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A time-domain digital simulation of hearing aid response

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Abstract—A time-domain digital simulation of an in-the-ear (ITE) hearing aid has been developed. The simulation allows modeling of nonlinear effects such as compression and amplifier distortion in addition to linear processing and acoustics. The simulation includes a microphone, two-channel compression processing, an amplifier with clipping distortion, a receiver, an ear canal and ear drum, and feedback and direct sound transmission through the vent. Simulation results for a linear hearing aid are similar to those obtained for frequency-domain representations of the analog system. Examples of responses for nonlinear systems are also provided.

Key words: *amplifier, analog system, in-the-ear (ITE) hearing aid, microphone, time-domain digital simulation, tone-burst responses, two-channel compression processing.*

INTRODUCTION

This paper presents a time-domain digital simulation of an in-the-ear (ITE) hearing aid. Previous computer simulations, such as those of Egolf, *et al.* (4), Bade, *et al.* (1), and Kates (5) have all been in the frequency domain and limited to representing the linear behavior of a hearing aid. The time-domain simulation, however, extends the modeling to include nonlinear effects such as compression and amplifier distortion. This results in a more complete computer simulation of a hearing aid than previously available, and one which can be used for many applications such as the development of hearing aid signal processing and hearing aid test systems.

A time-domain model is essential for simulating the effects of nonlinear or linear time-varying systems in which the processing changes in response to the input signal behavior. Examples in hearing aids include compression in one or more frequency channels, compression with adaptive time constants, adaptive filters, noise suppression, and feedback cancellation. The design and evaluation of such algorithms can be greatly facilitated by using a computer simulation during the processing development. Distortion is also of interest; a common example is amplifier saturation. Modeling the distortion characteristics of typical circuits makes it possible to isolate the effects of distortion on sound quality and speech intelligibility in hearing aids.

The time-domain simulation is based on the frequency-domain simulation of Kates (5). That simulation includes the linear effects of a microphone, amplifier, receiver, vent, ear canal and ear drum, and pinna. The results of the frequency-domain model were shown by Kates (5) to be accurate in reproducing the measured behavior of an ITE hearing aid mounted on the KEMAR anthropometric manikin (2) and to be useful in predicting the effects of varying the acoustics of the hearing aid and the ear (5,6). The purpose of the time-domain simulation is not, however, to reproduce the behavior of a specific hearing aid, but rather to simulate a generic hearing aid incorporating signal processing of interest.

In developing the time-domain simulation, most of the elements of the frequency-domain model of Kates (5) were kept by converting the frequency response of each element in the model into a corresponding time-domain digital filter

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response. The pinna effects, which convert the free-field excitation into the pressure at the microphone, have not been included because they are equivalent to a modification of the excitation signal and can be added at a later date if needed. The amplifier, instead of modeling a specific manufacturer's hearing aid, has been replaced by an ideal voltage or current source with an adjustable hard-clipping threshold to simulate an amplifier with distortion.

Because the simulated amplifier does not duplicate the behavior of any specific hearing aid, validation of the model is based on general behavior and the correspondence with earlier models rather than with matching a set of individual hearing aid temporal responses. Hearing aid processing has been added to the model to give a two-channel compression hearing aid with independent input-referred automatic gain control (AGC-I) in each channel and an adjustable crossover frequency between the two channels.

The purpose of this paper is to describe the time-domain simulation and to give some examples of its use. The underlying physics of the model has been treated by Kates (5) and is therefore not repeated here. The examples show the frequency response of a simulated linear system with acoustic feedback computed from the system impulse response; the temporal response of a simulated hearing aid to a 1-kHz tone burst for three processing options (ideal linear, the same instrument with clipping distortion, and two-channel compression); and the response of the three processing systems to a two-tone test signal used for the measurement of intermodulation distortion.

