Beyond the audiogram: identifying and modelling patterns of hearing deficits.

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Abstract

The choice of a hearing aid and its tuning parameters should benefit from a more detailed assessment of a patient’s hearing than occurs at present in clinical practice. The process of fitting and tuning might be facilitated even more if these data were used to generate a computer model of the patient’s hearing. A number of hearing impaired patients have been tested using a range of tests that were devised to be easy to administer under automatic computer control and easy for the, often elderly, listeners to use. The principal finding is that there is an unexpected variation among patients in the patterns of impairment that are revealed by the tests. This is true even when the patients have similar audiograms. The data have been used to develop individual computer models of the patients’ hearing which we call ‘hearing dummies’. These dummies have been successfully evaluated using the same automated tests that were used to collect the original data. The creation of these dummies raises interesting questions about the basis of different forms of hearing pathology and point to a future where all forms of hearing impairment and their treatment might be diagnosed in terms of the underlying pathology rather than the symptomatology as at present.

1.1 Introduction

The audiogram is the main basis for the fitting of hearing aids. However, experience shows that patients with the same audiogram may experience different degrees of benefit from the same hearing aid. It is likely that supplementary information such as frequency selectivity and measures of residual compression could improve the prescription process or drive the development of more appropriate hearing aid algorithms. Unfortunately, this additional information will not immediately help audiologists to prescribe existing aids because the manufactures fitting guidance does not take this information into account. Nor will it necessarily be of direct help to manufacturers unless some method is found for predicting the outcome for a particular patient of different aids, algorithms or settings.

To address the ‘data poverty’ problem, we are exploring methods for extending the range of measures that might usefully be deployed in a clinical context with an emphasis on speed of data collection and ease of use from the patient’s point of view. Using a single-interval, up-down automated procedure, we have been able to measure frequency-selectivity in impaired, often elderly, patients over time-periods that are clinically viable. The same approach has been used to measure residual compression. When this additional information is combined with other routine assessments, we obtain a rich audiological profile that reveals surprisingly diverse patterns across patients.

We have also begun to explore the problem of predicting hearing aid outcomes by developing computer models (‘hearing dummies’) of the hearing of individual patients. The models simulate the physiological processes underlying successive signal processing stages of the auditory periphery. Individual deficiencies are modelled by introducing pathological features such as reduced endocochlear potential, impaired OHC functioning, dead regions, etc. The model is, in effect, a hypothesis concerning the underlying pathology of the auditory periphery responsible for the deficit. The validity of the computer model is evaluated by testing it using the *same* software and procedures that are used with the patient. The ultimate aim of the project is to use the dummies to predict outcomes in an individual patient for a range of different hearing aids or different settings for a given aid.

The results reported below illustrate data collection and modelling results using data from one normal and three impaired listeners and our attempts to model their thresholds and frequency selectivity. The illustrations are restricted for reasons of space to absolute thresholds and frequency selectivity measured using iso-forward masking contours (IFMCs). The aim is to illustrate the general thrust of our on-going work and to stress that the driving motivation of the research is to characterise the hearing of an individual patient rather than make scientific statements about types of hearing impairment.

1.2 Methods

1.2.1 Psychoacoustic profiles of normal and impaired listeners

Psychoacoustic measures

Absolute thresholds: Detection thresholds were measured where possible for 500-ms pure tones (raised cosine onset and offset times of 4 ms) at frequencies 250, 500, 1000, 2000, 4000 and 8000 Hz.

Frequency selectivity was assessed using iso-forward masking contours (IFMC): In this measurement the patient's task is to report the presence or absence of a probe tone that follows a masking tone after a gap of 10 ms. The probe had a duration of 8 ms (including 4-ms ramps) and was always presented at 10 dB SL. 100-ms masking tones were varied to find the masker level that resulted in 50% detection of the probe. Masker frequencies were 0.5, 0.7, 0.9, 1, 1.1, 1.3, and 1.6 * probe frequency. Masker thresholds were measured for probe frequencies 250, 500, 1000, 2000, 4000 and 6000 Hz. No efforts were made to prevent off-frequency listening because this is a useful indicator of non-functioning regions of the cochlea. For the impaired listeners, not all frequencies could be tested because of too-high absolute thresholds. In some instances additional intermediate frequencies were tested. Each IFMC threshold is generally the mean of three measurements.

