Comparison of distortion-product otoacoustic emission growth rates and slopes of forward-masked psychometric functions^{a)}

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Slopes of forward-masked psychometric functions (FM PFs) were compared with distortion-product otoacoustic emission (DPOAE) input/output (I/O) parameters at 1 and 6 kHz to test the hypothesis that these measures provide similar estimates of cochlear compression. Implicit in this hypothesis is the assumption that both DPOAE I/O and FM PF slopes are functionally related to basilar-membrane (BM) response growth. FM PF-slope decreased with signal level, but this effect was reduced or reversed with increasing hearing loss; there was a trend of decreasing psychometric function (PF) slope with increasing frequency, consistent with greater compression at higher frequencies. DPOAE I/O functions at 6 kHz exhibited an increase in the breakpoint of a two-segment slope as a function of hearing loss with a concomitant decrease in the level of the distortion product (L_d). Results of the comparison between FM PF and DPOAE I/O parameters revealed only a weak correlation, suggesting that one or both of these measures may provide unreliable information about BM compression.

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I. INTRODUCTION

Compressive growth is a feature of normal cochlear responses. Evidence of compressive growth may be observed in the slopes of psychometric functions (PFs) for forward-masked thresholds (FMTs), which become shallower as compression increases. Distortion-product otoacoustic emission (DPOAE) input/output (I/O) functions presumably also reveal evidence of cochlear compression by growing more slowly with increasing stimulus level. In this study, measurements of PF slopes for FMTs are compared with measurements of DPOAE I/O functions to test the hypothesis that the compression observed in these measures is due to the compressive growth rate of basilar membrane (BM) vibration.

Compressive growth is observed in many measures of cochlear function, including BM (e.g., Rhode and Recio, 2000), neural (e.g., Yates *et al.*, 1990), psychophysical (Plack and Oxenham, 1998), and otoacoustic-emission (OAE) (e.g., Dorn *et al.*, 2001; Neely *et al.*, 2003) measurements. Direct BM I/O recordings (e.g., Rhode, 1971; Ruggero and Rich, 1991; Ruggero *et al.*, 1997; Recio *et al.*, 1998; Rhode and Recio, 2000) and estimates from auditory nerve fiber (ANF) rate-intensity functions (Yates *et al.*, 1990; Cooper and Yates, 1994) reveal compressive growth at frequencies near their characteristic frequency (CF). The CF is the frequency to which the ANF or the location of the BM measurement is most sensitive. ANF and BM responses are less compressive at frequencies below CF. At least for the basal sites of the

cochlea, where direct comparisons have been made, ANF neural tuning is almost identical to BM mechanical tuning (Narayan *et al.*, 1998). The compressive nonlinearity at CF may be characterized as a two-segment function: linear near threshold [i.e., slope of 1 dB/dB for low-levels <10–20 dB sound pressure level (SPL)], with a slope of about 0.2 dB/dB (i.e., range of 0.1–0.5 dB/dB) above a breakpoint at moderate intensities. There is debate regarding the breakpoint at which the compressive portion begins; consensus places it at 20–25 dB above hearing threshold for mechanical data in various animal species while for neural data it has been argued that it is much lower (Sachs *et al.*, 1989; Müller and Robertson, 1991a). There seems to be better agreement regarding the upper limits of mechanical compression, placing it as high as 110 dB SPL (Ruggero *et al.*, 1997). This compressive nonlinearity is fast-acting, i.e., the compression is visible within one cycle of the center frequency at the recording site (Recio *et al.*, 1998). Cooper and Yates (1994) derived BM I/O functions for fibers with CFs between 1.5 and 3.6 kHz that were less compressive (0.5 dB/dB) than those for fibers with CFs above 4 kHz (0.13 dB/dB) suggestive of a frequency- and intensity-dependent compressive function. Given the close correspondence between mechanical and neural data, this suggests increased compression at higher frequencies than in lower-frequency regions. In the presence of outer hair cell (OHC) damage, elevated thresholds in ANFs (e.g., Kiang *et al.*, 1970) and more linear (i.e., less compressive) BM growth (e.g., Ruggero and Rich, 1991) are observed. Although loss of sharp tuning is often thought to be associated with OHC damage, some observations suggest this is not always the case (e.g., Dallos and Harris, 1978; Liberman and Dodds, 1984).

Psychophysical and physiological FMTs have been used to estimate compression. Threshold shifts due to forward

