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Stimulus-frequency otoacoustic emissions measured with amplitude-modulated suppressor tones

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Abstract

Stimulus-frequency otoacoustic emissions (SFOAE) are typically derived as the difference in sound pressure in the ear canal with and without a suppressor tone added to the probe tone. A novel variation of this method applies a sinusoidal amplitude modulation (AM) to the suppressor tone, which causes the SFOAE to also be modulated. The AM-SFOAE can be separated from the probe frequency using spectral methods. AM-SFOAE measurements are described for four normal-hearing subjects using 6-Hz AM. Because the suppressor modulation is at a higher rate, the AM-SFOAE technique avoids the confounding influence of heartbeat, which also modulates the probe tone.

I. INTRODUCTION

Stimulus-frequency otoacoustic emissions (SFOAE) are signals that originate within the cochlea at the same frequency as the evoking tonal stimulus, which we call the *probe* tone. The separation of the emission from the stimulus is a challenge for SFOAE measurements because they occur at the same frequency. To separate these responses, a common method for SFOAE measurement is to use two intervals, one with and one without a *suppressor* tone (e.g., Driesbach et al., 1998; Schairer et al., 2002; Shera and Guinan, 2003). The suppressor frequency (f_s) is chosen to be close enough to the probe frequency (f_p) to interact with it within the cochlea, but far enough away that spectral techniques can be used to distinguish it from the probe within the ear canal. The suppressor inhibits the cochlear generation of the SFOAE and causes the SFOAE measured in the ear canal to be reduced or eliminated. The SFOAE is then extracted from the response interval without the suppressor by subtracting the response in the interval with the suppressor. This subtraction is sometimes called *vector subtraction* because it includes both real and imaginary parts of the complex spectral components at the probe frequency.

A variation on the vector subtraction technique is explored here, in which the suppressor amplitude is modulated sinusoidally instead of being presented at a constant level. When this method is used, the influence of the suppressor on SFOAE generation varies continuously. As a result, the SFOAE measured in the ear canal is modulated in its amplitude at the same rate that the suppressor amplitude is modulated. This amplitude modulation allows the SFOAE to be separated from the probe using spectral techniques instead of vector subtraction.

II. METHODS

The theory behind the amplitude-modulated (AM) SFOAE technique is described first for the simpler case of constant probe level. Our implementation of the AM-SFOAE technique included an additional variation in probe level which occurred at a much slower rate than the

amplitude modulation. This probe level variation allowed input/output (I/O) functions to be measured efficiently, but is not an essential part of the AM-SFOAE technique.

A. Theory

The total ear-canal pressure can be modeled as the sum of four components

$$p(t) = p_{\text{stim}}(t) + p_{\text{oe}}(t) + p_{\text{ignor}}(t) + p_{\text{noise}}(t),$$

where $p_{\text{ignor}}(t)$ contains pressure components that are the result of higher-order distortion sidebands (e.g., due to nonlinearities in the earphone) that are synchronized to the stimulus and $p_{\text{noise}}(t)$ contains random pressure components (e.g., cable rub, swallowing, and heartbeat) that are not synchronized to the stimulus. The stimulus pressure waveform includes the probe and suppressor

$$p_{\text{stim}}(t) = p_{\text{probe}}(t) + p_{\text{suppr}}(t).$$

The probe component is a tone with constant amplitude A_p and frequency $\omega_p = 2\pi f_p$

$$p_{\text{probe}}(t) = A_p \cos(\omega_p t).$$

The suppressor component is a tone with constant frequency $\omega_s = 2\pi f_s$ and sinusoidally modulated amplitude

$$p_{\text{suppr}}(t) = A_s [1 + \alpha_s \cos(\omega_m t)] \cos(\omega_s t),$$

where α_s is the modulation depth ($0 \leq \alpha_s \leq 1$), $\omega_m = 2\pi f_m$ is the modulation frequency, and A_s is the unmodulated amplitude of the suppressor. The suppressor component can also be represented as the sum of three spectral components,

$$p_{\text{suppr}}(t) = A_s \left[\cos(\omega_s t) + \frac{1}{2} \alpha_s \cos(\omega_{sl} t) + \frac{1}{2} \alpha_s \cos(\omega_{su} t) \right],$$

where $\omega_{sl} = \omega_s - \omega_m$ and $\omega_{su} = \omega_s + \omega_m$ are lower and upper sideband frequencies, respectively, relative to the suppressor frequency. For simplicity, the tones and modulation are specified here as having zero phases.

