Original Article

Trends in Amplification
2014, Vol. 18: 1–14

DOI: 10.1177/2331216514564253
tia.sagepub.com
SAGE

Forward-Masking Recovery and the Assumptions of the Temporal Masking Curve Method of Inferring Cochlear Compression

Patricia Pérez-González[1,2], Peter T. Johannesen[1,2], and Enrique A. Lopez-Poveda[1,2,3]

Abstract

The temporal masking curve (TMC) method is a behavioral technique for inferring human cochlear compression. The method relies on the assumptions that in the absence of compression, forward-masking recovery is independent of masker level and probe frequency. The present study aimed at testing the validity of these assumptions. Masking recovery was investigated for eight listeners with sensorineural hearing loss carefully selected to have absent or nearly absent distortion product otoacoustic emissions. It is assumed that for these listeners basilar membrane responses are linear, hence that masking recovery is independent of basilar membrane compression. TMCs for probe frequencies of 0.5, 1, 2, 4, and 6 kHz were available for these listeners from a previous study. The dataset included TMCs for masker frequencies equal to the probe frequencies plus reference TMCs measured using a high-frequency probe and a low, off-frequency masker. All of the TMCs were fitted using linear regression, and the resulting slope and intercept values were taken as indicative of masking recovery and masker level, respectively. Results for on-frequency TMCs suggest that forward-masking recovery is generally independent of probe frequency and of masker level and hence that it would be reasonable to use a reference TMC for a high-frequency probe to infer cochlear compression at lower frequencies. Results further show, however, that reference TMCs were sometimes shallower than corresponding on-frequency TMCs for identical probe frequencies, hence that compression could be overestimated in these cases. We discuss possible reasons for this result and the conditions when it might occur.

Keywords

cochlear nonlinearity, DPOAEs, otoacoustic emissions, middle-ear muscle reflex

Introduction

The mammalian cochlea compresses a wide range of sound pressure levels (SPLs) into a narrower range of mechanical responses. The amount of compression and the range of SPLs over which compression occurs depend on outer hair cell (OHC) function (Robles & Ruggero, 2001). Cochlear compression is thought to determine important auditory percepts such as absolute hearing threshold, the dynamic range of hearing, or auditory masking (Oxenham & Bacon, 2003, 2004). Detailed measurements of cochlear compression could thus be useful to diagnose hearing impairment (Lopez-Poveda & Johannesen, 2012; Plack, Drga, & Lopez-Poveda, 2004), to understand the impact of hearing loss on auditory perception (Bacon & Oxenham, 2004), or to fit hearing aids (Meddis, Lecluyse, Tan, Panda, & Ferry, 2010; Panda, Lecluyse, Tan, Jurgens, & Meddis, 2014). In humans, peripheral compression cannot be measured directly and so a number of psychoacoustical methods have been developed to infer it (Lopez-Poveda

[1]Instituto de Neurociencias de Castilla y León, Universidad de Salamanca, Salamanca, Spain
[2]Grupo de Audiología, Instituto de Investigación Biomédica de Salamanca, Salamanca, Spain
[3]Departamento de Cirugía, Facultad de Medicina, Universidad de Salamanca, Salamanca, Spain

Corresponding author:
Enrique A. Lopez-Poveda, University of Salamanca, Calle Pintor Fernando Gallego 1, Salamanca 37007, Spain.
Email: ealopezpoveda@usal.es

& Alves-Pinto, 2008; Lopez-Poveda, Plack, & Meddis, 2003; Nelson, Schroder, & Wojtczak, 2001; Oxenham & Plack, 1997; Plack & Arifianto, 2010; Plack & O'Hanlon, 2003; Plack & Oxenham, 2000; Yasin, Drga, & Plack, 2013). These psychoacoustical techniques are indirect and hence are based on a number of assumptions. Here, we test the assumptions of a technique known as the temporal masking curve (TMC) method.

The TMC method (Nelson et al., 2001) consists of measuring the level of a tonal forward masker required to just mask a fixed tonal probe as a function of the time interval between the masker and the probe. A TMC is a graphical representation of the resulting masker levels against the corresponding masker-probe intervals. Because the probe level is fixed, the masker level increases with increasing masker-probe time interval and hence TMCs have positive slopes. Nelson et al. (2001) argued that the slope of any given TMC depends simultaneously on the amount of basilar membrane (BM) compression affecting the masker at a cochlear place whose characteristic frequency (CF) equals approximately the probe frequency *and* on the rate of recovery from the internal (postmechanical or compression free) masker effect. By assuming that the postmechanical recovery rate is the same across masker frequencies, BM input/output functions may be estimated by plotting the masker levels of a reference TMC (i.e., the TMC for a masker that is processed linearly by the cochlea) against the levels for any other masker frequency, paired according to masker-probe delays (Nelson et al., 2001).

In their original study, Nelson et al. (2001) used a masker frequency about an octave below the probe frequency as the reference TMC on the grounds that BM responses are linear for tones well below the CF. Lopez-Poveda et al. (2003) argued that the latter is true for basal cochlear regions but not for apical cochlear regions (Rhode & Cooper, 1996) and proposed inferring compression at low probe frequencies by using a reference TMC for a high probe frequency. This version of the TMC method has been used in many studies (e.g., Johannesen & Lopez-Poveda, 2008; Johannesen, Pérez-González, & Lopez-Poveda, 2014; Jurgens, Kollmeier, Brand, & Ewert, 2011; Lopez-Poveda & Alves-Pinto, 2008; Lopez-Poveda & Johannesen, 2012; Lopez-Poveda, Plack, Meddis, & Blanco, 2005; Nelson & Schroder, 2004; Panda et al., 2014; Plack & Drga, 2003; Plack et al., 2004). An implicit assumption of this approach is that the postmechanical rate of recovery from forward masking is independent of probe frequency.

The TMC method thus rests on two assumptions regarding the postmechanical rate of recovery from forward masking: (a) for a given probe frequency, it is independent of masker frequency; and (b) it is independent of probe frequency. These assumptions are controversial. Wojtczak and Oxenham (2009) questioned the first assumption by showing that the rate of postmechanical recovery is actually faster when masker and probe frequencies are equal (on-frequency condition) than when the masker frequency is about an octave below the probe frequency (reference condition). Given that masker levels are typically higher for the reference than for the on-frequency TMC, an alternative explanation for their findings is that forward-masking recovery is actually dependent upon masker level rather than masker frequency. Indeed, a third, less explicit assumption of the TMC method is that forward-masking recovery is independent of masker level. Wojtczak and Oxenham concluded that for normal-hearing listeners, the first assumption of the TMC method held for masker levels below 83 dB SPL but not for higher levels.

Stainsby and Moore (2006) questioned the second assumption of the TMC method. They showed that for hearing-impaired listeners with nearly absent distortion product otoacoustic emissions (DPOAEs), and hence presumably linear cochlear responses, TMCs are steeper for low than for high probe frequencies. On the other hand, other authors have provided experimental support for the second assumption using other psychoacoustical methods that do not require a reference TMC (Lopez-Poveda & Alves-Pinto, 2008; Plack et al., 2008). In an attempt to reconcile these seemingly disparate findings, Lopez-Poveda and Alves-Pinto (2008) argued that "absence of measurable DPOAEs at low frequencies is not necessarily indicative of linear cochlear responses because it is hard to measure DPOAEs at low frequencies due to physiological and ambient noise" (p. 1553). In other words, Lopez-Poveda and Alves-Pinto were suggesting that the absence of DPOAEs in the subjects used by Stainsby and Moore could be more apparent than real due to their using insufficiently sensitive DPOAE techniques. Indeed, Stainsby and Moore used primary tones with a single level of 70 dB SPL each even though DPOAEs depend strongly on the levels of the primary tones (e.g., Figure 7 in Lopez-Poveda & Johannesen, 2009), particularly at low frequencies (Figures 1 and 2 in Johannesen & Lopez-Poveda, 2010). Therefore, it is conceivable that DPOAEs may have appeared as *absent* to Stainsby and Moore but might have been present if they had used different primary levels. In addition, Stainsby and Moore used a fixed measurement time of 2 s across test frequencies even though DPOAE detectability increases with increasing measurement time (on average, the DPOAE signal-to-noise ratio improves by 3 dB for every doubling of the measurement time; e.g., see Figure 1 in Zurek, 1992; see also Figure 1 in Popelka, Osterhammel, Nielsen, & Rasmussen, 1993). The use of short recording times can hinder DPOAE detectability more at low

