. At the other four f_2 frequencies, the median levels in subjects with cochlear implants were at or below system distortion. All of these findings support the view that high-level responses were of biological origin in normal human ears. In spite of these observations, we remain cautious in our interpretation of results for levels >85 dB SPL.

C. DPOAE I/O functions in subjects with hearing loss

Data from subjects with hearing loss were grouped according to degree of loss and are presented in Fig. 3. Within this figure, median data from the normal-hearing and hearing-loss groups are represented by a series of thinning solid lines; the thinner the line, the greater the hearing loss. In addition, median data from subjects with cochlear implants are included as dotted lines. There are several observations to be made from these data. DPOAE I/O functions from ears with hearing loss differ from those observed in ears with normal hearing. This was true regardless of the amount of hearing loss. I/O functions from ears with hearing loss were characterized by elevations in threshold (defined as the lowest level at which the DPOAE exceeds D+N), reduced range of stimulus levels over which compression was apparent, and, as a consequence, steeper slopes through the moderate-level range, compared to I/O functions in the normal-hearing group. At 2 and 4 kHz, this increase in DPOAE threshold and steepening of the slope through the

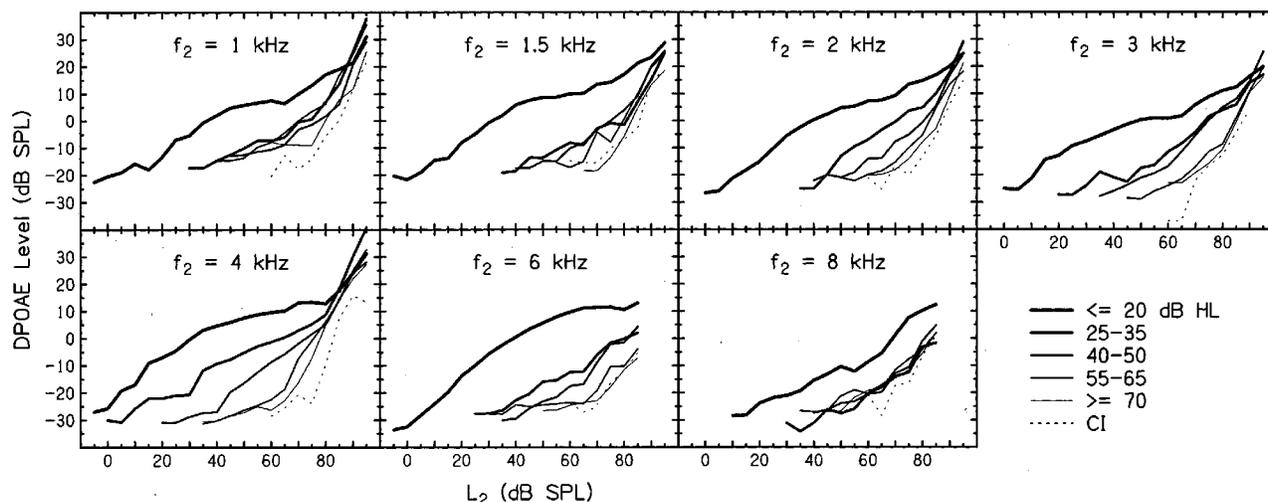


FIG. 3. DPOAE I/O functions at seven f_2 frequencies in subjects with normal-hearing and in subjects with hearing loss. Data are grouped according to audiometric threshold. Median data are represented by a series of thinning solid lines; the thinner the line, the greater the hearing loss. Data from the normal-hearing subjects (thickest solid line) are reproduced from Fig. 2 for comparison purposes. Dotted lines represent median data from cochlear-implant subjects.

compressive region systematically changed as hearing loss increased. At the other frequencies, changes in the I/O function were observed, but the pattern of change was less orderly with regard to magnitude of hearing loss. At 8 kHz, the median DPOAE I/O functions from all impaired ears (regardless of the magnitude of hearing loss) were clustered together and differed less from the values seen in patients with cochlear implants. With few exceptions ($f_2 = 8$ kHz), those cases in which the estimated DPOAE level was indistinguishable from the data produced by ears with cochlear implants were restricted to hearing losses of 55 dB HL or more. At high levels, the grouped data from most of the hearing-loss categories approximated the pattern observed in normal ears and exceeded the median level from cochlear-implant users, the exception to this being the data for an f_2 of 8 kHz. These observations should be considered with the knowledge that system distortion increases as stimulus level increases, thus making it more difficult to separate it from high-level responses in impaired ears.