Threshold estimation procedure

The basic procedure for measuring absolute thresholds is based on an adaptive yes-no paradigm previously evaluated by Lecluyse and Meddis (2009). A stimulus is presented to the participant who simply responds with a 'yes' or 'no' button press according to whether or not he hears the target. The level is changed from trial to trial using a one-up, one-down adaptive procedure. After an initial series of stimuli using a large step size to find the threshold region, the step size is reduced to 2 dB and the run then continues for 10 more trials not including an additional 20% of catch trials presented at random among the regular trials. These are trials where no stimulus is presented and the participant is expected to say ‘no’. On the few occasions when the participant is ‘caught out’, the run is stopped and restarted; possibly after resting the patient and giving further instructions. Patients are encouraged to be conservative in their judgments and false-alarms are, in practice, very rare. Our experience is that this method is fast, reliable and acceptable to often elderly patients who need little (or sometimes no) training before producing useful results. Similar methods have recently been explored by Leek et al (2000) in a variety of audiological research contexts.

The threshold is estimated at the end of the run by fitting a psychometric function of the form $p = (1 + \exp(-k(L\text{-}Th)))^{-1}$ to the responses, where, p is the proportion of ‘yes’ responses, L is the level of the stimulus (dB SPL), k is a slope parameter and Th is the threshold, the level of the stimulus at which the proportion of yes-responses is 0.5. The function is fitted using a least-squares, best-fit procedure, with Th and k as free parameters.

Stimuli were generated using a sampling rate of 96000 Hz, with 24-bit resolution, and presented monaurally via circumaural Sennheiser HD600 headphones

Listeners

The behavioural data presented in this report was obtained from one normal listener and three impaired listeners. The normal listener was 21 years old with no history of hearing problems. The three impaired listeners were aged between 68 and 76.

1.2.2 Computer modelling

The models were developed using an existing published computer model of the auditory periphery (Meddis 2006). Its parameters were adjusted iteratively to produce an approximate match to the data for an individual patient. The model consists of a cascade of signal processing operations representing successive processing stages in the auditory periphery. The input is an acoustic signal and the output is a multi-channel, multi-fiber representation of high spontaneous rate (HSR) spiking activity in the auditory nerve (see Figure 1). This activity forms the input (on a tonotopic basis) to more models of sustained-chopping cells in the cochlear nucleus with low (~10 spikes/s) spontaneous firing rates. The output from these units is used as the input to a further layer of modelled units representing inferior colliculus (IC) cell coincidence-detectors (bottom panel, Fig. 1B). These were parameterised to have no spontaneous activity. Any activity in one or more of these cells was used to indicate that an acoustic stimulus had been detected.

The parameters of the model were fixed at values consistent with the physiological literature and each stage of the model has been separately evaluated to be consistent with published data obtained using small mammals. The parameters were then adjusted as little as possible to give a good representation of a single listener with good hearing (see below). The parameters for this ‘normal’ model were used as the starting point of the explorations used to model individual patterns of abnormal hearing; again by making as few changes as possible. The illustrations below are restricted to a single change with no further adjustments to produce more flattering fits to the patient data.