ELEMENTS OF THE SIMULATION

The computer program simulates a standard (full-concha) ITE hearing aid. The major external features of the hearing aid are the microphone opening, battery compartment, volume control, vent openings, and receiver opening. A cross-section of the typical ITE hearing aid is presented in **Figure 1**. The microphone is positioned near the top of the hearing aid faceplate above the battery compartment and the volume control; the vent is at the bottom. Not shown is circuitry internal to the instrument; it is positioned where there is available space. The receiver is located in the canal portion of the hearing aid, and the receiver output is conducted to the ear canal via a short tube. The vent runs from the faceplate to the ear canal and can take several shapes, although a tube having circular cross-section is the most common and is a reasonable acoustic model for other geometries as well. The hearing aid fits entirely in the ear: the portion containing the receiver is inserted in the ear canal and the portion containing the microphone and battery compartment rests in the concha area of the outer ear.

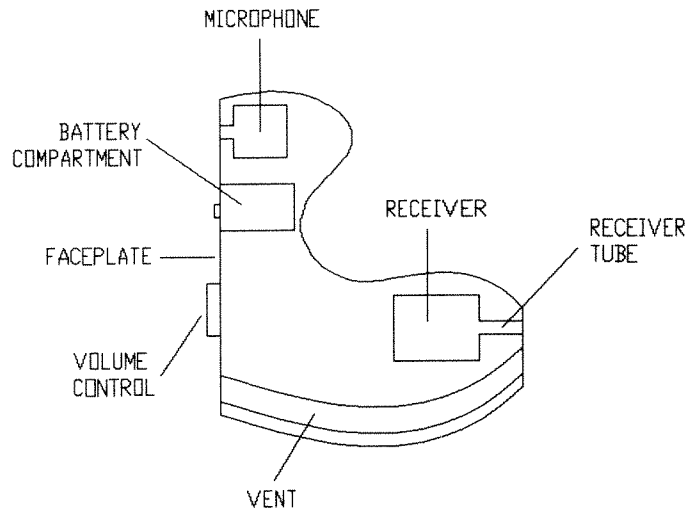

Figure 1.
Cross-section of an in-the-ear (ITE) hearing aid.

A block diagram of the simulated instrument for 2 cm^3 coupler measurements is presented in **Figure 2**. The input to the microphone is the free-field sound pressure generated by an ideal loudspeaker because no pinna or head effects are present. The microphone output is split into a low-pass and a high-pass channel, with independent input-referred compression (AGC-I) in each channel. The gain in each channel is then adjusted; following this, signals are summed. The power-amplifier behavior is represented as a symmetric clipping stage followed by a fixed gain. The receiver is assumed to be loaded acoustically by a tube leading to a 2 cm^3 coupler; any vent present is ignored because it is not connected to the coupler for this type of measurement.

A block diagram of the simulated instrument inserted into an ear is presented in **Figure 3**. The vent now becomes part of the system. In addition to the amplified signal path shown in **Figure 2**, there is also an unamplified signal path directly through the vent, so the sound pressure in the ear canal is the sum of the amplified and direct signals. Acoustic feedback is included in the model as sound from the ear canal being transmitted through the vent to the faceplate where it is reradiated and picked up by the microphone. The receiver, connected to the ear canal by a short tube, is loaded by tube, vent, and ear canal. The ear canal is terminated by the modified Zwislocki coupler used in the KEMAR manikin. The digital time-domain simulation has been implemented at a sampling rate of