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masking have been interpreted as reflecting either temporal integration (Oxenham, 2001) or adaptation produced by the masker (e.g., Duifhuis, 1973; Smith, 1977, 1979; Harris and Dallos, 1979; Jesteadt *et al.*, 1982; Kidd and Feth, 1982; Alves-Pinto *et al.*, 2010) or a combination of these two mechanisms. Psychophysical forward-masking due to temporal integration is conceptualized as an overlap of the internal representations of the signal and the masker that occurs centrally. It can also be thought of as masker persistence, although most recent evidence does not support masker persistence as having a role in forward masking (Alves-Pinto *et al.*, 2010). ANF excitation by a preceding masker results in adaptation which affects the response to a subsequently presented probe or signal (Smith, 1979; Harris and Dallos, 1979). Adaptation observed in eighth-nerve afferents may arise (1) presynaptically, during translation of the stimulus into hair-bundle deflection or while hair-bundle deflection is translated into a receptor potential (e.g., Jia *et al.*, 2007; Beurg *et al.*, 2008); (2) during transmitter release; or (3) postsynaptically in the primary-afferent neuron (Eatock, 2000). Attempts to apply psychophysical threshold estimation methods to single-ANF observations have led to the conclusion that psychophysical forward masking may occur at levels central to the cochlear nucleus (Relkin and Turner, 1988, Alves-Pinto *et al.*, 2010). The discrepancy between the psychophysical and physiological functions as well as evidence of forward masking in cochlear-implant patients (Chatterjee, 1999) has led investigators to suggest that psychophysical forward masking might be occurring at higher auditory centers, possibly at the level of the inferior colliculus (Nelson *et al.*, 2009). However, there are studies that localize adaptation in the auditory peripheral system to the inner hair-cell (IHC) – primary-afferent synapse (Spassova *et al.*, 2004; Goutman and Glowatzki, 2007). The involvement of IHC-primary-afferent synapse adaptation in psychophysical forward masking is also supported by recent computer modeling (Meddis and O'Mard, 2005) and electronic circuit studies (Huang and Chiueh, 2002).

Although psychophysical forward masking might be due to retrocochlear influence, FMTs have been used to infer nonlinear BM response growth (e.g., Lopez-Poveda *et al.*, 2003, 2008; Nelson *et al.*, 2001; Oxenham and Plack, 1997; Plack and Oxenham, 1998; Schairer *et al.*, 2003; Schairer *et al.*, 2008). Estimates of peripheral nonlinearity have been obtained using the comparison of slopes of growth of maskability (GMB) (Oxenham and Plack, 1997; Nelson *et al.*, 2001) or growth of masking (GOM) (Plack and Oxenham, 1998) for on- and off-frequency maskers, from the ratio of temporal-masking curve slopes (TMCs) for on- and off-frequency maskers (e.g., Nelson *et al.*, 2001; Plack *et al.*, 2004; Stainsby and Moore, 2006; Lopez-Poveda and Alves-Pinto, 2008), and from forward-masked psychometric functions (FM PFs) in on-frequency masking conditions (Schairer *et al.*, 2003; Schairer *et al.*, 2008). Studies of GMB, GOM, and TMCs report derived slope estimates for normal-hearing subjects ranging from 0.16 to 0.88 for that portion of the I/O function considered to be “compressive” (typically in the range from 50 to 80 dB SPL) based on data that were averaged across 3–16 listeners. The values obtained for subjects

with moderate-to-severe hearing loss are reportedly linear in some studies (Oxenham and Plack, 1997; Nelson *et al.*, 2001), while in other studies of mild-to-moderate impairment it has been suggested that shallower functions reflect “residual compression” or compression mediated by IHCs (Plack *et al.*, 2004; Lopez-Poveda *et al.*, 2005). Assumptions underlying these procedures include linear growth at the signal place for off-frequency maskers, frequency-independent decay of psychophysical forward masking and frequency-independent compression.

Additivity of masking (Plack and O'Hanlon, 2003; Plack and Arifianto, 2010), the variant temporal masking curve (Lopez-Poveda and Alves Pinto, 2008), and the PF-slope measure require data for only one frequency and do not rely on the assumption that growth of response at a given CF is linear when the masker frequency is well below CF. A PF relates the percentage of correct (PC) responses of the observer to the strength of the signal where only one variable is changing (i.e., masker or signal). For normal listeners, in a FM PF variable signal (VS) paradigm with masker level fixed, on-frequency forward masking behaves in a nonlinear fashion. For example, as signal level at threshold (i.e., masked threshold) increases to compensate for the increase in excitation caused by the masker, it transits through the more compressive region of the function, producing a shallower slope. Schairer *et al.* (2003) obtained FM PFs for on- and off-frequency maskers in a VS paradigm. Schairer *et al.* (2003) interpreted the decrease in FM PF-slope as a measure of compression in normal-hearing listeners. Subsequently, Schairer *et al.* (2008) investigated cochlear nonlinearity at low frequencies in both normal-hearing and hearing-impaired listeners. They observed that FM PFs in listeners with hearing loss were steeper, suggesting loss of compression associated with hearing loss, but found no differences in PF slope as a function of frequency.

DPOAE I/O measurements in humans evidence compressive nonlinearity (e.g., Dorn *et al.*, 2001; Neely *et al.*, 2003). DPOAEs originate in response to a simultaneously presented two-tone probe as byproducts of the cochlea's nonlinear mechanics. The lower-primary frequency of the two-tone probe is referred to as f_1 and its level as L_1 ; the higher frequency is denoted as f_2 and its level as L_2 . The response is labeled a distortion product because it originates from the cochlea as a tone (e.g., f_d) that is not present in the original two-tone probe. The response is typically measured at the $2f_1 - f_2$ frequency, because it is the largest distortion product in humans. It is generally assumed that DPOAEs arise from two sources on the BM: the f_2 place and the DPOAE-frequency place (f_d) (Zweig and Shera, 1995; Talmadge *et al.*, 1998; Shera and Guinan, 1999). The first component (distortion) is thought to arise from the nonlinear interaction between the f_1 and f_2 primary tones; the interaction creates energy at the DPOAE frequency place (f_d) that then travels both apically and basally within the cochlea. When the apically traveling energy reaches the f_d place on the BM some of the energy is reflected back basally, to produce a second DPOAE component reflected from the f_d place. These two DPOAE components combine to produce the emission signal measured in the ear canal. DPOAE amplitudes measured as a function of