When the probe and suppressor components combine within the cochlea, they will interact and produce additional distortion components due to the vibration of nonlinear elements. In addition, we assume that the probe will generate a reflected component at its tonotopic place. The reflected component $p_{\text{refl}}(t)$ will be modulated in amplitude by the presence of the suppressor at the tonotopic place where the probe is represented. We assume that $p_{\text{refl}}(t)$ will be at its maximum amplitude when the suppressor is at its minimum amplitude and vice versa

$$p_{\text{refl}}(t) = A_r \{1 - \alpha_r \cos[\omega_m(t - t_{in})]\} \cos[\omega_p(t - t_{in})],$$

where t_{in} represents the travel time (of both the probe and suppressor) to the tonotopic place of the probe and α_r is the modulation depth of the reflected component at the probe frequency. The reflected component can also be represented as the sum of three spectral components,

$$p_{\text{refl}}(t) = A_r \left\{ \cos[\omega_p(t - t_{in})] - \frac{1}{2} \alpha_r \cos[\omega_{pl}(t - t_{in})] - \frac{1}{2} \alpha_r \cos[\omega_{pu}(t - t_{in})] \right\},$$

where $\omega_{pl} = \omega_p - \omega_m$ and $\omega_{pu} = \omega_p + \omega_m$ are lower and upper sideband frequencies, respectively, relative to the probe frequency and A_r is the unmodulated amplitude of the reflected component. The reflected component will travel back to the ear canal, where we observe it as $p_{\text{oe}}(t)$. For simplicity, we choose to ignore the reflected component at the probe frequency and consider $p_{\text{oe}}(t)$ to be the sum of the upper and lower sidebands

$$p_{oae}(t) = A_{ol} \cos(\omega_{pl}t + \phi_{ol}) + A_{ou} \cos(\omega_{pu}t + \phi_{ou}).$$

If we assume that $p_{oae}(t)$ is related to $p_{ref}(t)$ by an outward travel time t_{out} and attenuation α_{out} , then $A_{ol} = A_{ou} = -\frac{1}{2}\alpha_{out}\alpha_r A_r$, $\phi_{ol} = -\omega_{pl} \cdot (t_{in} + t_{out})$, and $\phi_{ou} = -\omega_{pu} \cdot (t_{in} + t_{out})$. Note that when $\alpha_r = 1$, the sidebands contain half the total power of the reflected component. Note also that the difference in phase between the two sidebands provides an estimate of the round-trip travel time

$$\begin{aligned} \phi_{ol} - \phi_{ou} &= 2\omega_m \cdot (t_{in} + t_{out}) \\ t_{in} + t_{out} &= \frac{\phi_{ol} - \phi_{ou}}{2\omega_m}. \end{aligned}$$

B. Application

To explore the effectiveness of AM suppressors, we rapidly (6 Hz) modulated the *amplitude* of the suppressor while slowly (0.015 Hz) varying the *level* of both probe and suppressor. Our stimulus was 131 s in duration and contained two complete cycles of probe-level variation, ranging between 0 and 60 dB SPL. The suppressor, which was always present, varied in level between 30 and 70 dB SPL coincident with probe-level variation, while simultaneously being sinusoidally amplitude-modulated (99%) at a much higher rate of 6 Hz.

Our measurements were made using an ER-10C probe system (Etymotic Research) and locally-developed software (SYSRES, Neely and Stevenson, 2002). Frequency-domain heterodyne analysis (Kim et al., 2001; Guinan et al., 2003) was used to extract the slow variation of both stimulus and response component levels from the recorded waveforms.