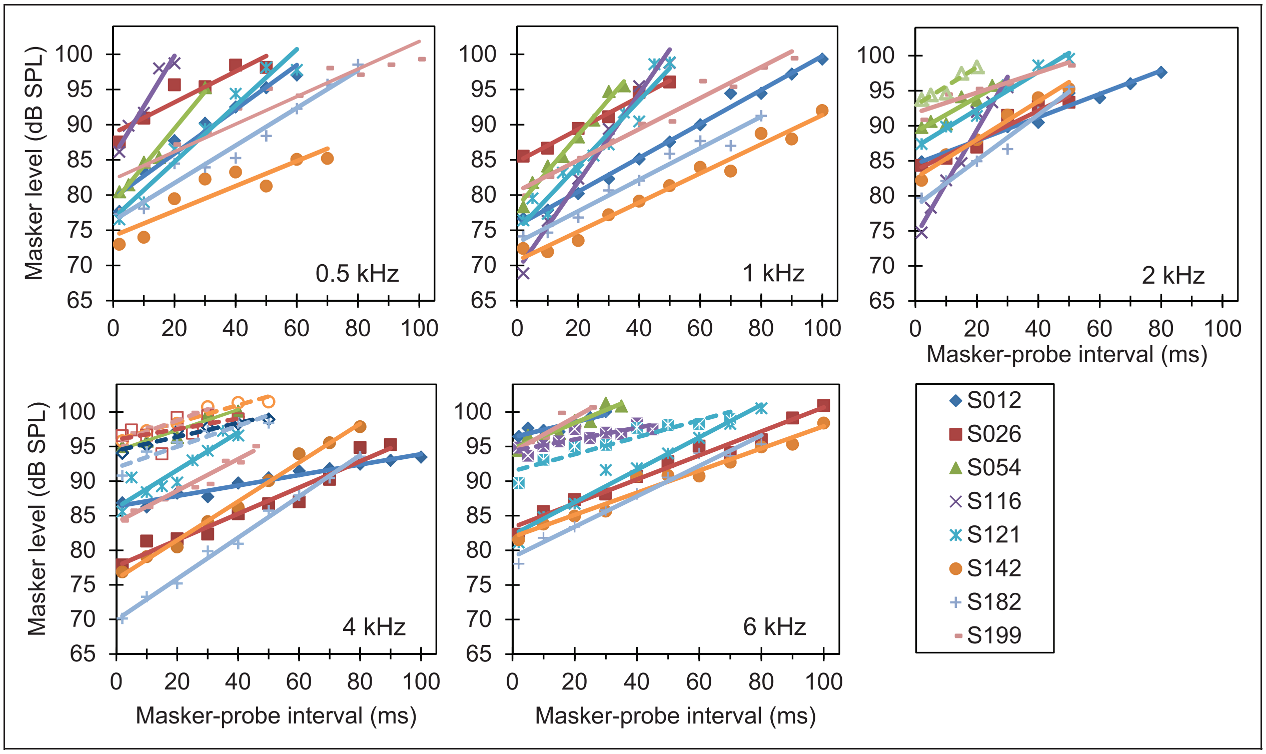

Figure 1. Experimental (symbols) and fitted (lines) TMCs. Each panel illustrates on-frequency TMCs (filled symbols, continuous lines) for a different probe frequency, as indicated by the numbers at the bottom-right corner of the panel. A panel also illustrates linear reference TMCs (dashed lines) if they were measured at the corresponding probe frequency.

frequencies where the physiological noise is comparatively higher. Therefore, it is conceivable that DPOAEs may have appeared as *absent* to Stainsby and Moore but might have been present if they had used longer recording times. An additional concern about the study of Stainsby and Moore is that they used only three subjects.

The aims of the present study were to revisit the two main assumptions of the TMC method using an approach inspired by Stainsby and Moore (2006) but with a larger sample size and improved methods to maximize DPOAE detectability.

Methods

Approach

The aim was to test if the postmechanical (i.e., compression free) rate of recovery from forward masking is independent of masker level and of probe frequency. To do it, we called back 68 hearing-impaired listeners who had participated in a related TMC study (Johannesen et al., 2014) and measured DPOAEs in these subjects at four test frequencies (0.5, 1, 2, and 4 kHz) and using eight different primary levels at each test frequency. Whenever possible, DPOAE primary levels were individually optimized to maximize DPOAE levels. Of the 68 listeners, we chose eight who showed absent or nearly absent DPOAEs and we assumed that they had linear cochlear responses. We analyzed the already available on-frequency TMCs for those listeners at test frequencies of 0.5, 1, 2, 4, and 6 kHz as well as their reference TMCs seeking correlations of TMC slope with probe frequency and masker level. Importantly, we overcame the limitations of the study by Stainsby and Moore (2006) by using a larger sample size ($N = 8$ vs. $N = 3$) and using improved DPOAE methods. Specifically, over a wide level range, we searched combinations of primary levels that maximize DPOAEs independently at each frequency compared with Stainsby and Moore who used primaries with only a fixed level of 70 dB SPL each; and we used longer measurement times of 30 s at 500 Hz and 10 s at higher frequencies compared with the 2s used by Stainsby and Moore. By including these improvements, we maximized the chance of detecting DPOAEs above the noise floor that might otherwise be missed, particularly at low frequencies. In other words, we were more confident that the lack of DPOAEs as observed using our methods was a better indicator of linear cochlear responses than a lack of DPOAEs as observed using the methods of Stainsby and Moore.

Experimental procedures were approved by the Ethics Review Board of the University of Salamanca. Informed consent was obtained from all participants.

DPOAE Measurements

Pairs of primary pure tones with frequencies (f_1, f_2) and corresponding levels (L_1, L_2) were presented, and the level of the $2f_1-f_2$ frequency component of the otoacoustic emission in the ear canal was recorded and regarded as the DPOAE level. DPOAEs were measured for f_2 of 0.5, 1, 2, and 4 kHz and for L_2 values from 35 to 70 dB SPL, in 5-dB steps (4 test frequencies × 8 levels = 32 conditions). For each test frequency f_2, f_1 was set equal to f_2/1.2. An attempt was made to individually set L_1 so as to maximize DPOAE levels. For L_2 values of 50 and 65 dB SPL, we empirically sought the L_1 value that maximized the DPOAE level, if any. When a pair of L_1 values was