stimulus level of the two-tone probe have been suggested as representative of the compressive growth of BM displacement (e.g., Mills and Rubel, 1994; Neely *et al.*, 2003; Gorga *et al.*, 2007). Neely *et al.* (2003) suggested that if DPOAE growth rate was defined as the slope of the DPOAE I/O function (in dB/dB), then compression could be viewed as the reciprocal of the growth rate of these cochlear responses. Consistent with I/O functions derived from direct BM measurements (e.g., Ruggero and Rich, 1991; Ruggero *et al.*, 1997; Rhode, 2007), DPOAE I/O functions exhibit nearly linear growth in response to low stimulus levels and compressive growth at moderate levels; the results can be represented by a two-line function (Neely *et al.*, 2009). When hearing loss is present, similar to physiological findings (Zurek *et al.*, 1982), the normally compressive portion of the function exhibits more linearity.

Much of the compression observed psychophysically is presumably due to compression in cochlear mechanics; however, there are likely to be differences between response growth at a single location, such as in the case of BM measurements, and the more spatially distributed cochlear responses that are represented in psychoacoustic measurements (Siegel *et al.*, 2005). If compression is viewed as the reciprocal of the growth rate of DPOAE I/O functions and of the FM PF-slope, a reasonable assumption is that DPOAE parameters might be mutually consistent with those observed in FM PF-slopes. The purpose of this study was to determine if the trends observed in FM PF-slopes could be predicted from DPOAE I/O functions.

II. METHODS

A. Subjects

Sixty subjects with ages ranging from 16 to 86 yr participated in this study. Twenty-five of these subjects had normal hearing while 35 had hearing loss. The hearing-impaired subjects had audiometric thresholds no greater than 40 dB HL (re ANSI, 2004). All subjects were recruited from a database of potential research subjects, which is maintained at BTNRH (Boys Town National Research Hospital); they were paid for their participation. In addition to inclusion criteria related to hearing sensitivity, described below, these subjects were selected for two reasons. First, they indicated that they would be willing to devote the amount of time that was required to collect the psychophysical (FM) and physiological (DPOAE I/O) data (about 7–9 h per subject). Second, they produced DPOAEs during a screening procedure and were able to perform the masked-threshold task. Additionally, subjects were required to have a normal 226-Hz tympanogram on each day on which DPOAE measurements were made. Behavioral thresholds were measured for octave and inter-octave audiometric frequencies from 0.25 to 8 kHz using routine clinical procedures. Subjects were assigned to hearing-loss categories (HLCs) from 0 to 40 dB HL at the two frequencies of interest (see below). Depending on hearing-loss configuration, data were collected at one or both frequencies. Only one ear of each subject was selected for study. Data are reported for 40 of the 60 subjects whose FM data yielded valid estimates of PF parameters.

B. Stimuli and apparatus

1. FM PFs

For the psychophysical experiments, the masker was set to 1 or 6 kHz, and masker level was held constant at levels ranging from 50 to 90 dB SPL (varied in 10-dB steps). The probe (f_p) was either a 1- or a 6-kHz 20-ms toneburst with 10-ms rise and fall Blackman windows, and no steady-state portion. The maskers were 300 ms in duration including rise- and fall-times of 2-ms. The masker-signal delay was set to 10 ms measured at the zero points of the waveforms. All stimuli for the psychophysical portion of the experiment were generated digitally via MATLAB at a sampling rate of 44 100 Hz and output with 24-bit resolution by a CardDeluxe (Digital Audio Labs, Minneapolis, MN) sound card. The headphone output of the soundcard was fed to a remote passive attenuator in a sound-treated room, and then to a Sennheiser HD 250 Linear II headphone (Wedemark, Germany).

2. DPOAE I/O functions

DPOAE I/O functions were measured for f_2 frequencies that were the same as those used in the psychophysical portion of the experiment (1 and 6 kHz). For all DPOAE measurements, response waveforms with duration of approximately 250 ms were averaged in two alternating buffers. These buffers were summed and, after a Fourier transformation, the frequency component in the $2f_1$ – f_2 bin was used to estimate DPOAE level. These two buffers were subtracted and the squared-magnitudes in the $2f_1$ – f_2 bin and the five bins above and below $2f_1$ – f_2 were averaged to provide an estimate of noise level. DPOAE stimuli were produced at a sampling rate of 32 000 Hz by a 24-bit soundcard (CardDeluxe, Digital Audio Labs) that drove a probe-microphone system (ER-10C, Etymotic Research, Elk Grove Village, IL). The “receiver equalization” of the ER-10C was bypassed to allow production of higher stimulus levels. Both channels of the soundcard and probe-microphone system were used during DPOAE measurements, with f_2 presented on one channel and f_1 presented on the other.