Regardless of the source of distortion, the combination of DPOAE threshold elevation and near-normal response levels at high stimulus levels resulted in DPOAE I/O functions that were steeper in impaired ears, compared to I/O functions in ears with normal hearing. In turn, these data suggest that response growth, as estimated by DPOAE I/O functions, is more rapid in impaired ears.

D. DPOAE I/O slope estimates in normal and impaired ears

In an effort to quantify the DPOAE I/O functions to help in comparisons between normal-hearing and hearing-impaired ears, the following equation was used to fit the DPOAE data:

$$d(s) = n + \frac{g_0 s}{1 + (g_1 s)^{1-p}} + (g_2 s)^2. \quad (2)$$

The term d refers to the distortion magnitude represented as a power. (To convert to level in dB, one would need to mul-

tiple the log of d by 10.) The signal power is represented by $s = 10^{L_2/10}$. The values of the parameters n , g_0 , g_1 , p , and g_2 , were selected to fit the data. These parameters were estimated separately for each measurement. Parameter g_0 can be interpreted to represent the product of the forward and reverse transfer functions for the middle ear (Kemp, 1980). Parameter g_1 can be interpreted to represent the forward transfer function for the middle ear (Keefe, 2001). Parameter p determines the maximum amount of compression. These three parameters relate to the low-level and compressive portion of the I/O function. The parameter g_2 determines the high-level portion of the function and n is noise (independent of signal level). This equation was fitted separately to the median data for each group, including the data from subjects with normal hearing, subjects with hearing loss, and cochlear-implant subjects. The fits to the DPOAE I/O functions accounted for 98.6% of the variance across all f_2 frequencies and all groups.

The derivative of Eq. (2) was used to provide an estimate of the slope of the DPOAE I/O functions with respect to L_2 . This approach provided a much smoother slope estimate than taking the difference between adjacent points on the fitted I/O function. Figure 4 presents slope estimates as a function of L_2 , following the convention used in Fig. 3 for representing data from different groups by using different line weights.

The slope estimates for data derived from ears with normal hearing at frequencies below 8 kHz were the most straightforward and will be described first. The slope estimates were characterized initially by a steep rise to a local maximum asymptotic value near 1 dB/dB for L_2 levels of 10–20 dB SPL, going from threshold to the level at which the slope began to decrease as level was increased. The upper L_2 limit for the initially steep slope seldom exceeded 30 dB SPL. Beyond this level, the slope decreased to minimum slopes of between 0.1 to 0.3 dB/dB. This range of reduced slopes corresponded to the compressive region of the DPOAE I/O function and extended up to an L_2 of about 80

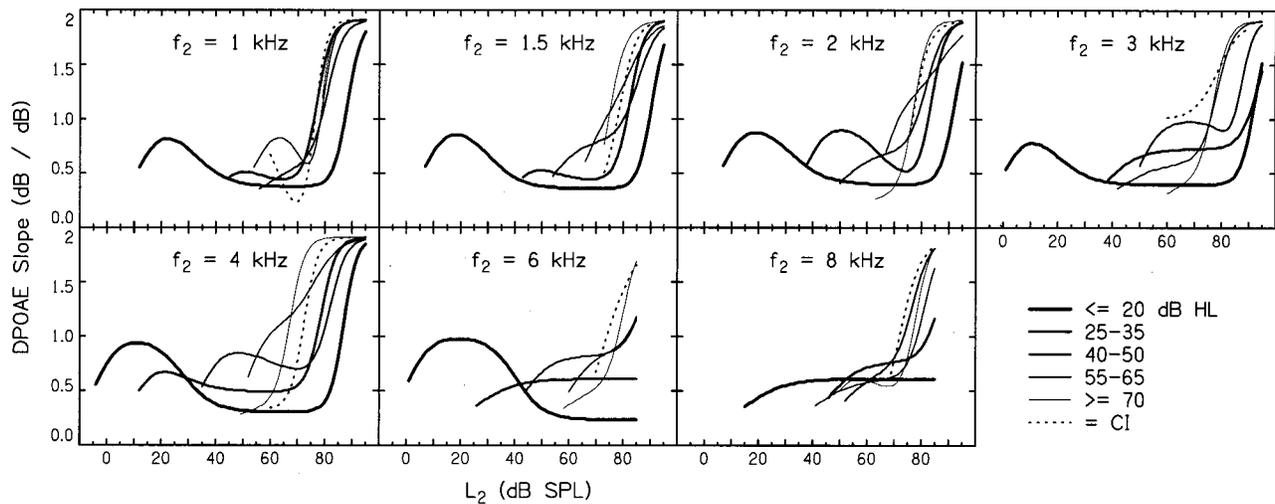


FIG. 4. Slope estimates from DPOAE I/O functions at seven f_2 frequencies. Data are grouped according to audiometric threshold, following the convention used in Fig. 3.

dB SPL. This region was followed by a rapid increase in slope as level increased above 80 dB SPL. The range (in dB) between the early peak at low levels and the steep rise at high levels was used as an estimate of the range of levels over which compression occurred (see Table I).