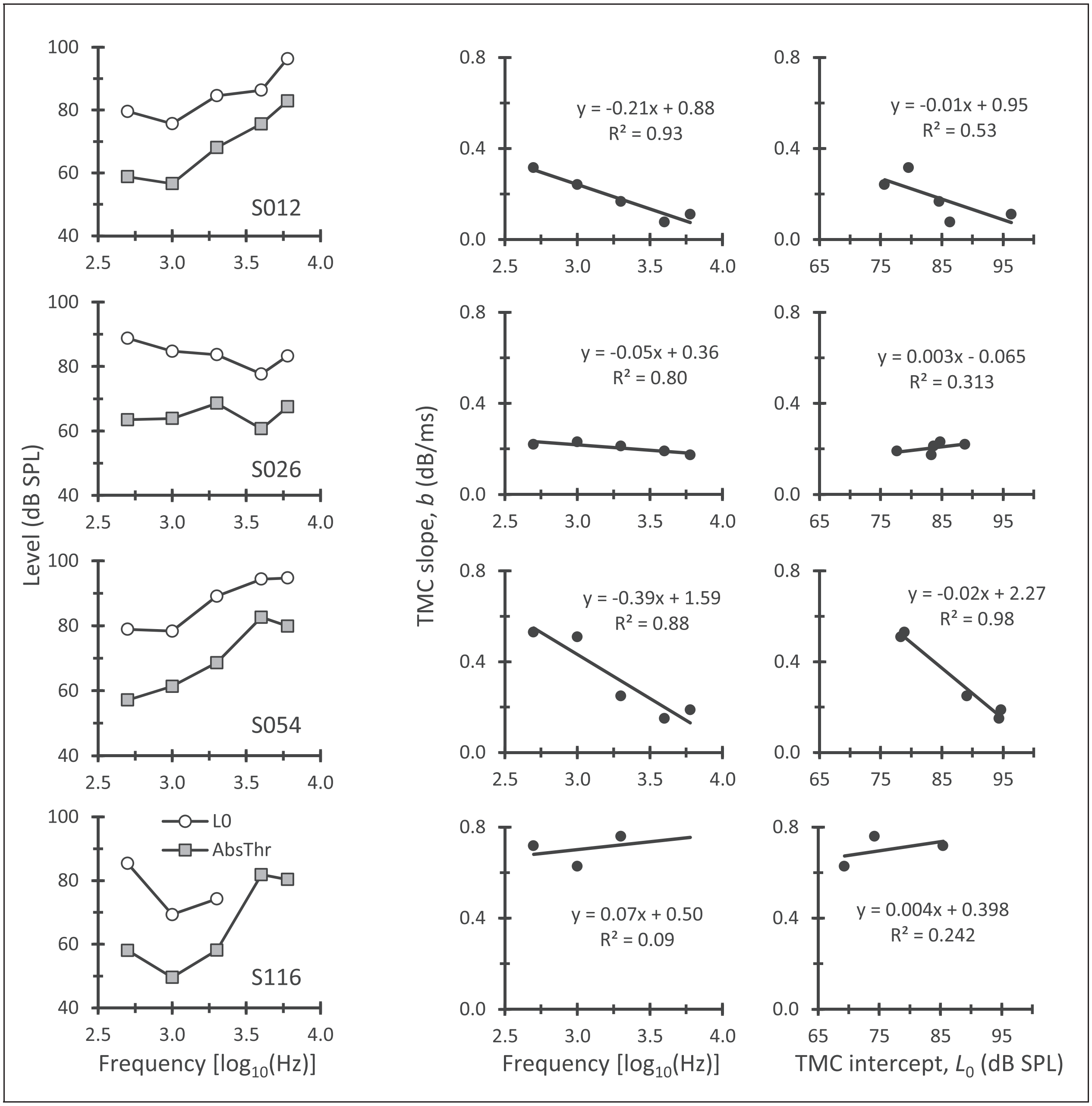

Figure 2. On-frequency TMC characteristics for four participants (S012, S026, S054, and S116). Left: Masker absolute thresholds (gray squares) and TMC intercept levels (L_0, open circles). Middle: TMC slope, *b*, as a function of frequency. Right: TMC slope, *b*, as a function of intercept level, L_0. Each row is for a different participant, as indicated in the bottom-right corner of the left panels. Straight lines and equations in the middle and right panels illustrate linear regression fits together with their corresponding equations and proportion of predicted variance (R^2).

found, then the values of L_1 for the other L_2 levels were obtained using linear regression. When individually optimal L_1 values were not found, we used the primary level rule of Neely, Johnson, and Gorga (2005) because it has been independently confirmed that this is the most appropriate rule to maximize DPOAE levels on average (cf. Figure 7 in Lopez-Poveda & Johannesen, 2009).

DPOAE measurements were obtained using an Intelligent Hearing System's Smart device (with SmartOAE software version 4.52) equipped with an Etymotic ER-10D probe. During the measurements, participants sat comfortably in a double-wall sound attenuating chamber and were asked to remain as steady as possible. The probe fit was checked before and after each recording session. The probe remained in the participant's ear throughout the whole measurement session to avoid measurement variance from probe fit. DPOAEs were measured for a preset measurement time of 30 s for $f_2 = 500$ Hz and 10 s for other f_2 frequencies. A DPOAE measurement was regarded as valid when it was 6 dB above the measurement noise floor (defined as the mean level over 10 frequency bins adjacent to the $2f_1{-}f_2$ component in the OAE spectrum). When a response did not meet this criterion, the measurement was repeated. If the required criterion was not met for at least two of three successive tries, we concluded that DPOAEs were absent for that condition.

DPOAE measurements were regarded as valid only when they were 6 dB above the system's artifact response. The rationale behind this rather strict criterion and the details of the procedure for controlling for system's artifacts and calibration can be found elsewhere (Johannesen & Lopez-Poveda, 2008).

Participants

Eight listeners participated in the study (Table 1). They were selected from a sample of 68 listeners with symmetrical sensorineural hearing loss and no sign of middle-ear pathology (Johannesen et al., 2014). These eight listeners were selected because (a) they had absolute thresholds equal to or higher than 40 dB HL (American National Standards Institute, 1996) at the TMC test frequencies, a criterion also met by the three subjects of Stainsby and Moore (2006); and (b) they had absent DPOAEs for at least 29 of the 32 conditions (4 test frequencies × 8 primary levels) that were attempted, as shown in Table 2.

Middle-Ear Muscle Reflex Measurements

As a control, the threshold of activation of the middle-ear muscle reflex (MEMR) was measured using a clinical middle-ear analyzer (Interacoustics AT235h). Middle-ear compliance for a probe tone of 226 Hz and 85 dB SPL was measured in the presence and in the absence of ipsilateral MEMR elicitor tones with frequencies 500, 1000, 2000, and 4000 Hz and levels 75 to 100 dB HL in 5-dB steps. The MEMR activation threshold was regarded as the lowest elicitor level that evoked a detectable change in middle-ear compliance (re the non-elicitor condition) minus 2.5 dB, that is, minus half the elicitor intensity step. Measured MEMR activation thresholds are shown in Table 1.

Table 1. Participants' Data.

				Reference TMC	MEMR activation threshold (dB SPL)			
Participant	Sex	Age	Ear	(f_P, f_M)	500	1000	2000	4000
S012	M	80	Left	4, 1.6	n.p.	n.p.	n.p.	n.p.
S026	F	51	Left	4, 1.6	96	95	?	?
S054	M	79	Left	2, 0.8	106	?	n.p.	n.p.
S116	M	53	Right	6, 2.4	n.p.	105	106.5	n.p.
S121	M	60	Left	6, 2.4	96	95	106.5	n.p.
S142	M	51	Right	4, 2.0	101	95	96.5	104.5
S182	F	55	Right	4, 1.6	101	90	101.5	99.5
S199	F	73	Left	4, 2.0	?	?	?	?

Note. TMC = temporal masking curve; MEMR = middle-ear muscle reflex; n.p. = MEMR not present; ? = MEMR could not be measured reliably. Age is in years. Also shown is (a) the linear reference TMC condition measured for each listener, expressed as a pair of probe and masker frequencies (f_P, f_M) in kHz; and (b) the threshold of activation of the MEMR for different eliciting frequencies.

Table 2. DPOAE Levels (dB SPL) for Those Participants (S#), Test Frequencies (Columns), and L_2 Levels (Rows) for Which DPOAEs Were Measurable.

	F_2 (kHz)							
	0.5		1		2		4	
L_2 (dB SPL)	S#	dB SPL	S#	dB SPL	S#	dB SPL	S#	dB SPL
70	S054	−1.5						
65	S199	−1.7	S116	−4.0	S199	3.9	S026	10.6
							S199	−0.3
60	S116	2.8						
	S121	7.5						
55	S054	−7.8	S012	−3.5				
	S142	1.8						
50			S142	−9.4				
45							S026	−6.0
40			S012	−5.6	S054	−14.1	S142	−6.3
					S121	−9.6		
35			S116	−5.2	S182	−14.1		

Note. DPOAE = distortion product otoacoustic emission. Missing values indicate absent DPOAEs.

TMC Measurements

Temporal masking curves for the eight selected listeners were taken from a previously published study (Johannesen et al., 2014). Procedures are fully described in that study and hence only a summary is given here.

On-frequency TMCs were measured for probe frequencies (f_P) of 0.5, 1, 2, 4, and 6 kHz. Maskers and probes were sinusoids. The duration of the maskers was 210 ms including 5-ms cosine-squared onset and offset ramps. Probes had durations of 10 ms, including 5-ms cosine-squared onset and offset ramps with no steady-state portion, except for the 500-Hz probe, whose duration was 30 ms with 15-ms ramps and no steady-state portion. The level of the probes was fixed at 10 dB above the individual absolute threshold for the probe. Masker-probe time intervals, defined as the period from masker offset to probe onset, ranged from 10 to 100 ms in 10-ms steps with an additional gap of 2 ms. Masker levels sometimes reached the maximum permitted sound level output (105 dB SPL) for time intervals shorter than the maximum 100 ms. If the number of measured data points was insufficient for curve fitting (see later), masker levels were measured for additional intermediate intervals (e.g., 5, 15, 25 ms).

A single reference TMC was measured for each participant. The reference TMC was for a probe frequency of 2, 4, or 6 kHz and for a masker frequency equal to $0.4f_P$ or $0.5f_P$. The selection of the reference condition depended on the participant's hearing loss at the reference probe frequency and on the maximum permitted sound level output (105 dB SPL). Following the indications of earlier studies (Lopez-Poveda & Alves-Pinto, 2008; Lopez-Poveda et al., 2003), the reference conditions were sought in the following order of priority (Johannesen et al., 2014): (4, 1.6), (4, 2), (6, 2.4), (6, 3), (2, 0.8), (2, 1), where the numbers in each pair denote probe and masker frequency in kHz, (f_P, f_M), respectively. Table 1 shows the reference TMC conditions measured for each participant.