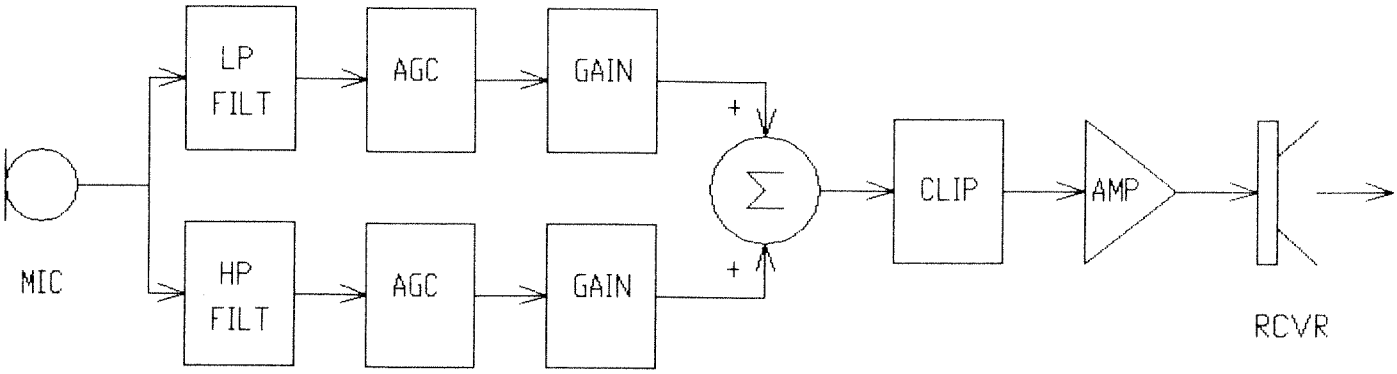

Figure 2.
Block diagram of the simulated hearing aid for 2 cm^3 coupler measurements.

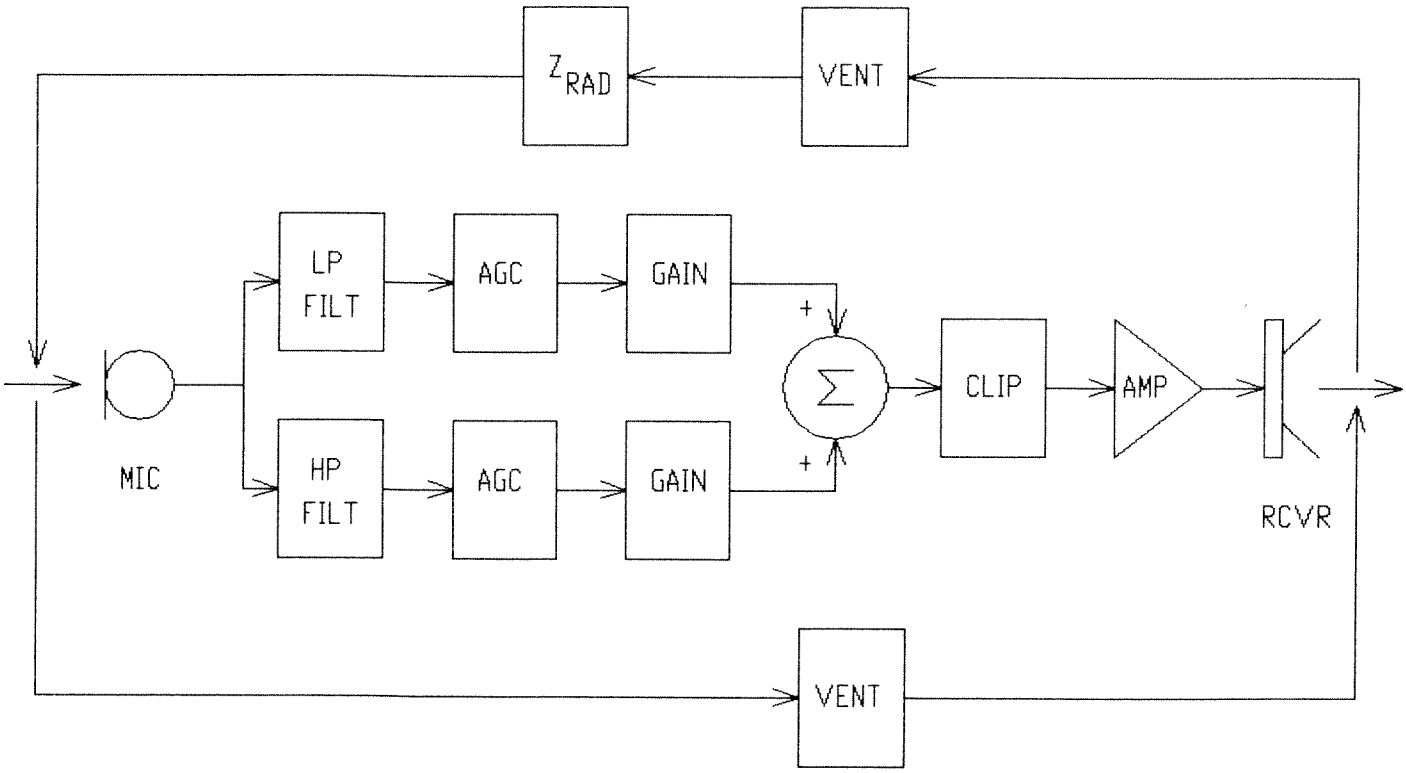

Figure 3.
Block diagram of the simulated hearing aid with vent.