C. Procedure

1. FM PFs

A novel two-interval forced choice (2IFC) adaptive procedure called interpolated quadrant probability (IQP) was used to measure FMTs. The IQP procedure specifies a signal level for the next trial that is based on the entire track of previous trials. Trials are sorted in the order of increasing level and then divided into four ordered subsets or quadrants. After the average level and average-proportion correct are computed for each quadrant, the specified level is found by interpolating between the quadrant-average levels according to the quadrant-average probabilities. Exceptions are required when interpolation is not possible. If the desired probability is higher than the highest quadrant probability, then the specified level is one step above the previous level. If the desired probability is lower than the lowest quadrant probability, then the specified level is one step below the previous level. In order to implement these exceptions,

the procedure requires that a step size be specified, like the [Levitt \(1971\)](#) procedure. The step sizes specified were 4 and 2 dB. Unlike maximum likelihood, IQP makes no assumption about the shape of the PF and only assumes that it is monotonically increasing; it works by constructing a four-point approximation of the PF. Whenever the resulting PF is nonmonotonic, the number of points used to represent the PF is reduced until a monotone PF estimate is found. An advantage of IQP over traditional staircase methods ([Levitt, 1971](#)) is the ability to target any desired PC. FM PFs were obtained using IQP with three-interleaved tracks targeted at 76, 84, and 90 PC. Trials were presented in blocks of 180 with a pause every 45 trials to avoid fatigue or attention loss. Each trial consisted of a 500-ms warning interval, two 400-ms observation intervals separated by 500 ms, and a 300-ms feedback interval following the response of the subject. There was an interval of 500 ms before the beginning of the next trial. Subjects were given visual markers for the warning and observation intervals and correct-interval feedback in a message window on a keypad that they used to indicate their responses. In one interval, the masker and the signal were presented together, while in the other interval, the masker was presented alone. Four to eight 180-trial repetitions were obtained during training sessions. For data analysis, two 180-trial repetitions were obtained in each condition, for a total of 360 trials or 120 trials per track.

2. Quiet Thresholds

Prior to FM PF data collection, thresholds were obtained for both probe and masker using the IQP procedure described above with only one track (at 76 PC). Two blocks of 50 and 30 trials were collected for the probe and masker, respectively.

3. DPOAE I/O functions

DPOAE data were collected with locally developed software (EMAV; [Neely and Liu, 1994](#)). The level of f_2 (L_2) ranged from -5 to 90 dB SPL in 5-dB steps. Preliminary DPOAE measurements were used to optimize f_1 levels (L_1) at each frequency in each individual ear. At 6 kHz, the optimum L_1 was determined using a stimulus that follows a Lissajous path ([Neely et al., 2003](#)) using custom-designed software (SYSRES; [Neely and Stevenson, 2002](#)). At 1 kHz, because the signal-to-noise ratio (SNR) was lower, the optimum L_1 was determined using discrete stimulus levels (5-dB steps). The f_2/f_1 ratio was 1.22 for both f_2 frequencies. The L_1 levels obtained in this fashion were incorporated into a MATLAB program that generated an I/O parameter file specific to each subject. For each f_2 , a single I/O function was collected; data collection began at the highest L_2 and continued until the SNR was <3 dB.

III. RESULTS

A. FM PF fits

For a two-alternative, forced-choice task, d' is related to proportion correct by the cumulative-normal-distribution function,

$$P_c = \Phi\left(\frac{d'}{\sqrt{2}}\right). \quad (1)$$

The model of the FM PF adopted for this study is that $\log d'$ is approximately a linear function of signal level L_s within the range of signal levels near $d' = 1$. The data were initially subjected to group analysis and those results are presented first. When PF data are combined from all subjects at each frequency, and both masker level L_m and masker threshold in quiet θ_m are included, in addition to L_s , as dependent variables, then the linear relation among these variables is described by the following equation:

$$10 \cdot \log_{10}(d') = (a_0 + a_1 L_m + a_2 \theta_m) + (b_0 + b_1 L_m + b_2 \theta_m) \cdot L_s, \quad (2)$$

where a_0 , a_1 , a_2 , b_0 , b_1 , and b_2 are coefficients to be determined by linear regression. All trials from the three tracks were combined to fit FM PFs. After eliminating observations that had d' outside the range from 0.1 to 10, the linear model in Eq. (2) was fit to 591 observations at each frequency (1 and 6 kHz). Although the original model described in Eq. (2) was rejected because it accounted for only 9% and 4% of the variance at 1 and 6 kHz, respectively, the p -value for this regression was less than 0.001 at both frequencies. The significance of the p -value was further explored by reducing the initial model and is presented further below. Coefficients of the regression in Eq. (2) are listed in Table I. A notable difference between the coefficients at the two frequencies is that b_0 is 50% smaller at 6 kHz compared to 1 kHz. This difference indicates a trend of decreasing PF slope with increasing frequency, which is consistent with greater compression at higher frequencies. The fact that b_1 is negative at both frequencies indicates a trend for PF slopes to decrease as masker level increases, which is consistent with greater compression at higher stimulus levels. The fact that b_2 is positive indicates a trend for PF slopes to increase in subjects with greater amounts of hearing loss.

Based on the initial analysis [i.e., Eq. (2)], a model with fewer parameters was implemented to analyze individual data and those results are described here. When fitting data from individual subjects at each frequency, a reduced linear model was used with only four parameters instead of six,

$$10 \cdot \log_{10}(d') = (a_0 + a_1 L_m) + (b_0 + b_1 L_m) \cdot L_s. \quad (3)$$

TABLE I. Coefficients of a multiple linear regression for all listeners combined of $10 \cdot \log_{10}(d')$ onto dependent variables L_m , θ_m , L_s , $L_m \cdot L_s$, and $\theta_m \cdot L_s$.