TMC Analysis

As in many previous studies (e.g., Lopez-Poveda et al., 2003, 2005; Nelson & Schroder, 2004; Nelson et al., 2001; Plack et al., 2004; Stainsby & Moore, 2006), TMCs were fitted using a straight line:

$$L_M(t) = L_0 + b \cdot t \tag{1}$$

where $L_M(t)$ is the masker level (in dB SPL) at masker-probe time interval t (in ms), b is the TMC slope (dB/ms), and L_0 is the intercept masker level (in dB SPL) for a masker-probe time interval of 0 ms. Given that the selected participants presumably had linear cochlear responses, parameter b was taken as indicative of forward-masking recovery rate. L_0 was used as indicative of the range of masker levels in a TMC.

Results

Distortion product otoacoustic emissions

Table 2 gives the DPOAE levels measured for each participant for each pair of test frequency, f_2, and primary level, L_2. Missing values indicate absent DPOAEs. DPOAEs were present for only 19 of the 256 possible cases (4 test frequencies × 8 primary levels × 8 participants). Furthermore, for no participant were DPOAEs present in more than 3 out of 32 conditions. The noise floor level was less than −4 dB SPL for all participants at 0.5 and 1 kHz, except for S121 at 500 Hz, for whom the noise floor was −1 dB SPL. The average noise floor level was −8.19 and −13.50 dB SPL at 0.5 and 1 kHz, respectively, and lower at higher frequencies. Altogether, these results suggest that the absence of DPOAEs for these participants is not due to high levels of noise. Therefore, we concluded the absence was due to their having linear (or almost linear) cochlear responses over the frequency range from 0.5 to 4 kHz. The accuracy of this conclusion will be discussed later.

Temporal masking curves

Figure 1 shows experimental (symbols) and fitted (lines) TMCs. On-frequency and reference fitted TMCs are illustrated using continuous and dashed lines, respectively. Note that 46 TMCs were measured (38 on-frequency plus 8 reference TMCs) and that on-frequency TMCs are missing for S116 at 4 and 6 kHz (Table 3). For S116, probe thresholds were so high at 4 kHz that we anticipated masker levels would be higher than the maximum system output level. Hence, we did not attempt measuring on-frequency TMCs at 4 kHz. For S116, we tried measuring on-frequency TMCs at 6 kHz but masker levels exceeded the maximum system output. Except for one case, missing points in Figure 1 are indicative that the corresponding masker levels would exceed the maximum system output level (105 dB SPL). The exception is the on-frequency TMC for S142 at 0.5 kHz. This TMC was nonmonotonic (i.e., masker levels decreased with increasing masker-probe time interval beyond 70 ms), probably because the subject had greater difficulty at keeping track of the probe for the longer masker-probe time intervals. We regarded the nonmonotonic trend as unrealistic and omitted the declining portion of the TMC.

Table 3 gives the parameters and goodness-of-fit statistics of the straight line fits to the TMCs (root mean square, RMS, errors and proportion of variance

Table 3. Linear Regression Parameters and Goodness-of-Fit for Each Subject (S#) and TMC.

		Linear regression model (equation (1))				
S#	kHz	L_0 (dB SPL)	b (dB/ms)	R^2	RMS (dB)	Num. points
S012	0.5	79.6	0.32	0.95	1.40	7
S026	0.5	88.8	0.22	0.86	1.45	6
S054	0.5	78.9	0.53	0.98	0.77	5
S116	0.5	85.4	0.72	0.94	1.13	5
S121	0.5	76.7	0.40	0.95	1.73	7
S142	0.5	74.2	0.18	0.83	1.80	8
S182	0.5	76.4	0.27	0.97	1.29	9
S199	0.5	82.3	0.20	0.89	2.18	11
S012	1	75.7	0.24	0.99	0.67	11
S026	1	84.7	0.23	0.99	0.41	6
S054	1	78.3	0.51	0.98	0.75	8
S116	1	69.3	0.63	0.98	1.29	6
S121	1	75.0	0.46	0.94	1.76	11
S142	1	70.7	0.21	0.97	1.07	11
S182	1	73.2	0.22	0.97	1.00	9
S199	1	80.5	0.22	0.97	1.02	10
S012	2	84.6	0.17	0.97	0.70	9
S026	2	83.7	0.21	0.94	0.88	6
S054	2	89.1	0.25	0.91	0.75	7
S116	2	74.2	0.76	0.97	1.24	7
S121	2	87.0	0.27	0.98	0.70	6
S142	2	82.6	0.27	0.98	0.70	6
S182	2	78.6	0.33	0.98	0.78	6
S199	2	91.9	0.14	0.90	0.78	6
S012	4	86.3	0.08	0.96	0.49	11
S026	4	77.7	0.19	0.96	1.16	10
S054	4	94.4	0.15	0.91	0.65	5
S116	4					
S121	4	86.1	0.27	0.86	1.39	9
S142	4	75.8	0.28	0.99	0.66	9
S182	4	69.9	0.30	0.99	0.63	9
S199	4	84.0	0.23	0.96	0.64	10
S012	6	96.3	0.11	0.82	0.49	7
S026	6	83.3	0.17	0.97	0.88	11
S054	6	94.7	0.19	0.91	0.65	8
S116	6					
S121	6	82.2	0.24	0.97	0.97	9
S142	6	81.9	0.16	0.96	0.97	11
S182	6	79.0	0.22	0.97	0.95	9
S199	6	94.2	0.25	0.80	1.00	6
S012	LR-4	94.4	0.10	0.92	0.49	6
S026	LR-4	96.0	0.08	0.29	1.44	8
S054	LR-2	92.9	0.27	0.92	0.51	5
S116	LR-6	94.4	0.08	0.74	0.69	10
S121	LR-6	91.4	0.12	0.87	1.07	8
S142	LR-4	96.0	0.13	0.93	0.56	6
S182	LR-4	92.0	0.15	0.88	0.93	6
S199	LR-4	95.5	0.16	0.91	0.49	7

(continued)

Table 3. Continued

Note. TMC = temporal masking curve; LR = linear reference. Each Line is for a different on-frequency or LR TMC. RMS is the root-mean-square error in decibels; R^2 is the proportion of variance explained by the linear regression model. Empty cells indicate that the corresponding TMC was not available (see main text).

explained, R^2). The variance explained by the fit was $\geqslant$90% for 36 of the 46 measured TMCs, between 80% and 90% for eight TMCs, and 74% and 29% for the remaining two TMCs. The RMS error was always less than 2.2 dB, with a mean value of 0.96 dB. These statistics justify the use of a linear regression model (Equation 1) to analyze the present TMCs.

Forward-Masking Recovery as a Function of Probe Frequency and TMC Intercept Level

The middle panels of Figures 2 and 3 show the slope of on-frequency TMCs (i.e., parameter b in equation (1)) as a function of probe frequency expressed as $\log_{10}$(Hz); the rightmost panels show TMC slope as a function of TMC intercept level (i.e., parameter L_0 in equation (1)). The leftmost panels in the two figures illustrate masker absolute thresholds and TMC intercept levels as a function of frequency. Each row of each figure shows results for an individual participant, as indicated in the leftmost panels of each figure. For some participants, TMC slope decreased with increasing frequency and with increasing L_0, while for other participants TMC slope remained approximately constant across frequencies and L_0. Linear regression functions were fitted to the trends in Figures 2 and 3. These are shown as straight lines together with their corresponding equations and proportion of explained variance (R^2). Note that a low value of R^2 does not necessarily imply a poor linear regression fit; indeed, low R^2 values also occur when TMC slope remains constant across frequencies or intercept levels.