Among ears with hearing loss, the clearest slope pattern was observed at 4 kHz. For the two mildest hearing-loss groups, the slope initially rose to a local maximum, but achieved this slope at a higher L_2 compared to data from normal ears. Furthermore, the reduction in slope following the initial peak was less than that observed in ears with normal hearing. The steeply rising, high-level portion of the I/O functions actually shifted towards lower L_2 levels, compared to normal. It follows, therefore, that with hearing loss, the slope estimates indicated that both the amount of compression (the extent to which slope decreased for moderate level primaries) and the range of levels over which compression was observed were reduced.

The pattern in ears with greater hearing loss and for some of the other f_2 frequencies was more complex. The initial peak in the slope function, evident in normal ears, was frequently not present in the slope functions of ears with hearing loss, especially among ears with greater degrees of hearing loss. In addition, it was not always possible to identify a range of L_2 levels over which reduced slopes (i.e., compression) were evident. Rather, many of these slope functions were characterized only by a steeply sloping, high-level portion.

Table I presents the range of compression (in dB) for the six f_2 frequencies at which it could be estimated from the slope functions of Fig. 4. The range of compression

TABLE I. Range of compression (dB).

Hearing group (dB HL)	f_2 frequency (kHz)					
	1	1.5	2	3	4	6
≤20	65	71	72	79	75	76
25–35	21	27	32		53	
40–50					30	

was defined from the level of the initial low-level “peak” to the high level at which the slope attained the same value as it did for the initial peak. For all conditions, in order to identify a compressive range, the (fitted) SNR had to be at least 3 dB at the level of the initial peak. At 6 kHz, the high-level steep slope was not observed; therefore, the highest level tested ($L_2=85$ dB SPL) was taken as the upper limit of the compressive range. Empty cells in the table (25–35 dB HL group at $f_2=3$ and 6 kHz; 40–50 dB HL group at $f_2=1, 1.5, 2, 3,$ and 6 kHz) as well as the absence of certain hearing-loss groups (55–65 and ≥70 dB HL groups at all f_2 frequencies) indicated that a compressive region (as defined from the slope functions) could not be identified for these conditions. Subjects with normal hearing produced the largest range of compression, varying from 65 to 79 dB. In the mild (25–35 dB HL) and moderate (40–50 dB HL) hearing-loss groups, the range of compression was always reduced compared to that found in the normal-hearing group.

Table II provides estimates of maximum compression, defined as the minimum slope within the compressive range. The smaller the values in the table, the greater the amount of compression. Lower minimum slopes were found for the normal-hearing group. Thus, greater reduction in output (i.e., compression) relative to linear growth with increases in level was observed in ears with normal hearing, compared to subjects with hearing loss. For all conditions in which it was possible to estimate a compressive range, compression was less in ears with hearing loss, as evidenced by the fact that minimum slopes were larger in these ears. Another way of stating this observation is that response grew more rapidly

TABLE II. Maximum compression (minimum DP I/O slope in dB/dB).

Hearing group (dB HL)	f_2 frequency (kHz)					
	1	1.5	2	3	4	6
≤20	0.31	0.29	0.34	0.34	0.24	0.16
25–35	0.39	0.39	0.47		0.44	
40–50					0.67	