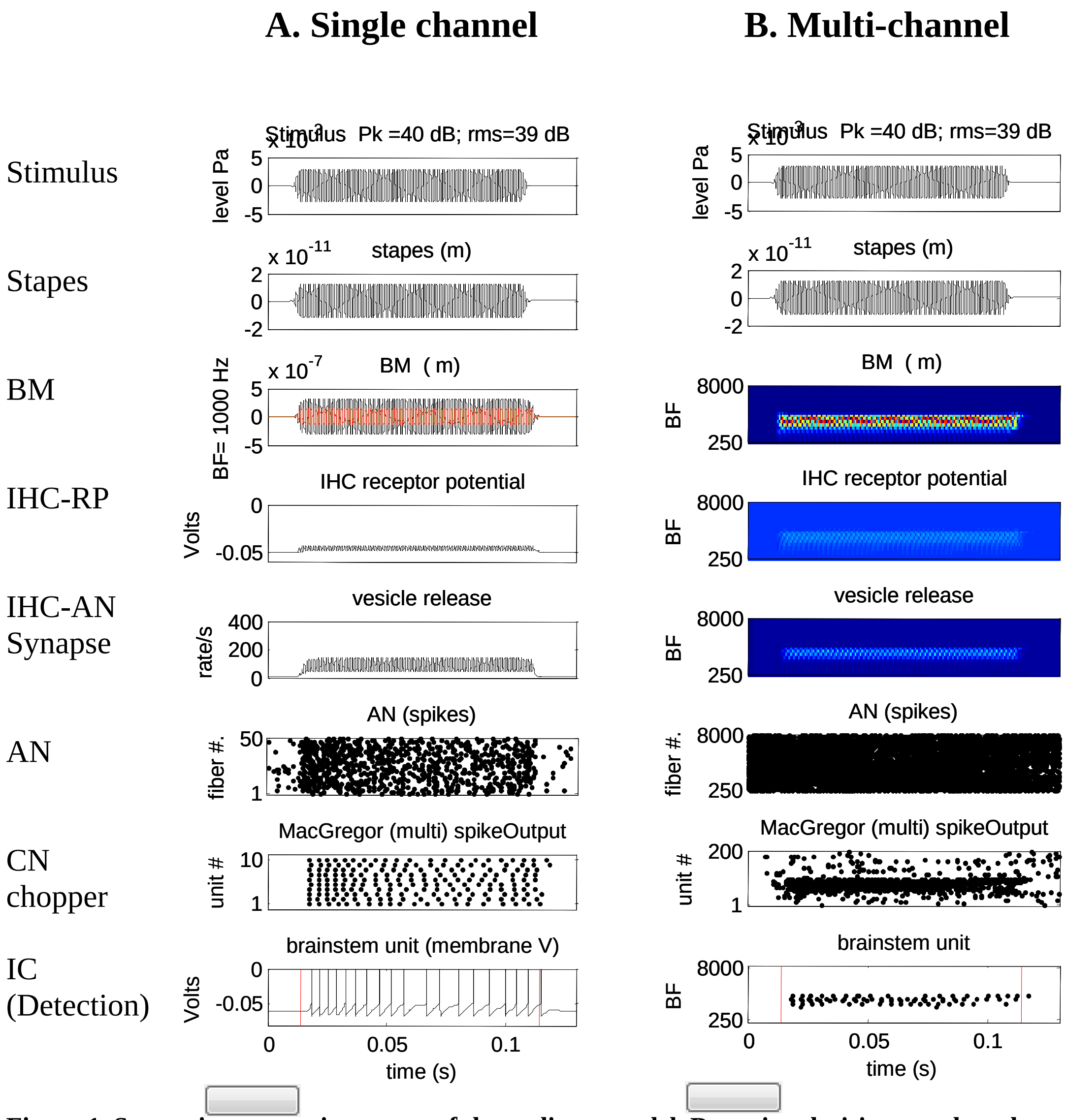

Figure 1. Successive processing stages of the auditory model. Detection decisions are based on the presence or absence of at least one action potential in the final (IC) stage during the presentation of the stimulus. The stimulus is a 100-ms, 1000-Hz pure tone presented at a level of 40 dB SPL (29 dB SL). A: single channel model (BF=1000 Hz). B: 40-channel model whose BFs range from 250-8000 Hz. In this illustration, the auditory nerve stage of the model has 50 AN fibers per channel. The CN stage of the model has 10 units per channel. The IC stage has one unit per channel. The IC units have no spontaneous activity and any activity in any of these units is taken to be indicative of the detection of an acoustic stimulus.

The model was evaluated using *exactly* the same procedures and software used to collect the participant's data. The model was harnessed to the psychometric software so that it 'pressed the yes-button' if one or more spikes were registered in the IC unit during the presentation of the stimulus probe. If no spikes occurred, it 'pressed the no-button'.

1.3 Results

1.3.1 Normal data and model

Figure 2A shows a hearing profile of a 21-year-old male listener (CMa) with excellent hearing. His hearing is the best of all normal subjects tested in the laboratory.