Figure 4 shows the slopes of the linear regression fits for each participant. Different symbols illustrate the slope of the linear regression trends for frequency (triangles) and L_0 (circles). Negative and positive values indicate that TMC slope decreased and increased with increasing frequency or L_0, respectively. For five participants (S026, S116, S142, S182, and S199), TMC slope

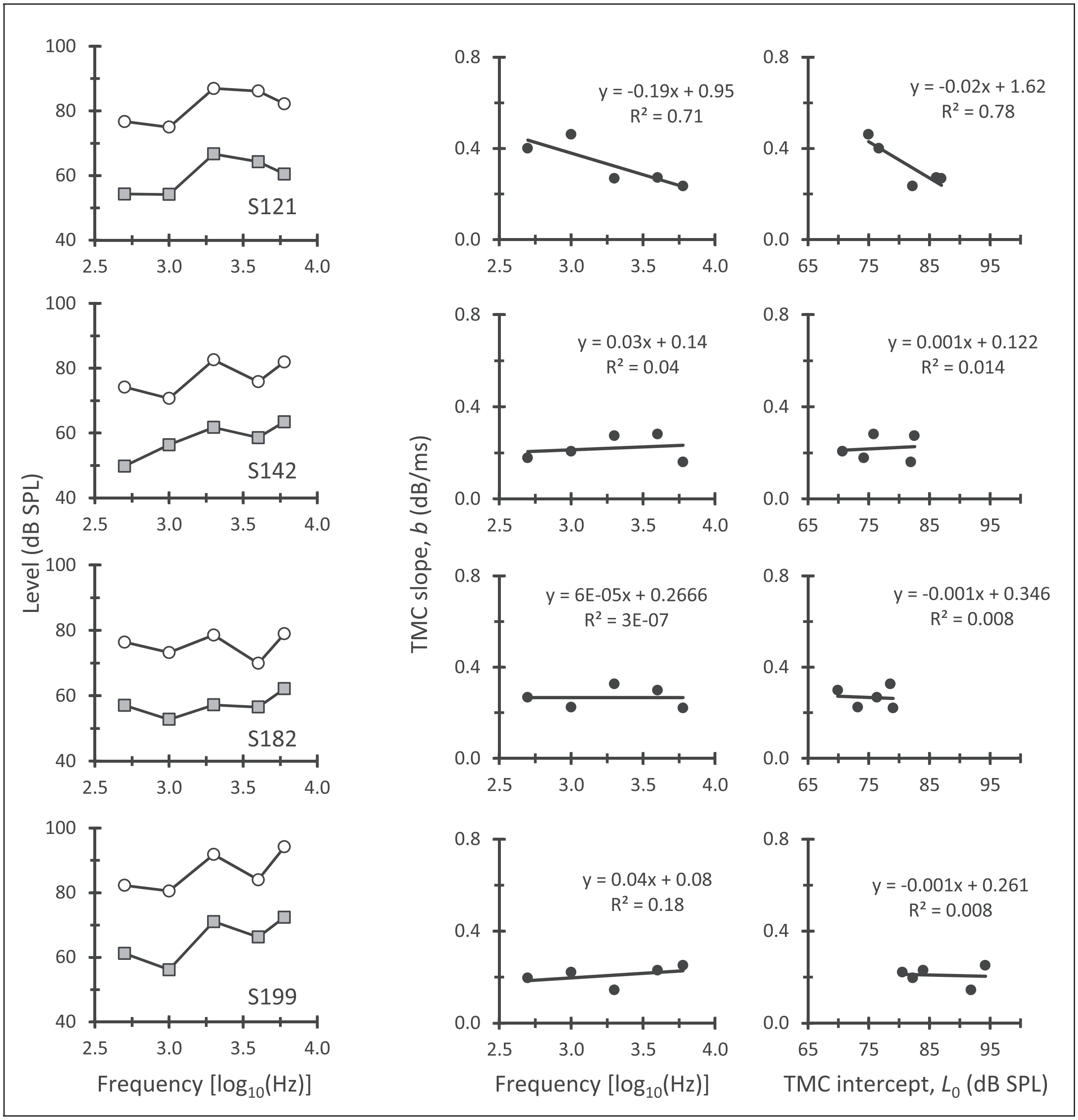

Figure 3. As Figure 2 but for four other participants (S121, S142, S182 and S199).

barely changed across frequencies or TMC intercept levels. For the remaining three participants (S012, S054, and S121), however, TMC slope decreased with increasing frequency and with increasing TMC intercept level. Interestingly, for the latter participants, TMC slope covaried with L_0 and with frequency, an aspect that will be further investigated later.

The filled symbols in Figure 4 show mean slopes of the linear regression trends across participants. On average, TMC slope decreased slightly with increasing frequency (mean $= -0.08$, $SD = 0.16$ dB/ms/log$_{10}$(Hz)) indicating that on-frequency TMCs were on average about 10% shallower at 6000 than at 500 Hz. Mean TMC slope also decreased slightly with increasing L_0 (mean- $= -0.005$, $SD = 0.0097$ dB/ms/dB) indicating that TMCs with $L_0 = 95$ dB SPL were on average about 10% shallower than those with $L_0 = 75$ dB SPL. Given the rather large variability across participants, however, the mean linear regression slopes were not statistically different from zero. In other words, mean TMC slope decreased slightly with increasing frequency and intercept level, across the probe frequency range

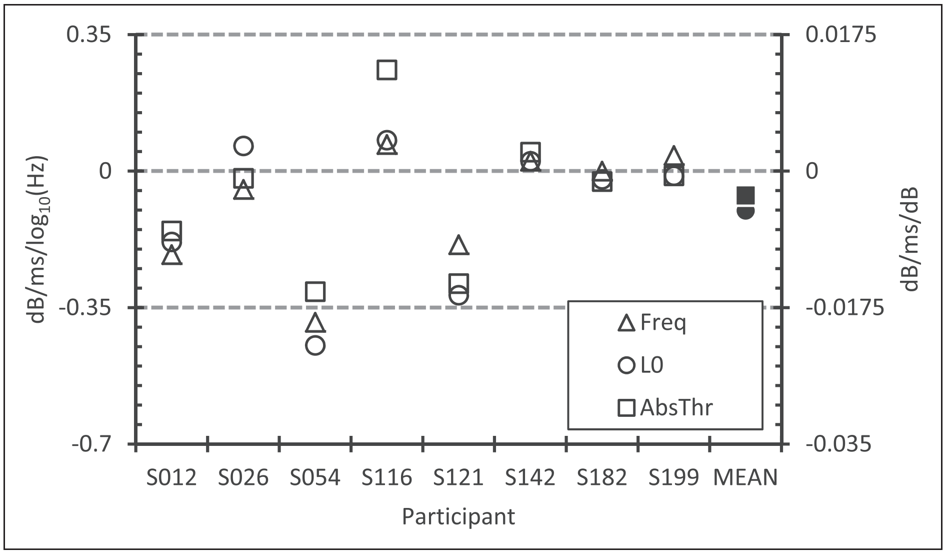

Figure 4. Trends of on-frequency TMC slope as a function of frequency or level for each participant or the mean. Each point depicts the slope of the linear regression lines in Figures 2 and 3. In other words, positive/negative values indicate that TMC slope increases/decreases with increasing frequency (left ordinate) or level (right ordinate).

(500–6000 Hz) and intercept level range (69–96 dB SPL) tested, but the trends were not significant.

Possible Frequency-Level Interactions on Forward-Masking Recovery

As shown in Figure 4, TMC slope covaried with frequency and with TMC intercept level. Probe frequency and intercept level were closely related with each other: Intercept level was higher at higher frequencies, particularly for those participants with sloping audiograms (left panels in Figures 2 and 3). An attempt was made to disentangle which of these two factors (probe frequency or intercept level) had a stronger influence on TMC slope. Our approach was based on the idea that if the main factor were level, then for a given probe frequency TMC slope should be negatively correlated with intercept level; however, if the main factor were frequency, then for TMCs with comparable intercept levels TMC slope should be negatively correlated with probe frequency. A third possibility could be that TMC slope concomitantly decreased with increasing probe frequency *and* intercept level.

Figure 5 illustrates the results of this analysis. The left panels show TMC slope against intercept level separately for each of the five frequencies tested. Note that the different points in a given panel correspond to different participants. TMC slope tended to be negatively correlated with level at frequencies of 1, 2, and 4 kHz. Despite the trends, however, the correlation was statistically significant only at 2 kHz (two-tailed t test, $N = 8$; $r = -.794$, $p = .0327$). The right panels in Figure 5 show TMC slope against probe frequency (expressed as $\log_{10}$(Hz)) for TMCs with intercept levels around approximately 76 (Figure 5(f)), 80 (Figure 5(g)), 85 (Figure 5(h)), and 89 dB SPL (Figure 5i), respectively. Slope also tended to be negatively correlated with frequency for intercept levels of 80, 85, and 89 dB SPL, but not for 76 dB SPL. Despite the trends, the correlations were not statistically significant at any of the four intercept levels.

In summary, the present data suggest that the rate of forward-masking recovery decreased with increasing level at frequencies of 1 to 4 kHz. They also suggest that the rate of forward-masking recovery decreased with increasing frequency, at least for TMCs that involved masker levels $\geqslant$80 dB SPL. Overall, however, the trends were not statistically significant possibly due to the small sample size.