The level variation of the probe (thick line) and suppressor (thin line) are illustrated in Fig 1. These responses were recorded with our measurement system in a 2-cm³ acoustic cavity, and show signs of being influenced by environmental noise at the lowest probe level. The bandwidth that was used to analyze the stimulus-component levels for this figure was wide enough to also reveal the 6-Hz modulation of the suppressor, shown by the rapid excursions of the upper line in the figure. Figure 2 shows the spectrum of the same cavity response. The (unmodulated) probe is situated at $f_p = 2$ kHz. The suppressor is centered at $f_s = 1920$ Hz with several sidebands at 6-Hz intervals from 1920 Hz. The spectrum of the digital stimulus had only one pair of sidebands. The other sidebands are presumably due to intermodulation distortion imposed by the ER-10C. However, this distortion does not contaminate the SFOAE measurement, because the cavity response demonstrates that there is no significant interaction between probe and suppressor in our measurement system. This conclusion is based on the observation that there are no sidebands within 40 Hz of the probe frequency f_p .

In addition to cavity measurements, we also made measurements in 4 normal-hearing ears and one deaf ear. The deaf ear contained a cochlear implant, which was not activated at the time of the measurement. The deaf ear was represents a biological system in which an SFOAE should not be present and provides a test of the linearity of our measurement system.

III. RESULTS

Spectral measurements from a cavity, deaf ear, and normal ear are compared in Fig. 3. The spectra are centered at $f_p = 2$ kHz and include a 10-Hz range on either side of f_p . The top panel (cavity) demonstrates the absence of interaction between the probe and suppressor in our measurement system as evidenced by the lack of any measurable response at $f_p \pm 6$ Hz. Likewise, the middle panel (deaf ear) shows no evidence of the influence of the suppressor, based on the absence of components at $f_p \pm 6$ Hz. However, the deaf ear shows clear evidence of modulation at 1.25 Hz, including harmonics of this frequency. The 1.25 Hz modulation is

presumably due to a heartbeat pulse rate of 75 per minute. Heartbeat modulation was frequently, but not always, observed in other ears. It was identified by the appearance of sidebands located 1 to 1.5 Hz relative to the probe frequency. In the lower panel, we see clear evidence of modulation at $f_p \pm 6$ Hz in addition to what is presumed to be heartbeat modulation. The modulation at $f_p \pm 6$ Hz is due to interaction between the probe and suppressor within the cochlea because it is not evident either in an acoustic cavity or in the deaf ear (i.e., a nonfunctioning cochlea). We used the level of the combined energy from both sidebands (L_{oae}) to represent the AM-SFOAE level. For the results shown in Fig. 3, the spectra were computed over the entire 131 s stimulus interval. As a result, the spectral levels are dominated by the maximum levels attained within that interval. In the case of a normal cochlea (bottom panel), the maximum SFOAE exceeded the noise floor by about 25 dB.

The probe and suppressor levels varied slowly in our stimuli to allow I/O functions to be determined for the range of levels over which the probe varied. These I/O functions were constructed by plotting the time course of the SFOAE level as a function of the time course of the probe level. Fig. 4 shows SFOAE I/O functions in four normal ears (superimposed) at each of four probe frequencies. The solid lines represent the combined energy of the upper and lower 6-Hz sidebands of the probe. Our stimulus actually produced two I/O functions: one for increasing probe level and one for decreasing probe level. Upward and downward excursions of the I/O function were averaged together to produce a single I/O function. The dashed lines represent the combined noise energy at both 6-Hz sideband frequencies, obtained by subtracting the first and second halves of the response waveform. Although there is variation in the I/O functions across subjects, some general trends were noted. The SFOAEs were largest at 4 kHz, but also were systematically present at 1 and 2 kHz. However, at 8 kHz, the SFOAEs were reduced in two subjects and did not exceed the noise floor at any probe level in the other two subjects.

IV. DISCUSSION

The AM-SFOAE technique described above may be attractive because it eliminates the confounding influence of probe-frequency modulation due to physiological influences, such as heartbeat. It also eliminates the need for vector subtraction, which may be an advantage when maintaining stationarity across two separate measurements is difficult. When combined with slow variation of probe and suppressor levels, the AM-SFOAE method can be used to obtain complete SFOAE I/O functions in a relatively small amount of time. However, the AM-SFOAE technique may also be attractive when used with constant-level probes because vector subtraction is not needed.