20 kHz using the elements described in the following paragraphs.

Microphone

The microphone is a digital transformation of the transfer function of a Knowles EA-1842 microphone as modeled by Kates (5). The microphone is represented as a one-pole high-pass filter at 300 Hz in cascade with a two-pole low-pass filter at 5100 Hz having a Q of 1.55. The digital high-pass filter was implemented using the bilinear transformation (7) and the digital low-pass filter was implemented using impulse invariance (7). The frequency response of the digital microphone simulation is shown in **Figure 4**. Note that the microphone response has been adjusted to give 0 dB gain at 1 kHz.

Filters

The low-pass and high-pass filters are three-pole Butterworth filters. The digital filters are bilinear transformations of the analog transfer functions, with the crossover

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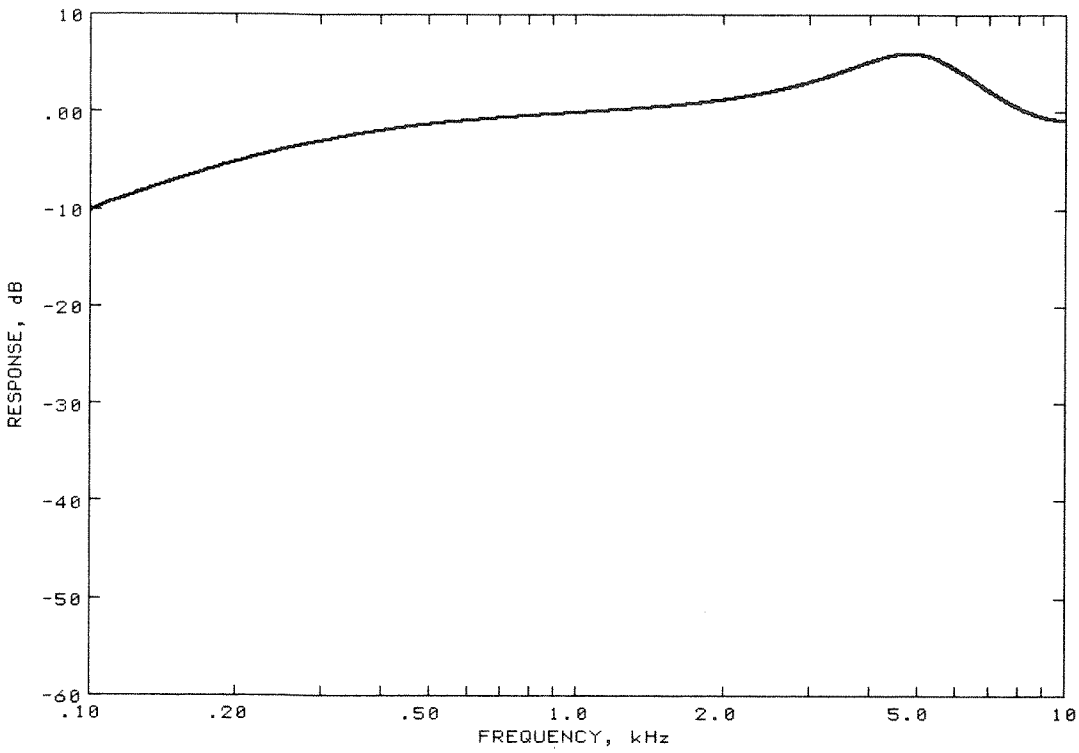

Figure 4.
Transfer function for the simulated Knowles EA-1842 microphone.

frequency a parameter of the hearing aid simulation. The digital filter responses are shown in **Figure 5** for a crossover frequency of 1 kHz. One property of these filters is that the sum of the low-pass and high-pass filters gives a smooth frequency response for any setting of the relative gains of the two filters. Other filter designs may not share this feature. The overall response is illustrated in **Figure 6**, where the gain of the high-pass filter is increased from 0 dB to 30 dB in steps of 10 dB, while the gain of the low-pass filter is kept at 0 dB. The combination of the two filters yields an overall frequency response that monotonically increases with increasing frequency.