Coefficients	Estimates		Pr(> t)	
	1 kHz	6 kHz	1 kHz	6 kHz
a_0	-7.769	-5.390	0.003	0.046
a_1	0.052	0.050	0.170	0.186
a_2	-0.121	-0.127	0.000	0.002
b_0	0.241	0.161	0.000	0.004
b_1	-0.002	-0.002	0.002	0.017
b_2	0.001	0.002	0.015	0.014

FIG. 1. (Color online) Examples of FM-PF fits to the individual PC data from the lowest (50 dB SPL) to the highest (90 dB SPL) masker-level conditions for three subjects. Each line represents a different masker level, with masker level increasing from left to right (i.e., masker levels in dB SPL: $\circ = 50$, $\triangle = 60$, $\nabla = 70$, $\diamond = 80$, $\times = 90$). FM-PF fits for 1 and 6 kHz maskers are shown in left and right panels, respectively. The top row shows PFs for one subject with HLC = 5 dB HL and the bottom row shows PFs for two subjects with HLC = 35 dB HL.

Dependence on θ_m was not included in this reduced model because masker threshold remains constant across observations for any given listener. On average, the reduced model accounted for 60% of the variance at each frequency for individual data. Four examples of linear fits to the data of individual listeners are shown in Fig. 1. Signal frequencies and subject identification numbers are indicated in the upper-left corner of each panel. Results for normal-impaired (HLC = 5) and hearing-impaired (HLC = 35) categories are represented in the top and bottom rows, respectively, with data for 1 and 6 kHz shown in left and right columns. Each line represents a different masker level, with masker level increasing from left to right (i.e., masker levels in dB SPL: $\circ = 50$, $\triangle = 60$, $\nabla = 70$, $\diamond = 80$, $\times = 90$). The top right panel demonstrates the expected trend for the PF slope to decrease as signal level increases in a subject with normal hearing, consistent with increasing compression with level. This pattern is associated with a negative value of b_1 in the reduced model; 75% of the fits at both frequencies showed this pattern, with the exceptions occurring primarily in subjects with larger degrees of hearing loss. The PF slopes in the bottom panels for two hearing-impaired subjects are steeper at all masker levels, suggesting loss of compression associated with hearing loss. This pattern is associated with smaller values of a_1 with larger degrees of hearing loss.

Distributions of the four parameters for each HLC are shown in Fig. 2. The a_1 parameter decreases with increasing hearing loss while b_1 increases. The mean and standard deviation of the four coefficients across all subjects are listed in

Table II. The coefficient values show no obvious difference between the two frequencies. A principal component analysis of the coefficients revealed a high degree of correlation, with essentially all of the variance being represented by the first coefficient, a_0 , alone.

If masked threshold T_s is defined as the signal level L_s at which $d' = 1$, then an equation for T_s can be derived from Eq. (3),

$$T_s = -\frac{a_0 + a_1 L_m}{b_0 + b_1 L_m}. \quad (4)$$

T_s reflects the horizontal spacing of the functions in Fig. 1 and is equivalent to the signal threshold parameter in Schairer *et al.* (2003) and Schairer *et al.* (2008).

B. DPOAE I/O-function fits

The DPOAE I/O functions were fit individually using the two-slope function described by Neely *et al.* (2009),

$$L_d = S_1 \cdot (L_2 - L_i) + 10 \cdot \log_{10} \left(10^{S_2 \cdot (L_d - L_o)/10} + 1 \right). \quad (5)$$

The two-slope fit has four parameters and required a nonlinear regression of DPOAE L_d level onto primary stimulus level L_2 . Two of the parameters (S_1, S_2) describe the slopes of the I/O function: S_1 is the slope of the lower-level, less compressive portion, while S_2 is the slope of the higher-level, more compressive portion. The other two parameters

FIG. 2. Distributions of the four parameters of the individual FM-PF fits as a function of hearing loss; results for 1 and 6 kHz, in left and right columns, respectively. Each box-and-whisker symbol indicates the median, inter-quartile range, and furthest non-outliers for selected distributions of a_0 (top panels), a_1 (second panels), b_0 (third panels), and b_1 (bottom panels).

(L_i, L_o) describe the location of the intersection of these two slopes along the L_2 axis: L_i is the value of L_2 at the breakpoint between linear and compressive slopes and L_o represents the value of L_d observed at L_i . Figure 3 shows an example of the two-slope fit to a DPOAE I/O function at $f_2 = 6$ kHz for a normal-hearing subject with 0 dB HL threshold. The circles represent L_d as a function of L_2 . The line is a two-slope fit to the I/O function. L_i (+ symbol) represents the input level (i.e., $L_2 = 50$ dB SPL) at the breakpoint between a near-linear slope (S_1) of the I/O function and a more compressive slope (S_2) at higher levels. In the model, L_o is the value of L_d (in this case, $L_d = 10$ dB SPL) where the breakpoint between linear and compressive slopes

occurs. The triangles represent noise levels. Distributions of the four parameters of the individual two-slope fits are shown in Fig. 4 for each HLC. The only apparent trends with HLC are for L_i and L_o at 6 kHz, where L_i increases and L_o decreases as a function of hearing loss. Consistent increases in slopes of the I/O functions with increasing hearing loss were not observed.

TABLE II. Average coefficients of multiple linear regressions for individual listeners of $10 \cdot \log_{10}(d')$ onto dependent variables L_m , L_s , and $L_m \cdot L_s$.