To further assess the effect of level on forward-masking recovery while minimizing the potentially concomitant effect of frequency, we compared the slope of reference and on-frequency TMCs measured at the same probe frequency. If forward-masking recovery were independent of level, on-frequency and reference TMCs should have comparable slopes. The relevant data are shown in Figure 6. Note that only seven of the eight possible pairs of reference and on-frequency TMCs (Table 1) were available because the on-frequency TMC was missing for S116 at 6 kHz. In all cases, reference TMCs had higher intercept levels than corresponding on-frequency TMCs. For four of the seven participants (S026, S121, S142, and S182), reference TMCs had shallower slopes than their corresponding

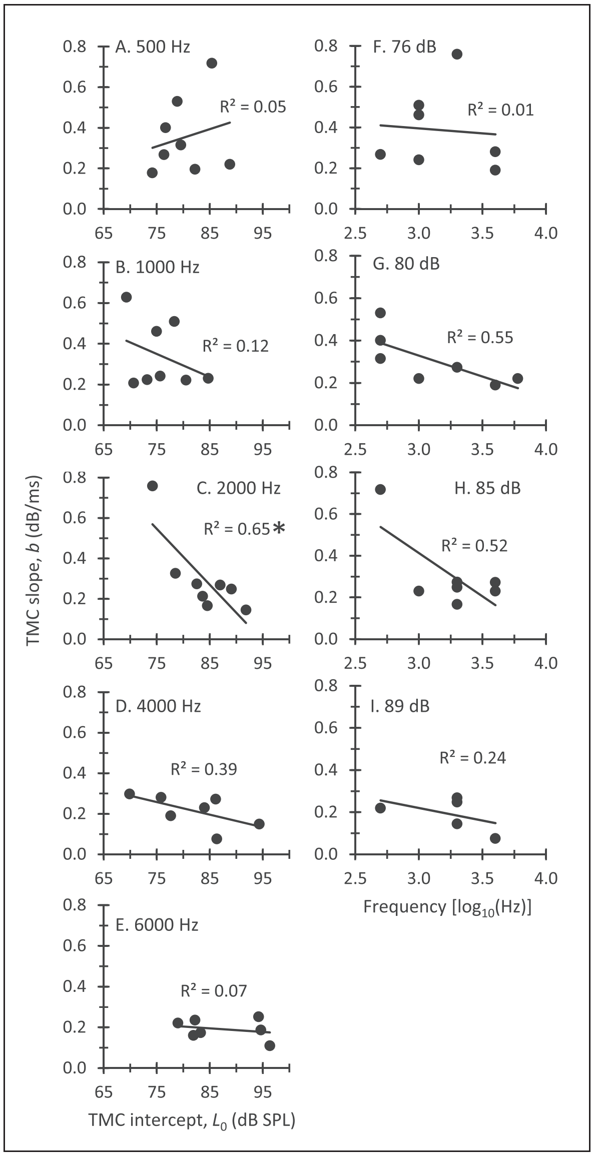

Figure 5. Left: On-frequency TMC slope as a function of intercept level. Each panel is for a different probe frequency, as indicated at the top of the panel. Right: On-frequency TMC slope as a function of frequency. Each panel is for a different intercept level, as indicated at the top of the panel. In each panel, different points are for different participants ($N = 8$). Missing points indicate that the corresponding TMC could not be measured. An asterisk (*) indicates a statistically significant correlation (two-tailed t test, $p < .05$).

on-frequency TMCs. For the remaining three participants (S012, S054, and S199), the slope of the reference TMC was comparable with or slightly greater than that of the corresponding on-frequency TMC. On average, reference TMCs had shallower slopes than on-frequency TMCs (0.15 vs. 0.22 dB/ms), but the difference was not statistically significant (two-tailed t test, $N = 7$, $p = .0851$).

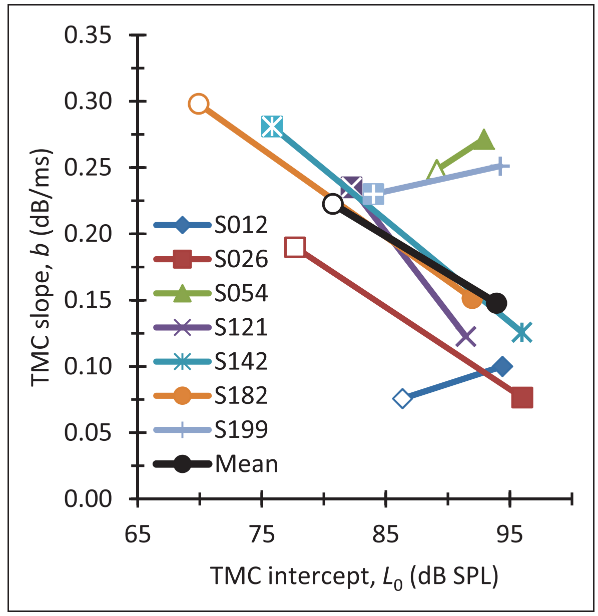

Figure 6. Slope of on-frequency (left symbols) and reference (right symbols) TMCs as a function of TMC intercept level. Each pair of data points is for a different participant.

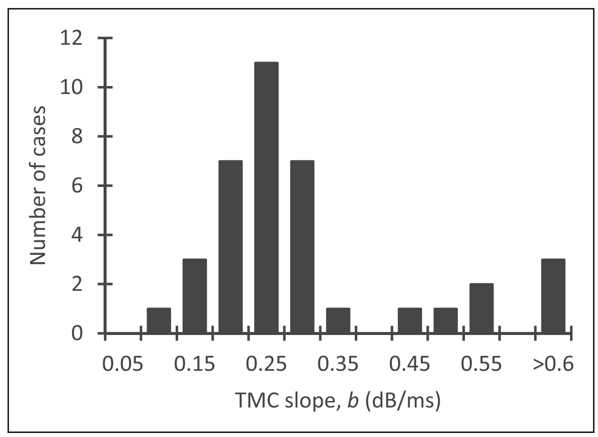

Figure 7. Histogram of on-frequency TMC slopes. Note that the total number of cases is 46 (8 participants × 6 frequencies minus two TMCs that could not be measured for S116, Table 3).

Discussion

We have investigated forward-masking recovery in a group of eight hearing-impaired listeners carefully selected to have absent or nearly absent DPOAEs over the range of primary L_2 levels from 35 to 70 dB SPL and over the range of primary f_2 frequencies from 500 to 4000 Hz (Table 2). We have shown that (a) for most cases, forward-masking recovery appeared constant across frequencies and levels; for some cases, however, forward-masking recovery decreased with increasing frequency and with increasing level (Figures 2 and 3);

(b) for those cases in which forward-masking recovery decreased with increasing frequency, forward-masking recovery also decreased with increasing level (Figure 4); (c) on average, however, forward-masking recovery did not change significantly across the range of probe frequencies (500–6000 Hz) or levels (70–100 dB SPL) tested; (d) for some individuals, forward-masking recovery measured using a fixed high-frequency probe was slower for low off-frequency maskers than for on-frequency maskers, while for others forward-masking recovery was comparable for on- and off-frequency maskers (Figure 6); and (e) on average, however, forward-masking recovery was not significantly different for on- and off-frequency maskers.

Limitations of the Present Data

Assuming that the absence of DPOAEs for levels below 70 dB SPL is indicative of linear cochlear responses, the present results would suggest that forward-masking recovery is frequency- and level-independent on average and for a majority of individuals but not for all individuals. One might argue, however, that the absence of DPOAEs below 70 dB SPL does not necessarily imply linear responses at the higher levels involved in the present TMCs (70–100 dB SPL, Figure 1). In other words, one might argue that the present TMCs could still be affected by compression. We could not rule this possibility out experimentally because the distortion generated by our DPOAE measurement system was too high at levels $L_2 > 70$ dB SPL to reliably assess the presence or absence of cochlear-generated DPOAEs at those levels. (We note that this limitation is common to most DPOAE measurements systems; see, e.g., Dorn et al., 2001.) In primates, however, cochlear gain, defined as the cochlear sensitivity at the CF premortem re postmortem, is about 40 dB at 6.5 to 8 kHz (see Table 1 of Robles & Ruggero, 2001). The present participants had hearing losses of at least 40 dB and typically greater at all test frequencies (Figures 2 and 3). Therefore, it is not unreasonable to assume that cochlear responses were linear for a majority of the present participants and conditions.