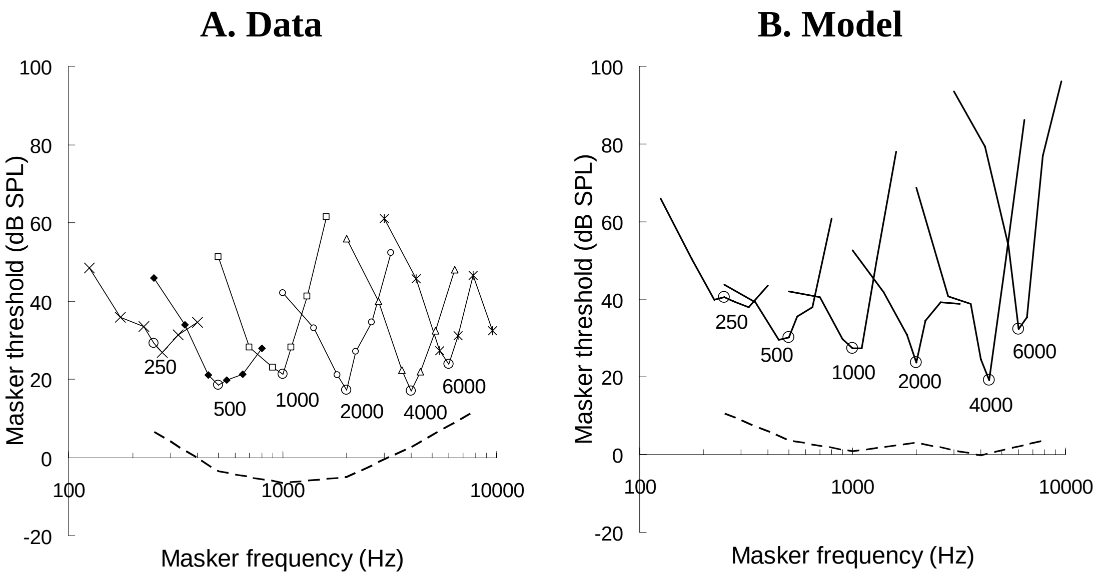

Figure 2. Absolute thresholds (dashed line) and IFMCs (full lines). For the IFMCs, the masker threshold when the masker and probe frequency are the same is represented by the large open circle. The data labels represent the probe frequency. A: Data for a normal listener (CMa, left ear). B: model results.

All thresholds are very low but within normal limits. Normal IFMCs were found at all frequencies. The computer model was carefully tuned to fit these data. Figure 2B shows the results of submitting the model to the same test procedures as the human listener. This model will be used as the baseline model for simulating impaired hearing. The impaired models will be described in terms of how they differ from the normal model.

1.3.2 Impaired data and models

Profile 1 (Participant ECr)

ECr is a 76-year-old male with bilateral moderate-severe sensory-neural hearing loss with normal middle ear function. His absolute thresholds are raised at low frequencies and are difficult to measure at frequencies above 1 kHz. The IFMCs show greatly reduced frequency selectivity and follow the contour of the absolute thresholds (Fig. 3A).

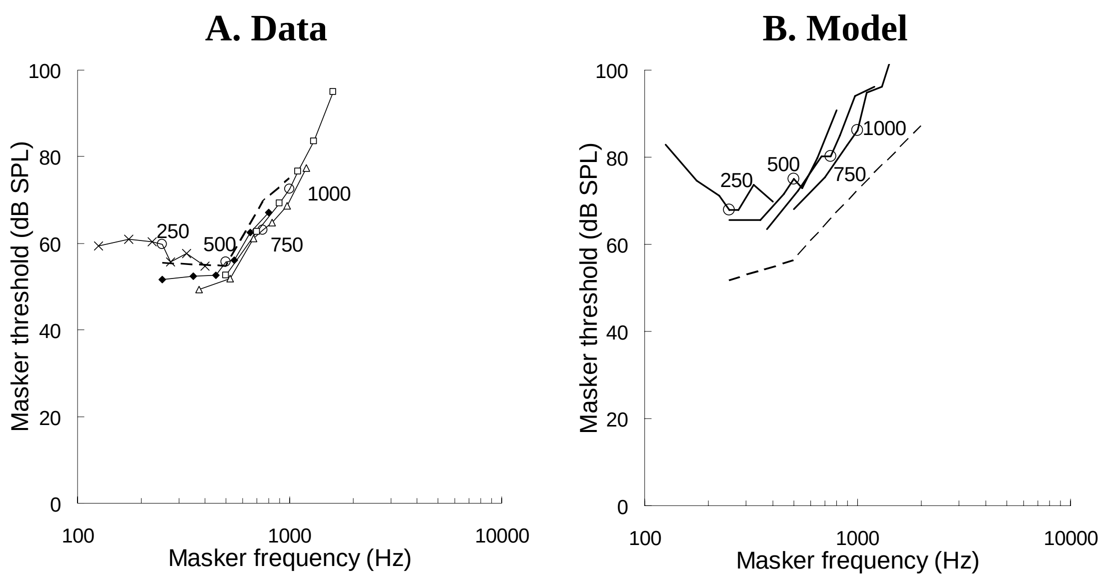

Figure 3. A: Data for an impaired listener (ECr, left ear). See Fig. 2 for more details. B: the 'impaired' model based on the normal model (Fig. 2B) with all channels disabled except BF=250 Hz (see text).