Coefficients	Mean		SD	
	1 kHz	6 kHz	1 kHz	6 kHz
a_0	-17.637	-31.265	32.891	31.837
a_1	-0.215	-0.055	0.538	0.495
b_0	0.833	0.996	0.665	0.698
b_1	-0.003	-0.005	0.009	0.009

FIG. 3. (Color online) Example of an I/O function fitted by the two-slope model. The solid line is superimposed on the DPOAE measurements at $f_2 = 6$ kHz in an ear with 0 dB HL threshold. The open triangles represent noise levels.

FIG. 4. Distributions of the four parameters of the individual two-slope fits as a function of hearing loss; results for 1 and 6 kHz, in left and right columns, respectively. Each box-and-whisker symbol indicates the median, inter-quartile range, and furthest non-outliers for selected distributions of S_1 (top panels), S_2 (second panels), L_i (third panels), and L_o (bottom panels). N for each distribution at 1 kHz (0–40 dB HL) is 4, 6, 11, 8, 8, 5, 4, 2, 4; N at 6 kHz (–5 to 40 dB HL) is 4, 5, 5, 5, 5, 5, 4, 4, 4, 4.

C. FM PF and DPOAE I/O-function comparison

The primary objective of this study was to assess the degree of correlation in individual ears between slopes of FM PFs and parameters that describe DPOAE I/O functions. DPOAE I/O functions with less than three points were not included in the analysis. Accordingly, a total of 35 and 40 functions at 1 and 6 kHz, respectively, were included. The scatterplots in Fig. 5 of every possible pair of FM PF and DPOAE I/O parameters at 6 kHz provide visual clues to possible relationships that may exist among these variables. Parameters a_0 and a_1 describe the horizontal position of the FM PFs along the signal level (L_s) axis, while parameters b_0 and b_1 describe FM PF-slope (see Fig. 1). Both the horizontal position and slope are linearly dependent on masker level. Parameters S_1 and S_2 describe the DPOAE I/O function slope at low levels and the decrease in slope at high levels, respectively, while parameters L_i and L_o describe the location of the breakpoint between these two slopes along the L_2 and L_d axes (see Fig. 3).

The clustering of points along the diagonal in all of the panels in the upper-left quadrant of Fig. 5 suggests strong correlations among the four FM PF parameters. In contrast, the relative lack of any pattern in the lower-left quadrant (or upper-right quadrant) suggests little correlation between

the FM-PF and DPOAE I/O sets of parameters. Because the ability to predict slopes of FM PFs from DPOAE I/O parameters was of particular interest, several linear regressions were considered. At 1 kHz, when four ears with $b_0 > 1.5$ were excluded from the analysis, thereby reducing the number of observations to 31, the regression that accounted for most of the variance (22%) was $b_1 \sim L_i + S_1$ (i.e., b_1 is linearly dependent on $L_i + S_1$), which had a p -value of 0.03. At 6 kHz, including all 40 observations, the regression that accounted for most of the variance (15%) was $b_1 \sim L_i$ (i.e., b_1 is linearly dependent on L_i), which had a p -value of 0.01. Although both of these regressions indicate a significant relationship between FM PF-slope and DPOAE I/O parameters, the variance accounted for by these regressions was less than anticipated. The 31 and 40 observations of b_1 used in these regressions at 1 and 6 kHz, respectively, are plotted in Fig. 6 as a function of L_i . In this figure, as with the linear regressions, there appears to be insufficient correlation to make a meaningful prediction of b_1 from L_i . Inasmuch as both estimates of compression have been purported to be functionally related to BM response growth, the regression results raise the possibility that FM PFs or DPOAE I/O functions (or both) might not be reliable estimates of compression.

FIG. 5. Scatterplots of correlations for all pairs of FM PF and DPOAE I/O parameters at 6 kHz: Parameters a_0 and a_1 describe the horizontal position of the FM PFs along the signal level (L_s) axis, while parameters b_0 and b_1 describe FM PF-slope, L_i (dB SPL) is the value of L_2 at the breakpoint between linear and compressive slopes of the DPOAE-I/O function, L_o is the value of L_d observed at L_i , S_1 is the lower-level slope of the DPOAE-I/O function, and S_2 is the higher-level, more compressive slope of the DPOAE-I/O function.

FIG. 6. Linear-regression coefficients of b_1 (dB/dB) as a function of L_i (dB SPL); results for 1 and 6 kHz are shown in left and right panels, respectively. At 1 kHz, the linear regression model was $b_1 \sim L_i + S_1$; at 6 kHz, the linear regression model was $b_1 \sim L_i$.

IV. DISCUSSION

A. FM PFs

The aim of this study was to determine whether trends in FM PFs might be predicted from DPOAE I/O functions. Although the use of FM PF-slopes seems like a reasonable alternative to other compression estimates (i.e., do not require assumptions regarding the decay of forward masking or the degree of off-frequency compression), it is not without controversy. Simulation studies (Leek, 2001) have suggested that the slopes of the PF estimated from adaptive tracking procedures tend to be too high unless the tracks are unusually long. Leek has suggested that adaptive procedures with more than two intervals, combined with a decision rule that estimates a higher point (79%) on the PF, provide more efficient and more accurate estimates of the PF slope. Kaernbach (2001) suggested that the source of the slope overestimation may be found in the adaptive algorithms themselves. Wichmann and Hill (2001) report that the lapse rate in trained observers ranges from 0% to 5% and this can have a significant effect on the slope parameters; to minimize this effect we allowed a pause every 45 trials. Another source of potential slope bias is the equation used to fit the PFs and estimate the slope parameter. The PF model represented by Eq. (3) was motivated by Neely and Jesteadt (2005), but is equivalent to the PF model proposed by Dai (1995) [and used by Schairer *et al.* (2003) and Schairer *et al.* (2008)] with $10 \cdot \log_{10}(x) = (a_0 + a_1 L_m)$ and $\beta = (b_0 + b_1 L_m)$. However, Dai's fitting method differs from ours in the relative weighting of points used in the regression.