AGC-I

The two frequency channels have identical independent AGC-I circuits. The analog prototype peak-detection circuit used in the AGC is shown in **Figure 7**. The circuit input is the absolute value of the low-pass or high-pass filter output. The attack time constant is given by $\tau_1 = R_1C$, under the assumption that the attack time is much shorter than the release time. The release time constant is given by $\tau_2 = R_2C$. The peak voltage v(t) is used to control the AGC gain for the channel. Circuit equations were transformed using backwards differences (7) to give the digital simulation, with the attack and release times in each channel parameters of the model.

The input/output behavior of each AGC circuit is shown in **Figure 8**. The system is linear as long as the control voltage v(t) is below the AGC threshold. Compression is engaged when the control voltage exceeds the threshold, with the threshold and the compression ratio a parameter of the simulation for each channel. The compression behavior is therefore governed by the estimated peak input signal levels. The compression threshold is specified in terms of the equivalent input sound-pressure level in dB sound-pressure level (SPL) at 1 kHz, so the actual compression threshold is modified by the microphone response of **Figure 4** and the low-pass or high-pass filter response from **Figure 5**.

Amplifier

The gain of the simulated system is specified for each frequency channel separately; following this, the signals are combined. The simulated amplifier then converts this signal into a form suitable for driving the receiver. The amplifier is modeled as a voltage or current source with adjustable clipping distortion; a voltage source is commonly used in high-power hearing aids and a current source is commonly used in low- to moderate-power hearing aids. The amplifier output impedance is either zero ohms for