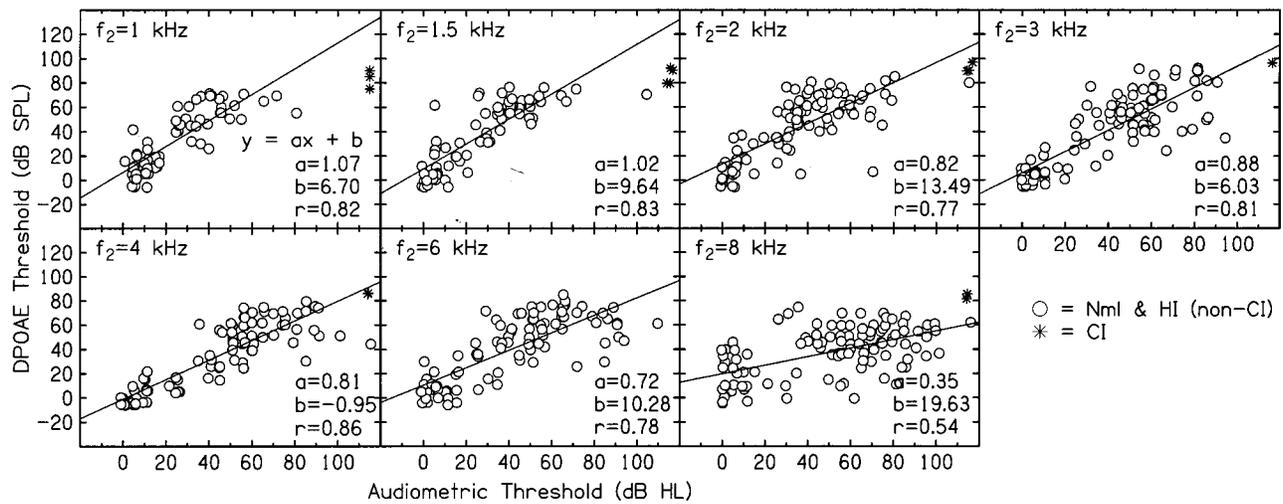


FIG. 5. Scatter plots of DPOAE threshold as a function of audiometric threshold for all subjects at each f_2 . Circles represent individual data from normal-hearing and hearing-impaired (non-cochlear-implant) subjects, while asterisks represent individual data from cochlear-implant subjects. DPOAE threshold was defined as a response that was 3 dB above the estimate of D+N for each subject. Within each panel, the line was derived from a linear regression that was fit to the data, excluding cochlear implant data and data in which DPOAE thresholds in any ear were equivalent to the lowest “DPOAE thresholds” from among the implanted ears (see text for more detail). The correlation coefficient (r), slope (a), and intercept (b) are provided within each panel for each linear regression.

above DPOAE threshold in ears with hearing loss than it did in ears with normal hearing.

It may be important to recognize some of the limitations in the present data, the most important of which relates to system distortion, which would present greater problems for the high-level portions of the I/O functions. We are confident that the measured DPOAEs in ears with normal hearing were biological in origin because they exceeded estimates of system distortion. We are less certain regarding the DPOAEs that were measured in ears with hearing loss, especially as the hearing loss increased above 50–60 dB HL. It is possible that other sources of distortion may have contributed to these measurements. Still, there were many conditions in which the results suggested that the measured DPOAE level in ears with milder hearing losses were biological in origin (i.e., not due to system distortion), and there is little doubt that the slopes of these functions differ in ears with normal hearing from those observed in ears with any degree of hearing loss.

In summary, the data presented in Figs. 3 and 4 and in Tables I and II indicated that the range of stimulus levels over which compression was evident and the amount of compression was reduced in ears with hearing loss compared to subjects with normal hearing. In addition, only the steep, high-level portion of the DPOAE I/O function remained as the hearing loss worsened.

E. DPOAE threshold as a function of audiometric threshold

DPOAE I/O functions can also be used to estimate DPOAE thresholds. The expectation is that as audiometric thresholds become poorer, DPOAE thresholds should worsen since both are indirect measures of peripheral auditory-system function and are associated with OHC function. Behavioral thresholds were measured using standard clinical audiometric procedures with a step size of 5 dB. DPOAE threshold was defined as the lowest level at which the

DPOAE was ≥ 3 dB above the estimate of D+N for each subject. If the DPOAE was never 3 dB above D+N, then no measurable threshold could be identified. In the interest of simplicity, DPOAE and audiometric thresholds were compared for the case when f_2 was equal to audiometric frequency, even though we know that correlations exist across frequency, both for DPOAEs and for audiometric threshold (Dorn *et al.*, 1999). Figure 5 shows scatterplots of DPOAE threshold as a function of audiometric threshold for all subjects at each f_2 . Cochlear-implant subjects’ threshold data (plotted at the upper audiometric limit of 115 dB HL) are shown with asterisks and are included to serve as an upper bound on DPOAE threshold. The best cochlear-implant DPOAE threshold for a given f_2 occurred at a level that was equivalent to the poorest DPOAE threshold included for all other subjects. It could be argued that comparisons between DPOAE and audiometric thresholds should be made only for those conditions in which OHCs might be contributing to both responses. Since it is unlikely that there are any surviving OHCs in the cochlear-implant ears, their “thresholds” were not included in subsequent analyses. In addition, the threshold data from any other subject whose DPOAE threshold was equivalent to the best “threshold” among implanted ears was excluded from analyses. This approach resulted in the exclusion of data points only for hearing-impaired ears mainly at 1 and 1.5 kHz. The remaining threshold data were fit with a linear regression (solid lines, Fig. 5), from which the correlation coefficient (r), slope (a), and intercept (b) were determined and are provided in each panel. As expected, DPOAE thresholds increased as audiometric thresholds increased, with slopes of 0.35 (8 kHz) to 1.07 (1 kHz), although the relationship was variable. With the exception of 8 kHz [a frequency for which the correlation coefficient (0.54) was also the lowest], the slopes ranged from 0.72 to 1.07. Correlations of 0.77 to 0.86 were observed at other f_2 frequencies, including lower frequencies, such as 1 kHz,