To develop a suitable model, we hypothesized that ECr retains only one functioning location along the BM that has a best frequency (BF) of 250 Hz and that the response is linear. The results in Fig. 3B were produced using a reduced version of the normal model illustrated earlier in Fig. 2B. All BM locations in the normal model other than the 250 Hz location were disabled and the nonlinear path of the dual resonance nonlinear (DRNL) simulation of BM activity in the remaining 250 Hz channel was also disabled leaving *only a single linear filter*. While the only remaining location has a BF of 250 Hz, the lowest threshold occurs at 500 Hz because of the high-pass contribution of the outer-middle ear which attenuates all frequencies below 1000 Hz. The IFMCs for probes greater than 500 Hz form simple diagonals along the contour of absolute thresholds as expected for probes in a non-functioning region.

Profile 2 (Participant JEV)

JEv is a 69 year-old male with a bilateral, moderate, sensory-neural, sloping hearing loss with normal middle ear function. The absolute thresholds show a moderate loss up to 1000 Hz and an increasing loss at higher frequencies. The loss at 8000 Hz is approximately 80 dB. IFMCs are shallower than normal at all frequencies especially from 2000 Hz onwards but still clearly V-shaped (Fig. 4A). Raised thresholds particularly at high frequencies combined with some preservation of tuning is consistent with a hypothesis that the deficit results from a low endocochlear potential (Ep). Schmiedt et al. (2002) has shown that thresholds for high frequency tones are more affected than low tones by loss of Ep.

Figure 4B shows the assessment of a model that is the same as the normal model in all respects except that Ep has been reduced from -0.1 V to -0.09 V. The raised thresholds are caused by a reduced response in the inner hair cells (IHC) resulting from the reduction in Ep. While the tuning curves are broader than for the normal subject, the IFMCs are what we would expect from a normal listener if tested using a more intense probe tone (i.e. at the same masker levels as used for JEv) (Nelson et al. 1990). The model reproduces Schmiedt's observations of greater threshold increases at 'higher frequencies' as an 'emergent property'.

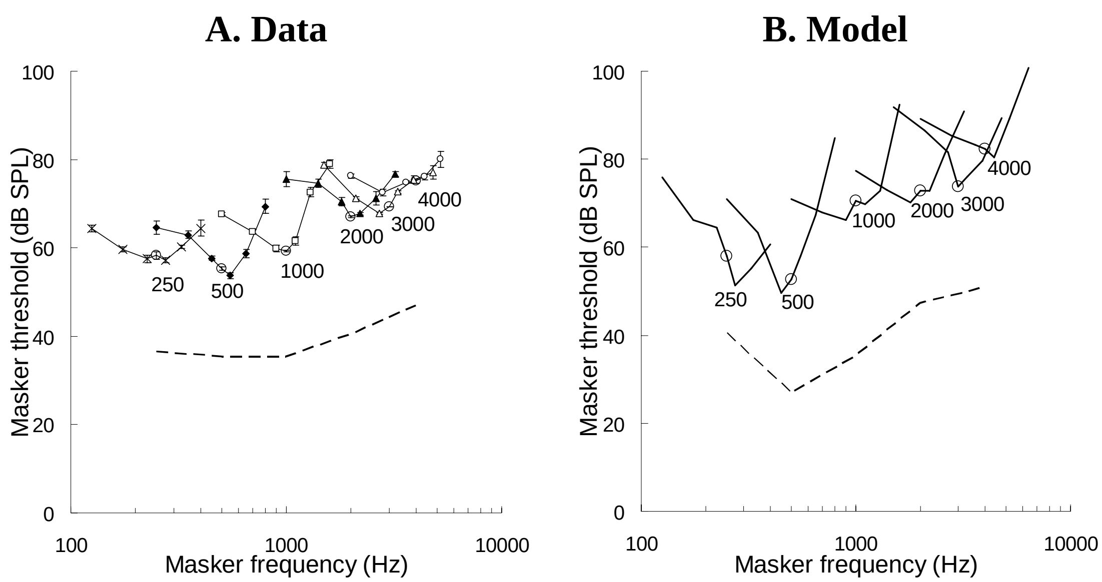

Figure 4. A: Data for an impaired listener (JEv, left ear). See Fig. 2 for more details B: the 'impaired' model based on the normal model (Fig. 2B) with Ep reduced from 0.1 to 0.09 V (see text).