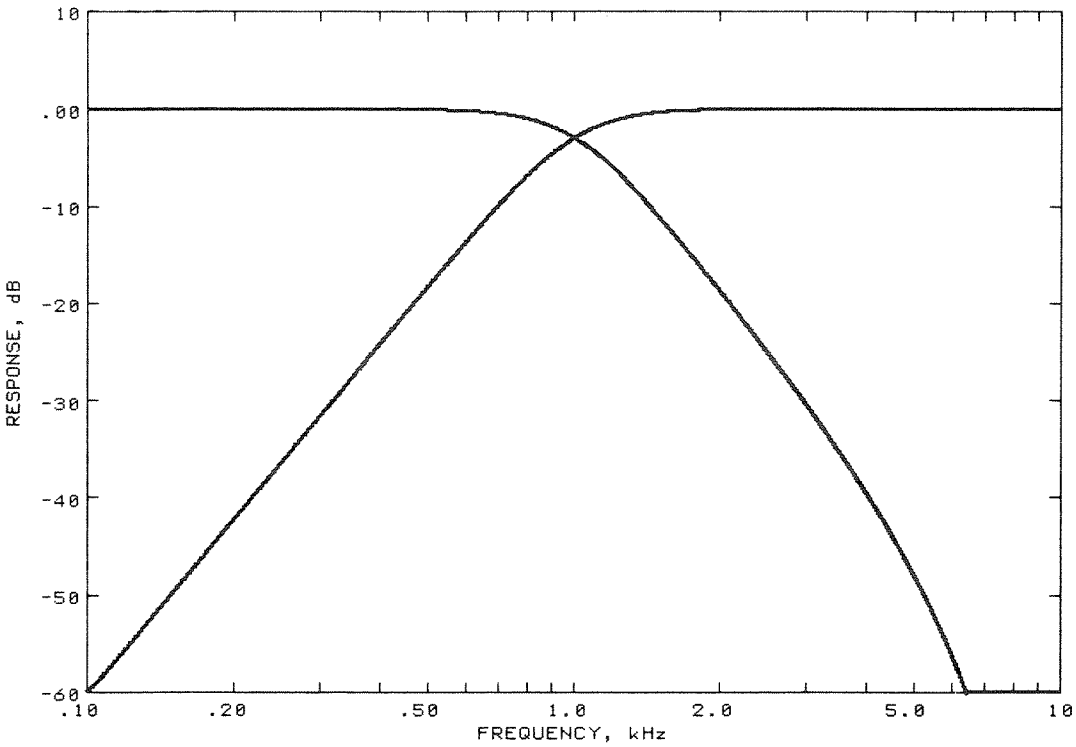

Figure 5.
Low-pass and high-pass filter frequency responses for the simulated hearing aid.

the voltage source or 100,000 ohms for the current source. The amplifier distortion is simulated as symmetric clipping of either the voltage or the current, depending on the type of source selected.* The clipping level is specified in terms of the equivalent input peak SPL at 1 kHz with 0 dB gain in both frequency channels, so the actual clipping level is modified by the microphone response and the gain in each frequency channel.

Receiver

The digital receiver simulation, based on the electrical analog of Carlson (3), was implemented by Kates (5) for the frequency-domain simulation. The output impedance of the amplifier is included in the receiver model, as is the specified acoustic load of ear canal, ear drum, and vent. The resultant receiver equivalent circuit model is very complicated, so a frequency-sampling approach (7) was used to generate the time-domain response instead of an analytical technique such as the bilinear transformation.

To compute the time-domain simulation for the sampling rate of 20 kHz, the complex analog frequency-domain transfer function was evaluated at 512 points uniformly spaced from 0 to 10 kHz. This frequency response was transformed into the time-domain via an inverse fast Fourier transform (FFT) to give a 1,024-point impulse response. The number of samples was chosen to minimize aliasing in the impulse response while keeping a reasonable computational burden. The impulse response was then windowed to 256 samples using the function given below:

$$w(n) = \begin{cases} 1 & 0 \le n \le 127 \\ 0.5\left[1 + \cos(\pi \frac{n-128}{128})\right] & 128 \le n \le 255 \\ 0 & 256 \le n \le 1023 \end{cases} \quad [1]$$

The windowed sequence yields a finite impulse response (FIR) digital approximation to the analog system. The amplifier output signal is then convolved with the FIR receiver response to give the receiver output.

The resultant frequency response for a Knowles ED-1913 receiver connected to a 2 cm^3 coupler is shown in **Figure 9**. The solid line is the frequency response computed from the frequency-domain model of the analog system. The dashed line is the Fourier transform of the windowed digital impulse response. The pressure measurement point is at the junction of the receiver tube with the ear canal at the end of the hearing aid. The agreement between the two curves is quite good since they overlap over almost the entire frequency range, thus showing the

*Personal communication, S. Armstrong, Gennum Corp., 1989.