where the inherently poor SNR makes DPOAE measurements less reliable, especially near DPOAE threshold. Even after applying the above exclusion criteria (related to thresholds in implanted ears), some caution is necessary in the interpretation of data points for audiometric thresholds exceeding about 60 dB HL. Assuming that DPOAEs are generated by the OHC system, and assuming that OHC damage can account for hearing losses up to about 60 dB HL and that greater losses must involve other parts of the auditory system (such as the inner hair cells), it is not expected that a relationship should exist between DPOAE and audiometric thresholds for losses exceeding 60 dB HL. This issue relates to other concerns regarding the correct interpretation of high-level DPOAEs, expressed as part of the description of DPOAE I/O functions in ears with moderate-to-severe hearing loss.

IV. DISCUSSION

A. DPOAE I/O functions in normal ears

This study demonstrates that it is possible to measure DPOAE I/O functions across a wide range of stimulus levels in humans. In normal ears, the response growth exhibits a three-part function, being linear at low levels, compressive at moderate levels, and linear to expansive at high levels (see Fig. 2). These functions resemble direct measurements of BM I/O functions (Ruggero and Rich, 1991; Ruggero *et al.*, 1997), as well as indirect measurements, such as DPOAE recordings made in healthy lower animals (Whitehead *et al.*, 1992a; Norton and Rubel, 1990). In the present study and others (Whitehead *et al.*, 1992a; Ruggero *et al.*, 1997), where the lower end of the I/O function (<20 dB SPL) is measured, linear or near-linear response growth is seen. Consistent with the observations made from other studies in lower animals (Ruggero and Rich, 1991; Ruggero *et al.*, 1997; Norton and Rubel, 1990; Whitehead *et al.*, 1992a, b), compressive nonlinear growth occurs at moderate stimulus levels. In the present study, the amount of normal compression, defined by the slope of the DPOAE I/O function at moderate levels, ranges from 0.16 to 0.34 dB/dB across f_2 frequencies (1 to 6 kHz). These values are similar to slope estimates derived from chinchilla BM velocity-intensity functions (Ruggero *et al.*, 1997), where the measured slopes were in the range of 0.2 to 0.5 dB/dB.

At high stimulus levels, previous work has shown that response growth is steeper, compared to the slope at moderate levels, with slopes that approach linearity (Ruggero *et al.*, 1997; Whitehead *et al.*, 1992a). In the present study, DPOAE level grew expansively with slopes exceeding 1 dB/dB, at least for the five f_2 frequencies at which it was possible to make measurements (1–4 kHz). The reasons for this expansion are not entirely clear. The high-level portion of the I/O function has few data points associated with it, due to limitations in system output and distortion. Thus, the slope of this portion of the function might be less reliably estimated. One could also speculate that the expansive growth might reflect intermodulation distortion produced by the recording system. However, normal DPOAE levels at maximum stimulus levels were between 9 dB (8 kHz) and 20 dB (1 kHz)

greater than estimates of system distortion (mean difference of 14.4 dB). Thus, the estimated DPOAE amplitude was between 2.5 and 10 times larger than the estimated amplitude of system distortion. Assuming that our estimate of system distortion is reliable, it would be likely that these high-level estimates of DPOAE level are dominated by sources other than the system hardware. While we do not have a good explanation of the source of the high-level, expansive portion of the DPOAE I/O function, it would appear to be biological in origin, at least in ears with normal hearing.