In general, our FM data are consistent with previously reported findings (e.g., Jesteadt *et al.*, 1982) for similar stimulus parameters. For example, Jesteadt *et al.* found forward masking to be independent of frequency at 0.25, 1, and 4 kHz. These results are consistent with the slope coefficient values (b_0 and b_1) listed in Table II and depicted in Fig. 2, which are similar at 1 and 6 kHz.

Schairer *et al.* (2003) employed FM PFs to observe trends in compression growth rate for seven normal listeners. The on-frequency masker and signal (200- and 10-ms duration, respectively) was a 4-kHz pure tone. The masker was presented at masker-signal intervals of 0, 10, or 30 ms and varied from 30 to 90 dB SPL in 10-dB steps. They reported that PF slope appeared to be more highly correlated with masker level than with signal level. Although the high correlation between masker level and signal level made it difficult to separate the two effects, partial correlation of the PF slope was lower with signal level (-0.06) than with masker level (-0.71). Schairer *et al.* (2008) extended their study to investigate cochlear nonlinearity at low and high frequencies in NH and HI listeners. They obtained FM PFs for 0.25 and 4 kHz for on-frequency conditions in nine listeners: four NH and five HI. Signal and masker were pure tones of 10- and 200-ms duration, respectively; the masker was presented at 50, 70, and 90 dB SPL with a masker-signal interval of 10 ms. The purpose of their study was to compare relative compression between frequencies and groups of NH and HI, not to estimate specific compression values. Their findings revealed that signal level at threshold in on-frequency VS conditions increased as a function of masker level and PF

slopes decreased as a function of signal level for both frequencies. Our data reveal similar trends: for example, Fig. 1 illustrates the increase in PF threshold with masker level, the effect of compression on the FM PFs for a normal-hearing (5 dB HL) subject (top row) and the linearization of the PF-slope for two subjects with mild (35 dB HL) hearing loss (bottom row). Schairer *et al.* suggested that in some cases, the steeper functions in HI listeners might be the result of a reduced dynamic range, as a result of loss of amplification and compression mechanisms. None of our subjects had hearing loss greater than 40 dB HL; however, some subjects in the 30–40 dB HLCs were unable to track PFs for 50 and 60 dB SPL masker levels. Thus, we agree that the increased linearization observed in FM PFs of the subjects in the 25–40 dB HL group is most likely a function of compression loss and reduced dynamic range. Schairer *et al.* (2008) found that in their two VS on-frequency experiments, masker level and FM-PF threshold (i.e., signal level) were significantly correlated with each other and with PF slope. As mentioned above, multivariate analysis of our data yielded similar results.

Several studies (e.g., Lopez-Poveda, 2005; Plack *et al.*, 2004; Schairer *et al.*, 2008) have reported nearly normal compression in some of their HI listeners. Both Plack *et al.* (2004) and Lopez-Poveda *et al.* (2005) measured compression in listeners with moderate sensorineural hearing loss using the TMC procedure. Both noted that there was not a clear relationship between maximum amount of compression and absolute threshold and Lopez-Poveda *et al.* found that compression was close to normal in two of three ears. Similarly, our data did not reveal a systematic correspondence with absolute threshold and revealed a very slight decrease in PF slope as a function of PF threshold in only 5 of 35 HI listeners, all of whom were in the 30–35 dB HLCs. Plack *et al.* and Lopez-Poveda hypothesized that the presence of compression in listeners with mild-to-moderate hearing loss might involve undetected IHC damage. This would run counter to the standard assumption that when thresholds are elevated by less than 50 dB, most often the main cause of the loss is damage to the OHCs, with consequent loss or impairment of the active mechanism (Liberman *et al.*, 1986).

The effect of masker level as well as the correlation between signal level and masker level observed here and in Schairer *et al.* (2003) and Schairer *et al.* (2008) deserves consideration. Schairer *et al.* (2003) suggested that as masker level increased there might be an increase in internal noise that interferes with detection of the probe. It is possible that the effect of masker level might be due to both an increase in internal noise with concomitant variability in signal detection as well as a reduction in the internal representation of the signal due to a saturation process in the OHC cilia (for which there is no evidence at this time) and/or adaptation of the IHC synapse (Spassova *et al.*, 2004; Goutman and Glowatzki, 2007) as well as a more central process. As mentioned in the introduction, attempts have been made to explain psychophysical forward masking as a function of neural spike count (e.g., Relkin and Turner, 1988; Nelson *et al.*, 2009). Recently, Alves-Pinto *et al.* (2010) estimated FM thresholds in primary auditory area (AI) neurons using