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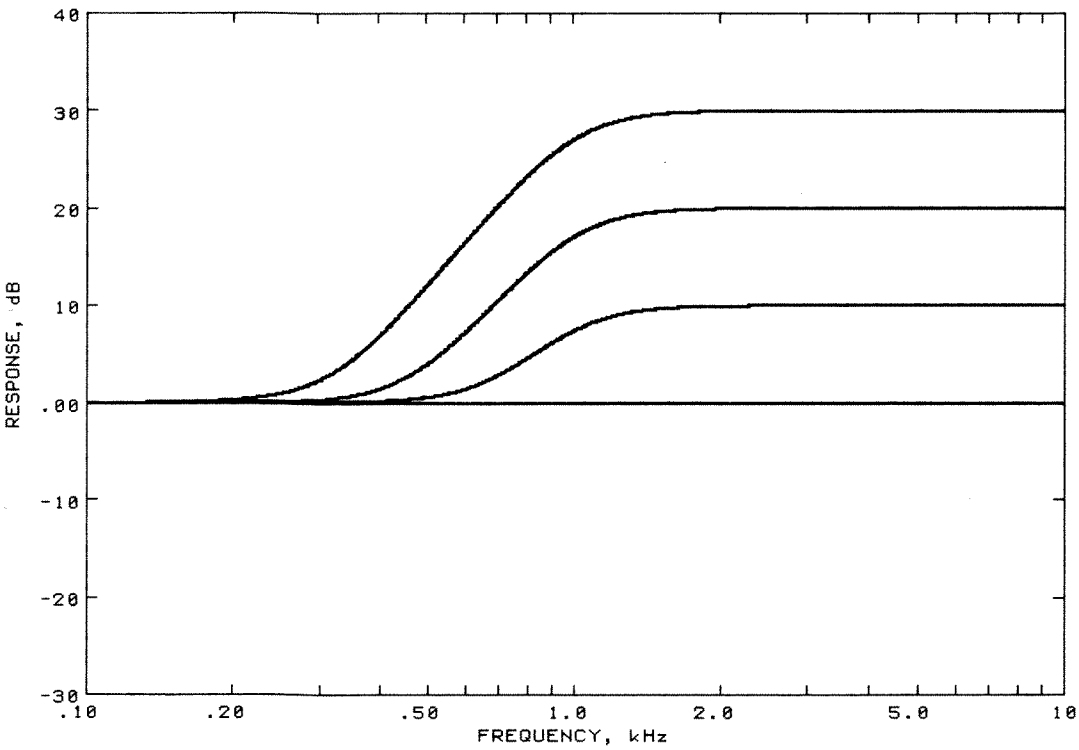

Figure 6.
Summed filter outputs as the gain of the high-pass filter channel is increased from 0 dB to 30 dB in steps of 10 dB. The gain of the low-pass filter channel is 0 dB.

accuracy of this simulation approach.

Figure 10 shows the frequency response for the same receiver loaded with a simulated ear and vent, but with the feedback and feedforward signal paths through the vent left unconnected. The parameters for the ear and vent are given in **Table 1**; these values are similar to those used by Kates (5,6) in simulating an actual vented hearing aid. The solid curve in the figure is again the frequency response from the frequency-domain simulation; the dashed line is the Fourier transform of the windowed impulse response. There is some minor discrepancy between the two curves in the low-level response regions, approximately 40 dB below the peaks, due to the effects of the window and the sampling rate. Such deviations will occur for any transformation of an analog system into the digital domain. The agreement at high response levels, however, is still very good. In particular, the effect of the vent at low frequencies, where the mass of the air in the vent and the compliance of the air in the ear canal interact to produce a high-pass filter, are quite apparent in the peak at 600 Hz and the reduction in response below that frequency. Because there is no feedback for this response calculation, the effect at high frequencies is primarily a notch at 7 kHz caused by the vent half-wave resonance modifying the acoustic load on the receiver.

Table 1.
Acoustic parameters of the simulated ear and hearing aid.

Receiver termination model	
Type:	tube
Length:	0.6 cm
Radius:	0.06 cm
Ear canal model (unoccluded portion)	
Type:	tube
Length:	1.2 cm
Radius:	0.33 cm
Ear-drum impedance:	modified Zwislocki coupler
Vent model	
Type:	tube
Length:	2.2 cm
Radius:	0.12 cm
Vent-Mic distance:	1.8 cm

Vent

The vent carries sound in both directions, as shown in the block diagram of **Figure 3**. It requires two different digital simulation responses because the acoustic source

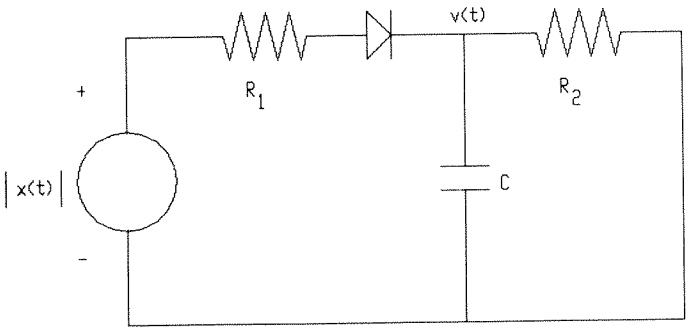

Figure 7.
Peak-detection circuit simulated in each channel of the hearing aid.