At 6 and 8 kHz, measurements were possible only up to 85 dB SPL, because higher levels could not be achieved by the sound card output and the probe loudspeakers. Given the observation that compressive nonlinear regions were observed up to approximately 80 dB SPL for lower f_2 frequencies, it is possible that our inability to produce higher levels was the reason why the expansive, high-level region was not observed in the present DPOAE I/O functions at these two frequencies. Further, the slope remained relatively constant at approximately 0.6 dB/dB in ears with normal hearing when $f_2=8$ kHz, once L_2 exceeded 20 dB SPL. Thus, we did not observe a range of moderate levels over which compression was observed, which was unlike all other f_2 frequencies, including 6 kHz. It is possible that the L_1, L_2 level paradigm (in which the differences between L_1 and L_2 increased as primary levels decreased) did not maximize responses from normal ears at 8 kHz. Perhaps a different separation between primary levels was needed to maximize distortion in the cochlea at this frequency. In addition, it may be important that middle-ear transmission at 8 kHz is reduced compared to middle-ear transmission for the other f_2 frequencies (Voss *et al.*, 2000; Keefe, 2001). This may have reduced L_2 and/or L_1 , and altered the effective relationship between L_1 and L_2 , as represented in the cochlea.

B. Changes in DPOAE I/O functions due to hearing impairment

In comparison to normal-hearing subjects, the DPOAE I/O functions for the subjects with hearing loss were characterized by (1) threshold elevation, (2) a reduction in response levels, (3) a linearization of response growth at moderate levels, and (4) steeper response growth at high levels. This steepening occurred at levels that were similar to, or lower than, the levels at which this steepening occurred in ears with normal hearing. Thus, the range of input levels over which compression occurred was reduced in ears with hearing loss. In many respects, these observations are similar to those made in lower animals whose ears have undergone trauma (Ruggero and Rich, 1991; Whitehead *et al.*, 1992b; Norton and Rubel, 1990). In general, these changes were most clearly observed in ears with mild and moderate degrees of hearing loss. With greater degrees of hearing loss at essentially all f_2 frequencies, and for all hearing-loss groups at some f_2 frequencies (1.5 and 8 kHz), the compressive region was abolished, and all that remained was linear (or expansive) response growth (see Figs. 3 and 4). The outcome at 8 kHz, where any degree of hearing loss drives most of the data points into the overall noise estimate (D+N), could be due to the measurement issues raised in the section above.

The data reported by Ruggero and Rich (1991) suggest that the OHCs are responsible for the compressive nonlinear behavior of the cochlea for CF tones, showing that reversible damage to the OHC system (through the administration of furosemide) compromises the cochlea's ability to respond nonlinearly. Interestingly, furosemide treatment had little or no effect on the I/O function when the driver frequency was significantly lower than CF. This means that nonlinear response behaviors at a specific place along the cochlea are observed only when that place is driven by frequencies at or near its CF. Direct BM measurements to moderate-level stimuli made in post-furosemide-treated chinchillas resulted in a linearization of the I/O function at CF, which returned to a normal nonlinear pattern when the animal was allowed to recover from the treatment. As always, direct assessments of cochlear response properties in humans are impossible to obtain. However, given the present results in humans with normal hearing and in some subjects with hearing loss, it would appear that DPOAE measurements in humans could be used to describe cochlear mechanical response properties over a wide range of levels. Furthermore, the similarity between direct measurements in lower animals and indirect measurements in humans suggests that the indirect data can be used to estimate both the range of levels over which compression is observed and the amount of compression that exists.

C. Generation of the I/O function

The existence of two sources, one dominant at low levels of stimulation and the other dominant at high levels, has been hypothesized by others (Whitehead *et al.*, 1992a, b; Norton and Rubel, 1990; Mills and Rubel, 1994). The low-level source is considered to be an active process powered by the micromechanical feedback loop between the motile OHCs and BM vibration. This process is responsible for compressive nonlinearities and is physiologically vulnerable, reflecting the vulnerability of the OHCs to cochlear insult. This view appears to be consistent with the present data from humans, in which DPOAE threshold is elevated and the range and amount of compression is reduced in the presence of hearing loss. The high-level source has been viewed as a passive process, whereby BM activity occurs due to macro-mechanical properties of the system. It dominates at high levels and/or when the active low-level source has been abolished. Having described the hypotheses about the high-level portion of the I/O function, it is important to note that the mechanisms responsible for the low- and moderate-level portions of the I/O function (whether directly or indirectly measured) are known, while the mechanisms responsible for (and perhaps even the existence of) the high-level portion of the function are not well understood. In the context of the single-source model (Lukashkin and Russell, 1999), the mechano-electrical transduction of the outer hair cells could be the source for the high-level portion of the DPOAE I/O functions observed in human ears. Regardless of whether a one- or two-source model provides the most parsimonious explanation for the slope of DPOAE I/O functions, there clearly are differences between the DPOAE I/O functions observed in normal and impaired ears at low and moderate levels of stimulation.