signal detection theory (SDT) analysis. Their method of deriving detection thresholds takes into account not only the mean spike count of the response but also the variability in cortical responses, a procedure which approximates the behavioral process and allows for comparison with the psychophysical task. Their neurometric functions, equivalent to PFs, describe the PC responses as a function of probe level from which a masked threshold is derived (they used 60% correct criterion for masked threshold). The authors found that the increase in mean spike count in response to the probe was accompanied by an increase in variance and this variability followed Poisson statistics below the 60% criterion, whereas above 60%, as signal level increased, the variability did not follow Poisson statistics. That is, there was an increase in variability but not as big as would be expected if it followed Poisson statistics. Thus, the increase in detectability of a probe was observed to be mediated by increases in firing rates as signal level increased, and this increase was competing with increases in signal detection variability. Schairer *et al.* (2003) suggested a similar process by implicating internal noise effects. Alves-Pinto *et al.* summarized the main effect of a masker as reducing the spike count in response to the probe, directly determining the detectability of the probe. Accordingly, the effect of a forward masker on the response to a probe is, in most cases, similar to reducing the sound level of the probe. The authors also found that the effect of a short masker–signal interval is stronger the higher the masker level and, although in most of their stimulus conditions masker persistence did not seem to play a role, they suggested that the possibility that masker persistence might contribute to forward masking cannot be discounted. It would appear then that at lower masker levels, the correlation between signal level and masker level might not have as much impact as when both signal level and masker level are presented at higher stimulus levels.

B. DPOAE I/O functions

Previous DPOAE studies have employed different methods to model individual I/O functions (e.g., Boege and Janssen, 2002; Gorga *et al.*, 2003; Goldman *et al.*, 2006; Neely *et al.*, 2009). We applied the two-slope fit recommended by Neely *et al.* (2009) because it provides an estimate of response-growth rate that is not tied to threshold as well as parameters that might correlate with psychophysical measures. The findings reported here are in agreement with previous studies: NH subjects produced I/O functions characterized by compressive nonlinear regions over a range of moderate levels while HI subjects produced steeper I/O slopes. As hearing loss increased, the breakpoint (i.e., L_i) between the two slopes increased and L_o decreased, consistent with previously reported findings (Dorn *et al.*, 2001, Gorga *et al.*, 2003, Neely *et al.*, 2009); however, these trends were apparent only for 6 kHz. The decrease in output (L_o) as hearing loss increases is not surprising and has been observed in previous studies. As seen in Fig. 4, the median values observed for S_2 , the compressive portion of the two-line fit, were smaller at 6 kHz than at 1 kHz, but, due to the variability, the difference was not statistically significant.

The correlations between the parameters of the FM PFs and DPOAE I/O functions shown in Fig. 5 were too weak to be able to permit meaningful predictions about the PF slope. Although many observations suggest that DPOAE, psychophysical estimates of compression, and BM compression growth rates are at least qualitatively similar, we would not expect this relationship to be exact. The weak correlation between individual FM PFs and DPOAE I/O functions was unexpected, however, and given the qualitative correspondence between the latter and BM measurements, casts doubts regarding the use of FM PFs to estimate compression. On the other hand, the apparent variability in the parameters that quantified the DPOAE I/O functions may have been a factor that made it difficult to observe a relation between psychophysical and acoustical estimates of compression.

At present, we do not know how well either FM-PF or DPOAE I/O compression estimates compare with BM compression. Much of the compression observed psychophysically is presumably due to compression in cochlear mechanics; however, there likely are differences between response growth at a single location, such as in the case of BM measurements, and the more spatially distributed cochlear responses that are represented in psychoacoustic measurements. Psychoacoustic measurements are not the only measurements besieged by procedural issues. At present, DPOAE generation seems to be less *local* than BM measurements and less *spatially distributed* than psychoacoustic measurements (Siegel *et al.*, 2005; Dong and Olson, 2010; Martin *et al.*, 2010); thus, they too might be subject to similar effects observed in psychophysical measures such as forward masking. If, as Martin *et al.* have recently suggested, there are basal DPOAE components that influence standard measures of DPOAEs, then it is plausible that these components might have influenced compression estimates for some hearing-impaired subjects, particularly those with low-frequency ascending hearing-loss configurations. As part of this study, we collected DPOAEs for four frequencies (i.e. 1, 2, 4, and 6 kHz); however, we report data for only 1 and 6 kHz because those were the two frequencies for which we collected FMTs. It is noteworthy that in a few instances, in the HI group, the compression estimates at lower frequencies (i.e., 1 and 2 kHz) exhibiting more hearing loss had stronger compression than higher frequencies with better hearing. Closer examination revealed that in those instances the hearing threshold an octave above was better. This raises the possibility that if indeed basal-generated components exert an influence on standard DPOAE measures, then that would explain the apparent discrepancy. However, since our measurements were not obtained in the same fashion as in Martin *et al.*, these comments remain speculative. On the other hand, these observations underscore the conclusion that there is a correlation between FMTs and DPOAE compression estimates but methodological as well as analysis issues need to be resolved before direct comparisons between the two can be made. Moreover, the influence of these basal components might underlie some of the variability observed in previous efforts aimed at relating behavioral threshold to DPOAE measures.

V. CONCLUSIONS

The goal of this experiment was to measure trends in FM PF-slopes that might be predicted from DPOAE I/O functions. The results reveal insufficient correlation to make a meaningful prediction of compression growth rate in FM PF-slope from DPOAE I/O data. The results suggest that one or both of these measures may provide unreliable indicators of BM compression.

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