Note that the model does not reproduce the flatter, asymmetric IFMCs at 3000 and 4000 Hz. This suggests that an improved model might be realised by disabling all BM locations above 2700 Hz. The simpler model has been retained here because it illustrates the effect of a single parameter change to the normal model. The emerging pattern of results is typical of what might be expected from an uncomplicated case of presbyacusis.

Profile 3 (Participant JJo)

JJo is a 68-year old man with a moderate bilateral, sensory-neural, ski-slope loss with normal middle ear function but no detectable acoustic reflex. Below 2000 Hz, thresholds and IFMCs are unexceptional (Fig. 5A). Above this, thresholds rise steeply and the IFMCs roughly follow the absolute threshold contour. This is consistent with the hypothesis that the BM is severely compromised above 1800 Hz.

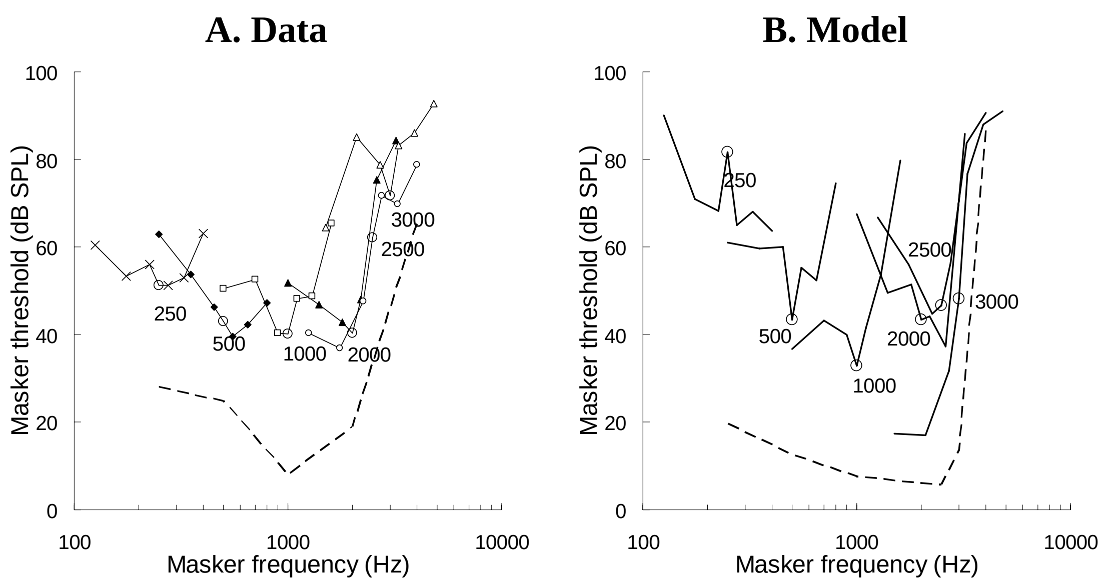

Figure 5. A: Data for an impaired listener (JJo, left ear). See Fig. 2 for more details. B: the 'impaired' model based on the normal model (Fig. 2B) with all channels with BFs> 1800 Hz disabled (see text).

This hypothesis is illustrated in Fig. 5B where the model is the same as the normal model except that all channels with BFs above 1800 Hz have been disabled.

Naturally, this results in raised thresholds for high frequency probes because they are heard only through lower-frequency channels. More interestingly, it gives rise to a pattern of IFMCs characteristic of non-functioning regions where the IFMC follows the contour of the absolute thresholds.

1.4 Discussion

While these results remain preliminary, they are encouraging in a number of respects. It is clear that the additional data obtained from the IFMCs allow us to identify different patterns of response in hearing impaired individuals. The addition of IFMC data helps to narrow the range of hypotheses that might account for the impairment. We also collect temporal masking curves (TMCs) and DPOAE data (not shown) and these help to clarify the picture further. We have been surprised by the variety of patterns of impairment in our group of volunteer (self-selected) participants in the study.

The results indicate that hypotheses concerning the underlying pathology can be tested for consistency with the data using computer models. As we gain confidence in the usefulness of the models and gain further experience in the admittedly black art of finding appropriate hypotheses and appropriate model parameters, we hope to be able to evaluate the effectiveness of different kinds of hearing aid algorithms and to make predictions as to their effectiveness for individual patients. To this end we are linking the model to automatic speech recognition devices. We acknowledge that the main complaint of patients involve difficulty in following conversations in noisy environments. We hope that our modelling efforts will eventually contribute to improved prostheses to minimize this problem.

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