The nonlinear compressive region, evident at low-to-moderate stimulus levels in the human data presented here, presumably corresponds to a low-level active source. In the presence of hearing loss, the active process diminishes and the passive process dominates. In the present study, it was assumed that the site of lesion for the hearing-impaired subjects was the OHC system, a reasonable assumption given what is known about the effects of cochlear insults in lower animals. Therefore, their hearing losses would represent dysfunction of the active-energy source. The range of levels over which the OHC and BM feedback loop operates and the amount of compression were calculated from the slopes of the DPOAE I/O functions in normal and impaired ears (Fig. 4), and presented in Tables I and II. The active range (in dB) is largest and the amount of compression greatest for normal-hearing subjects. Similar estimates of range and amount of compression were possible for only a subset of the present data from subjects with hearing loss, restricted to the milder hearing-loss groups for f_2 frequencies of 1, 1.5, 2, and 4 kHz. A comparison of the data from these groups with similar measurements from subjects with normal hearing indicated that the range of levels over which the active process operates is diminished and responses are less compressed as hearing loss increases. The fact that this pattern was observed only for groups of subjects with milder hearing losses is consistent with the view that DPOAEs are generated by the OHC system, and that the OHC system contributes to response properties only for levels up to about 60 dB HL. For greater hearing losses, the loss cannot be attributed solely to the OHCs. Thus, it may not be surprising that changes to the compressive nonlinearity associated with OHC function cannot be quantified with the present analysis approach once the loss exceeds the dynamic range of the OHCs.

The reason why similar patterns were not observed among the milder hearing-loss groups at other f_2 frequencies is not obvious. This partly results from the fact that the slopes of DPOAE I/O functions frequently did not show the initial peak that was characteristic of responses from ears with normal hearing, even though the responses were reliably measured, exceeding the estimate of D+N. At 8 kHz in the normal-hearing ears, the shape of the I/O function at low and moderate levels was different from that observed at the other f_2 frequencies. Further, at 8 kHz, much of the hearing-impaired data were indistinguishable from DPs measured in ears with cochlear implants. This observation is surprising, in that it would suggest that DPOAEs are present in normal ears but absent for any degree of hearing loss at this frequency. We would expect that a range of levels exists over which there is a relationship between DPOAE level and audiometric threshold (e.g., Martin *et al.*, 1990; Gorga *et al.*, 2002). Taking the present observations to their logical conclusion, the data at 8 kHz indicate that OHCs either are present and functional (ears with normal hearing) or absent or completely dysfunctional (ears with any degree of hearing loss). We do not believe that this is, in fact, the case; as a consequence, the mechanisms responsible for the observations at 8 kHz remain unclear. There are other possible causes for the outcome at 8 kHz. Perhaps a non-optimal

L_1/L_2 ratio is, in part, responsible for the measurements obtained at this frequency, thus leading to a situation in which f_1 had a greater suppressing effect on f_2 compared to other primary pairs. Other considerations are middle-ear transmission and standing waves in the ear canal, both potentially altering the signal level that is actually delivered to the cochlea, thereby altering the effective primary levels.

D. System distortion estimates and high-level DPOAEs

The cochlear-implant subjects served as a biological system that was not expected to produce distortion. At higher levels (above 85 dB SPL) and for some f_2 frequencies, the measured responses in their ear canals were not as low as the D+N estimates for normal ears; however, the responses seen in ears with cochlear implants were always less than the DPOAE levels from normal ears. In normal ears at these higher levels, estimates of noise and DPOAE levels were distinct from each other at all f_2 frequencies, suggesting that these responses were biological in origin. While the response levels in ears with cochlear implants were less than those seen in normal ears, they sometimes exceeded the estimates of D+N. One interpretation of the data in cochlear-implant ears is that some part of the biological system (excluding the OHC system) is responsible for the measured DPs, assuming that our estimate of D+N accurately describes system distortion. Given this constellation of outcomes, several possible explanations exist for the source of high-level distortion. These DPOAEs could be thought of as emanating from three possible sources: (1) a biological source dominated by normal cochlear activity, primarily the OHC system, (2) a biological source, but excluding the OHC system (e.g., the BM or middle ear), or (3) a nonbiological source, related to system distortion. We consider the second possible source to be the most speculative, representing the least likely scenario. Effort was expended to control for and understand the limits of system distortion. We recognize the possibility that our estimate of system distortion may underestimate the actual level at which distortion is produced by the hardware. At this point, we can only speculate as to the source of the measured distortion and provide the above three possible explanations. The present data do not allow us to more definitively identify the source in cochlear-implant ears and ears with moderate-to-severe hearing losses.