On the other hand, residual compression would lead to abnormally steep TMCs (Nelson et al., 2001). Figure 7 shows a histogram of the present on-frequency TMC slopes. The figure clearly suggests two groups of TMCs: a normally distributed group with slopes $\leqslant 0.35$ dB/ms and a group with higher slopes. It is tempting to speculate that the former group (with shallower slopes) possibly corresponds to TMCs unaffected by compression while the latter group (with steeper slopes) corresponds to TMCs that might be affected by residual compression. The latter group includes the steeper TMCs for participants S054, S116, and S121. These three participants had *sloping* audiograms; that is, greater losses at high than at low frequencies (left panels in Figures 2 and 3). Coincidentally, S054 and S121 are two of three cases for whom TMC slopes decreased with increasing frequency (S012, S054, and S121). If the present data were reanalyzed omitting slopes greater than 0.35 dB/ms, there would remain only one case (S012) for whom TMC slope would still change with frequency or level; for all other cases, TMC slope would be approximately constant across frequencies and levels. Therefore, it is tempting to conclude that TMC slope decreased with increasing frequency or level for some of the present participants because they had residual compression and that in the absence of compression forward-masking recovery would be constant across frequencies and levels.

In designing the present study, we have carried over the assumption from seminal reports that the TMC slope depends simultaneously on the amount of BM compression affecting the masker *and* on the rate of recovery from the internal (post-BM) masker effect (Nelson et al., 2001). Recent physiological, psychophysical, and modeling studies have shown or suggested sources of post-BM nonlinearity in the cochlea on responses that provide the input to the auditory nerve. For example, a recent study has shown that the motion of the reticular lamina shows more compression than the corresponding BM motion (Chen et al., 2011), indicating that the motion of the inner hair cell (IHC) stereocilia is not directly coupled to BM motion as is commonly thought (Guinan, 2012). In addition, Lopez-Poveda et al. (2005) noted that reference TMCs are shallower for some hearing-impaired than for normal-hearing listeners and argued that this could be due to frequency-unspecific compression in the IHCs that is present in normal-hearing listeners but reduced or absent in hearing-impaired listeners. This idea that IHC nonlinearities could be steepening the TMC slope has been later supported by model simulations of IHC potentials (Lopez-Poveda & Eustaquio-Martin, 2006) and by other psychoacoustical studies (Plack & Arifianto, 2010). This recent evidence suggests that in addition to BM compression, the slope of a TMC may also be affected (steepened) by compression added by the reticular lamina or the IHC. In the present context, this implies that even if the present TMCs were unaffected by BM compression (see the preceding paragraphs), they might still be affected by post-BM compression. In that case, the present analysis would still be correct if the post-BM compression were comparable across all the conditions tested here, something that is admittedly uncertain.

Of course, the TMC method was designed to infer BM compression specifically. Because it consists of comparing the slopes of two TMCs measured with different frequencies and because there is no evidence (to our knowledge) that post-BM compression is frequency selective, post-BM compression effects on individual TMCs would be cancelled in the comparison and hence the TMC method may still be useful for its purpose.

Relationship With Earlier Studies

Using an approach similar to the present one, Stainsby and Moore (2006) concluded that forward-masking recovery was negatively correlated with frequency. The present study uses a larger sample selected with more rigorous DPOAE criteria and a different analysis. The present results suggest that the trends reported by Stainsby and Moore could be due to their participants having residual compression at low frequencies.

Implications for Estimating Compression From TMCs

In inferring peripheral cochlear compression from TMCs, it is assumed that the post-mechanical (or compression free) rate of recovery from the masker effect is independent of probe frequency and of masker level (Lopez-Poveda et al., 2003; Nelson et al., 2001). The present mean results support the assumptions of the TMC method. This is not to say, however, that it would be accurate to infer cochlear compression from comparisons of on-frequency and reference TMC slopes in all cases. Efferent effects might affect forward-masking recovery in normal-hearing listeners or in hearing-impaired listeners with residual OHC function (Jennings, Strickland, & Heinz, 2009; Wojtczak & Oxenham, 2009; Wojtczak & Oxenham, 2010; Yasin et al., 2013). Therefore, that post-mechanical forward-masking recovery is generally frequency- and level-independent for hearing-impaired listeners with absent compression does not imply that compression estimates inferred with the standard TMC method are accurate. Wojtczak and Oxenham (2009) showed that for normal-hearing listeners, forward-masking recovery is slower for levels above than below 83 dB SPL. They reasoned that the TMC method can overestimate compression by approximately a factor of 2 when reference TMCs involve levels above 83 dB SPL (i.e., in those cases, the actual compression exponent could be half of the inferred value). Wojtczak and Oxenham argued that this was possibly due to high-level off-frequency masker activating the MEMR.

The present results suggest that something similar may also happen for hearing-impaired listeners. Reference TMCs had shallower slopes than on-frequency TMCs measured with the same probe frequency (Figure 6). One explanation for this result might be that despite our precautions, the participants in question still had residual compression at high frequencies. On the other hand, the present reference TMCs always involved values higher than 90 to 95 dB SPL (their L_0 is illustrated by the rightmost points in the Figure 6), hence comparable with the threshold levels of activation of the MEMR for the present participants (shown in Table 1). Indeed, the actual activation threshold of the MEMR can be 8 to 14 dB lower than estimated with clinical methods similar to the one employed here (Feeney, Keefe, & Marryott, 2003; Neumann, Uppenkamp, & Kollmeier, 1996). The MEMR can be elicited by sounds with a duration of 116 ms (Keefe, Fitzpatrick, Liu, Sanford, & Gorga, 2010), which is approximately half the duration of the present maskers. The MEMR hinders the transmission of frequencies between 300 and 1000 Hz and has no significant effect on the transmission of frequencies higher than 2000 Hz but *facilitates* the transmission of frequencies between 1000 and 2000 Hz (see the top panels in Figure 1 of Feeney et al., 2003 and in Figure 2 of Feeney, Keefe, & Sanford, 2004). The maskers used to measure the reference TMCs were long enough that the MEMR could be active during the course of the masker and had frequencies (800–2000 Hz) within the range of the facilitating effect of MEMR. Therefore, it is conceivable that MEMR facilitated the transmission of the reference maskers, thereby reducing the masker level at the probe masked threshold. The MEMR would have a much lesser effect for corresponding on-frequency TMCs because the involved masker frequencies were higher than 2000 kHz, where the MEMR effect is negligible. Therefore, an alternative explanation for the shallower slopes of reference TMCs could be that forward-masking recovery did depend on masker level possibly due to the activation of the MEMR. If the latter explanation were correct, the present data (Figure 6) would indicate that compression inferred from comparisons of on-frequency and reference TMCs can be twice as much as the actual compression for hearing-impaired listeners whose reference TMCs involve masker levels above the individual threshold of activation of the MEMR.

Conclusions

On the basis of the analysis of TMCs for hearing-impaired listeners with presumably linear BM responses, we conclude that forward-masking recovery is independent of probe frequency and of masker level, hence that it is reasonable to use a reference TMC for a high-frequency probe to infer cochlear compression at lower frequencies.

Reference TMCs can be sometimes shallower than corresponding on-frequency TMCs for identical probe frequencies. The reason is uncertain. It might occur when the masker used to measure the reference TMC is of sufficient duration and intensity to activate the MEMR. Whatever the reason, BM compression could be overestimated in these cases by as much as a factor of two.

Acknowledgements

The authors thank Sridhar Kalluri and Bill Woods for their comments. The authors also thank two anonymous reviewers and the editor, Andrew J. Oxenham, for many useful suggestions during the review process.

Declaration of Conflict of Interests

The authors declared no potential conflicts of interest with respect to the research, authorship, and/or publication of this article.

Funding

The authors disclosed receipt of the following financial support for the research, authorship, and/or publication of this article: This study was financially supported by Starkey Ltd. (USA), Junta de Castilla y León, and the Spanish Ministry of Economy and Competitiveness (ref. BFU2012-39544-C02).

References

American National Standards Institute. (1996). *S3.6 Specification for audiometers*. New York, NY: American National Standards Institute.

Bacon, S. P., & Oxenham, A. J. (2004). Psychophysical manifestations of compression: Hearing impaired listeners. In S. P. Bacon, R. R. Fay, & A. N. Popper (Eds.), *Compression: From cochlear to cochlear implants* (pp. 107–152). New York, NY: Springer-Verlag.

Chen, F., Zha, D., Fridberger, A., Zheng, J., Choudhury, N., Jacques, S. L., Wang, R. K., Shi, X., & Nuttall, A. L. (2011). A differentially amplified motion in the ear for near-threshold sound detection. *Nature Neuroscience*, *14*, 770–774.