The AM-SFOAE technique could be used to estimate the round-trip travel time from the ear canal to the SFOAE generation site and back to the ear canal. As described above, this travel time estimate is based on the phase difference between the sidebands, and should be similar to other SFOAE round-trip travel time estimates (e.g., Shera and Guinan, 2003). Travel-time resolution might be improved by increasing the modulation rate above 6 Hz; however, the extent to which travel-time resolution can be improved is limited because the spectrum of the suppressor with higher modulation rates will spread to higher frequencies eventually overlap the probe frequency. The possible contamination of the response due to overlapping spectra makes it unclear whether a higher modulation rate would be beneficial to the AM-SFOAE technique.

The suppressor level was selected in this study to keep it above the probe level without becoming excessively loud at the highest probe levels. Some attempts were made to determine suppressor levels that would produce the largest AM-SFOAE levels in our four subjects;

however, the results were not consistent across subjects or frequencies. Hence, it is still undetermined what suppressor levels provide the best results with this technique.

SFOAE I/O functions recently described by Schairer et al. (2003) were about 10 dB higher in level than those shown in Fig. 4. This difference might be attributed to two factors. (1) The sidebands contain only half the power of the total reflected component, so the AM-SFOAE level would be expected to be 3-dB lower than the SFOAE level obtained by vector subtraction. (2) Our suppressor levels were lower than those used by Schairer et al. and probably did not achieve 100% amplitude modulation of the reflected component level. Despite the lower I/O function levels, the AM-SFOAE may still have the advantage of greater signal-to-noise ratio. The I/O functions in Fig. 4 are sometimes 25 dB above our measured noise, whereas the Schairer et al. I/O functions were rarely more than 15 dB above their measured noise. They attributed the elevated noise at the highest probe levels to an unknown biological source. In light of the present results, the increased noise that they observed may have been due to heartbeat modulation, which was not separated from the SFOAE by their measurement technique.

A different SFOAE technique that involved amplitude modulation was recently described by Goodman et al. (2004). In that case, the probe (not the suppressor) was modulated and the purpose of the modulation was to investigate OAE latency, not SFOAE level. Their latency estimates were based on temporal features of the SFOAE measurement, instead of the spectral methods typically employed to measure SFOAE latency. Although the use of AM-SFOAE measurements to estimate travel time was described above as a spectral method, an alternative approach based on temporal features should work equally well.

V. CONCLUSIONS

Measurement of SFOAEs using an amplitude-modulated suppressor appears to be a viable technique which may have some advantages over other methods. In particular, the spectral separation of the SFOAE allows the contaminating influence of heartbeat modulation to be eliminated and does not require vector subtraction to identify the SFOAE.

It is not necessary to vary the probe level to use this method (although doing so provides a means of obtaining the entire SFOAE I/O function quickly). Amplitude-modulated suppressors may also provide an effective means of measuring SFOAEs with constant probe levels and stimuli less than 1 s in duration.

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REFERENCES

- Dreisbach LE, Siegel JH, Chen W. Stimulus-frequency otoacoustic emissions measured at low and high frequencies in untrained human. *Assoc. Res. Otolaryngol.* 1998Abs. 349
- Goodman SS, Withnell RH, de Boer E, Lilly DJ, Nuttall AL. Cochlear delays measured with amplitude-modulated tone-burst-evoked OAEs. *Hear. Res* 2004;188:57–69. [PubMed: 14759571]
- Guinan JJ, Backus BC, Lilaonitkul W, Aharonson V. Medial olivocochlear efferent reflex in humans: otoacoustic emission (OAE) measurement issues and the advantages of stimulus frequency OAEs. *J. Assoc. Res. Otolaryngol* 2003;4:521–540. [PubMed: 12799992]
- Kim DO, Dorn PA, Neely ST, Gorga MP. Adaptation of distortion-product otoacoustic emission in humans. *J. Assoc. Res. Otolaryngol* 2001;2:31–40. [PubMed: 11545148]
- Neely, ST.; Stevenson. Tech. Memo. 19, Boys Town, Nat. Res. Hosp. Omaha, NE: 2002. SysRes.

- Schairer KS, Fitzpatrick D, Keefe DH. Input-output functions for stimulus-frequency otoacoustic emissions in normal-hearing adult ears. *J. Acoust. Soc. Am* 2003;114:944–966. [PubMed: 12942975]
- Shera CA, Guinan JJ. Stimulus-frequency-emission group delay: A test of coherent reflection filtering and a window on cochlear tuning. *J. Acoust. Soc. Am* 2003;113:2762–2772. [PubMed: 12765394]

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