E. Relationship between DPOAE and audiometric thresholds

The range of slope and correlation values found when DPOAE and audiometric thresholds were compared are similar to the results reported by Martin *et al.* (1990), Nelson and Kimberley (1992), and Sukfüll *et al.* (1996), indicating that a positive relationship exists between the two threshold measures. Gorga *et al.* (1996) also found a similar relationship with the further observation that DPOAE thresholds in the mild HL group often overlapped those observed in ears with normal hearing, indicating that the most ambiguity in separating normal from impaired ears occurred with mild hearing losses. Harris (1990) noted that the correspondence between

the amount of reduction in DPOAE level and degree of loss when behavioral thresholds were between 20 and 50 dB HL was not clear. In contrast, relationships between DPOAE level and audiometric threshold have been described within the range of normal hearing (Dorn *et al.*, 1998) and within the range from normal hearing up to moderate hearing loss (Gorga *et al.*, 2002). While the correlation is not sufficient to permit precise predictions of auditory threshold from DPOAE measurements, both DPOAE threshold and DPOAE level are related to audiometric threshold. Such a relationship should not be surprising, since DPOAE measures and auditory sensitivity are related to the same underlying system (the OHCs) for thresholds up to about 50–60 dB HL.

F. DPOAE I/O functions in clinical applications

The observation of steeper DPOAE I/O functions in ears with hearing loss as compared to normal ears may correlate with other measures of abnormally rapid response growth, such as loudness recruitment. Schlauch *et al.* (1998) described a relationship between BM I/O functions and psychophysical measures of loudness. They measured loudness in humans and, from those data, derived I/O functions. The derived I/O functions were compared to the BM data of Ruggero *et al.* (1997). The results were promising in that a relationship between the psychophysical and physiological measures was found in individuals with loudness recruitment. Zhang and Zwislocki (1995) provided evidence to support the notion that loudness recruitment is present at the hair-cell level by making pre- and post-noise-exposure I/O measurements in the gerbil cochlea. They concluded that abnormal growth in loudness was locally determined and not a consequence of abnormal spread of excitation.

A connection between behavioral, basilar membrane, and hair-cell measures with respect to loudness is important when considering possible clinical applications of DPOAE I/O functions. Behavioral measures of loudness may not be possible in many populations (e.g., infants, children, patients with developmental delays). If DPOAE I/O measures can be associated with growth of loudness, then the development of a clinical DPOAE I/O measure would make it possible to identify those impaired ears that exhibit loudness recruitment. That information could then be applied to the selection of appropriate amplification characteristics, such as compression threshold and compression ratio.

V. SUMMARY

The main observations from the present study include the following:

- (1) It was possible to measure DPOAE I/O functions over a wider range of levels in both normal and impaired ears than in previous studies.
- (2) Predicted estimates of system distortion, based upon a linear model derived from several cavity measurements and tested on a group of cochlear-implant subjects, provides evidence that the measured DPOAEs in normal ears are biological in origin, even at high levels. In spite of these efforts, responses obtained at stimulus levels above 85 dB SPL must be viewed with caution.

- (3) The DPOAE I/O functions from normal ears are reminiscent of basilar membrane I/O functions derived from direct measurements and from DPOAE measurements in lower animals. These functions are steep at low and high stimulus levels, with a compressive region at moderate levels.
- (4) I/O functions obtained in subjects with hearing loss may be characterized as having steeper slopes, and the amount of compression and the range of levels over which compression occurs is reduced, as compared to subjects with normal hearing.
- (5) The normal compressive growth of DPOAEs may be interpreted as an indication of reduced cochlear-amplifier gain with increases in stimulus level. The transition from a compressive region to a region of steep slope may mark the upper limit of cochlear-amplifier function, assuming that high-level distortion is due to the OHCs.
- (6) The observation of steeper DPOAE I/O functions in ears with hearing loss compared to normal ears might relate to other measures of abnormally rapid response growth, such as loudness recruitment.

ACKNOWLEDGMENTS

This work was supported by NIH Grant No. DC02251. Portions of this work were presented at the 24th Midwinter Meeting of the Association for Research in Otolaryngology. We thank two anonymous reviewers for their helpful comments and suggestions.

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