Dorn, P. A., Konrad-Martin, D., Neely, S. T., Keefe, D. H., Cyr, E., & Gorga, M. P. (2001). Distortion product otoacoustic emission input/output functions in normal-hearing and hearing-impaired human ears. *The Journal of the Acoustical Society of America*, *110*, 3119–3131.

Feeney, M. P., Keefe, D. H., & Marryott, L. P. (2003). Contralateral acoustic reflex thresholds for tonal activators using wideband energy reflectance and admittance. *Journal of Speech, Language, and Hearing Research*, *46*, 128–136.

Feeney, M. P., Keefe, D. H., & Sanford, C. A. (2004). Wideband reflectance measures of the ipsilateral acoustic stapedius reflex threshold. *Ear and Hearing*, *25*, 421–430.

Guinan, J. J. (2012). How are inner hair cells stimulated? Evidence for multiple mechanical drives. *Hearing Research*, *292*, 35–50.

Jennings, S. G., Strickland, E. A., & Heinz, M. G. (2009). Precursor effects on behavioral estimates of frequency selectivity and gain in forward masking. *Journal of the Acoustical Society of America*, *125*, 2172–2181.

Johannesen, P. T., & Lopez-Poveda, E. A. (2008). Cochlear nonlinearity in normal-hearing subjects as inferred psychophysically and from distortion-product otoacoustic emissions. *Journal of the Acoustical Society of America*, *124*, 2149–2163.

Johannesen, P. T., & Lopez-Poveda, E. A. (2010). Correspondence between behavioral and individually "optimized" otoacoustic emission estimates of human cochlear input/output curves. *Journal of the Acoustical Society of America*, *127*, 3602–3613.

Johannesen, P. T., Pérez-González, P., & Lopez-Poveda, E. A. (2014). Across-frequency behavioral estimates of the contribution of inner and outer hair cell dysfunction to individualized audiometric loss. *Frontiers in Nuroscience*, *8*, 214.

Jurgens, T., Kollmeier, B., Brand, T., & Ewert, S. D. (2011). Assessment of auditory nonlinearity for listeners with different hearing losses using temporal masking and categorical loudness scaling. *Hearing Research*, *280*, 177–191.

Keefe, D. H., Fitzpatrick, D., Liu, Y. W., Sanford, C. A., & Gorga, M. P. (2010). Wideband acoustic-reflex test in a test battery to predict middle-ear dysfunction. *Hearing Research*, *263*, 52–65.

Lopez-Poveda, E. A., & Alves-Pinto, A. (2008). A variant temporal-masking-curve method for inferring peripheral auditory compression. *Journal of the Acoustical Society of America*, *123*, 1544–1554.

Lopez-Poveda, E. A., & Eustaquio-Martin, A. (2006). A biophysical model of the inner hair cell: The contribution of potassium currents to peripheral auditory compression. *Journal of the Association for Research in Otolaryngology*, *7*, 218–235.

Lopez-Poveda, E. A., & Johannesen, P. T. (2009). Otoacoustic emission theories and behavioral estimates of human basilar membrane motion are mutually consistent. *Journal of the Association for Research in Otolaryngology*, *10*, 511–523.

Lopez-Poveda, E. A., & Johannesen, P. T. (2012). Behavioral estimates of the contribution of inner and outer hair cell dysfunction to individualized audiometric loss. *Journal of the Association for Research in Otolaryngology*, *13*, 485–504.

Lopez-Poveda, E. A., Plack, C. J., & Meddis, R. (2003). Cochlear nonlinearity between 500 and 8000 Hz in listeners with normal hearing. *Journal of the Acoustical Society of America*, *113*, 951–960.

Lopez-Poveda, E. A., Plack, C. J., Meddis, R., & Blanco, J. L. (2005). Cochlear compression in listeners with moderate sensorineural hearing loss. *Hearing Research*, *205*, 172–183.

Meddis, R., Lecluyse, W., Tan, C. M., Panda, M., & Ferry, R. (2010). Beyond the audiogram: Identifying and modeling patterns of hearing loss. In E. A. Lopez-Poveda, A. R. Palmer, & R. Meddis (Eds.), *The neurophysiological bases of auditory perception* (pp. 631–640). New York, NY: Springer.

Neely, S. T., Johnson, T. A., & Gorga, M. P. (2005). Distortion-product otoacoustic emission measured with continuously varying stimulus level. *Journal of the Acoustical Society of America*, *117*, 1248–1259.

Nelson, D. A., & Schroder, A. C. (2004). Peripheral compression as a function of stimulus level and frequency region in normal-hearing listeners. *Journal of the Acoustical Society of America*, *115*, 2221–2233.

Nelson, D. A., Schroder, A. C., & Wojtczak, M. (2001). A new procedure for measuring peripheral compression in normal-hearing and hearing-impaired listeners. *Journal of the Acoustical Society of America*, *110*, 2045–2064.

Neumann, J., Uppenkamp, S., & Kollmeier, B. (1996). Detection of the acoustic reflex below 80 dB HL. *Audiol Neurootol*, *1*, 359–369.

Oxenham, A. J., & Bacon, S. P. (2003). Cochlear compression: Perceptual measures and implications for normal and impaired hearing. *Ear & Hearing*, *24*, 352–366.

Oxenham, A. J., & Bacon, S. P. (2004). Psychophysical manifestations of compression: Normal-hearing listeners. In S. P. Bacon, R. R. Fay, & A. N. Popper (Eds.), *Compression: From cochlea to cochlear implants* (pp. 62–106). New York, NY: Springer.

Oxenham, A. J., & Plack, C. J. (1997). A behavioral measure of basilar-membrane nonlinearity in listeners with normal and impaired hearing. *Journal of the Acoustical Society of America*, *101*, 3666–3675.

Panda, M. R., Lecluyse, W., Tan, C. M., Jürgens, T., & Meddis, R. (2014). Hearing dummies: Individualized computer models of hearing impairment. *International Journal of Audiology*, *53*, 699–709.

Plack, C. J., & Arifianto, D. (2010). On- and off-frequency compression estimated using a new version of the additivity of forward masking technique. *Journal of the Acoustical Society of America*, *128*, 771–786.

Plack, C. J., & Drga, V. (2003). Psychophysical evidence for auditory compression at low characteristic frequencies. *Journal of the Acoustical Society of America*, *113*, 1574–1586.

Plack, C. J., Drga, V., & Lopez-Poveda, E. A. (2004). Inferred basilar-membrane response functions for listeners with mild to moderate sensorineural hearing loss. *Journal of the Acoustical Society of America*, *115*, 1684–1695.

Plack, C. J., & O'Hanlon, C. G. (2003). Forward masking additivity and auditory compression at low and high frequencies. *Journal of the Association for Research in Otolaryngology*, *4*, 405–415.

Plack, C. J., & Oxenham, A. J. (2000). Basilar-membrane nonlinearity estimated by pulsation threshold. *Journal of the Acoustical Society of America*, *107*, 501–507.

Plack, C. J., Oxenham, A. J., Simonson, A. M., O'Hanlon, C. G., Drga, V., & Arifianto, D. (2008). Estimates of compression at low and high frequencies using masking additivity in normal and impaired ears. *Journal of the Acoustical Society of America*, *123*, 4321–4330.

Popelka, G. R., Osterhammel, P. A., Nielsen, L. H., & Rasmussen, A. N. (1993). Growth of distortion product otoacoustic emissions with primary-tone level in humans. *Hearing Research*, *71*, 12–22.

Rhode, W. S., & Cooper, N. P. (1996). Nonlinear mechanics in the apical turn of the chinchilla cochlea in vivo. *Auditory Cognitive Neuroscience*, *3*, 101–121.

Robles, L., & Ruggero, M. A. (2001). Mechanics of the mammalian cochlea. *Physiological Reviews*, *81*, 1305–1352.

Stainsby, T. H., & Moore, B. C. (2006). Temporal masking curves for hearing-impaired listeners. *Hearing Research*, *218*, 98–111.

Wojtczak, M., & Oxenham, A. J. (2009). Pitfalls in behavioral estimates of basilar-membrane compression in humans. *Journal of the Acoustical Society of America*, *125*, 270–281.

Wojtczak, M., & Oxenham, A. J. (2010). Recovery from on- and off-frequency forward masking in listeners with normal and impaired hearing. *Journal of the Acoustical Society of America*, *128*, 247–256.

Yasin, I., Drga, V., & Plack, C. J. (2013). Estimating peripheral gain and compression using fixed-duration masking curves. *Journal of the Acoustical Society of America*, *133*, 4145–4155.

Zurek, P. M. (1992). Detectability of transient and sinusoidal otoacoustic emissions. *Ear and Hearing*, *13*, 307–310.