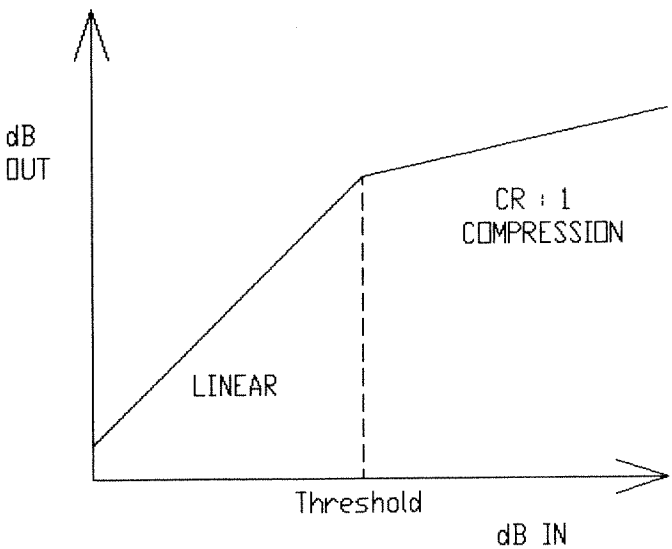

Figure 8.
Input/output characteristics of each channel in the simulated hearing aid.

and load impedances are different for the two transmission paths through the vent (5). The vent responses were computed using the same procedure as used for the receiver response.

The feedback-path frequency response, including the radiation impedance at the vent opening in the faceplate and the propagation from the vent to the microphone, is shown in **Figure 11** for the vent and ear parameters given in **Table 1**. The solid line is the frequency response computed from the frequency-domain simulation; the dashed line is the Fourier transform of the windowed impulse response. The agreement between the two curves is quite good, with both showing an overall gain of about −56 dB at low frequencies. The feedback path has a sharp resonance peak near 7 kHz. At this frequency, the effective length of the vent loaded by the radiation impedance is approximately a half wavelength.

The peak in the vent feedback response at 7 kHz does not mean that the hearing aid will tend to go into oscillation at this frequency. Oscillation is symptomatic of an unstable system, in which the amplitude and phase of the feedback path combine to give nearly unity gain. The vent response near 7 kHz is compensated by the low receiver output in the same frequency region, and the overall phase shift results in a notch rather than a peak for the system represented in **Figure 11**. In most cases, oscillation occurs at frequencies in the vicinity of the receiver response peaks since this is where the instrument has the highest gain. The exact frequency will depend on the phase shifts as well as the amplitudes of the amplified and feedback signal paths.

The direct-path response for the same simulated vent and ear is shown in **Figure 12**. The basic behavior of the direct path through the vent is a low-pass filter. The resonance frequency is determined by the interaction of the mass of air in the vent and the compliance of the air in the ear canal, and the sharpness of the resonance affected in addition by the losses in the vent and ear. The vent response also shows the high-frequency resonance, but the resonance frequency has moved to about 7.5 kHz due to the difference in the vent load impedance. Again, the frequency response computed from the windowed impulse response (dashed line) is in good agreement with the frequency-domain response (solid line).

EXAMPLES

Frequency response

The frequency response of the simulated linear system was obtained from the Fourier transform of the system impulse response for a hearing aid with and without a vent. The parameters of the simulated linear instrument are given in **Table 2**. The parameters for the ear are from **Table 1**, and the vent (when included) is also specified in **Table 1**. The time-domain simulation was run at a sampling rate of 20 kHz.

The frequency response of the hearing aid without the vent is given by the dashed line in **Figure 13**. Because both the low-pass and the high-pass processing channel have the same gain, the overall magnitude frequency response of the simulated hearing aid is that of the microphone plus receiver. This curve is quite similar to the unvented response curves computed by Kates (5) using the frequency-domain simulation for the same microphone, receiver, and model parameters, but with a different simulated amplifier.

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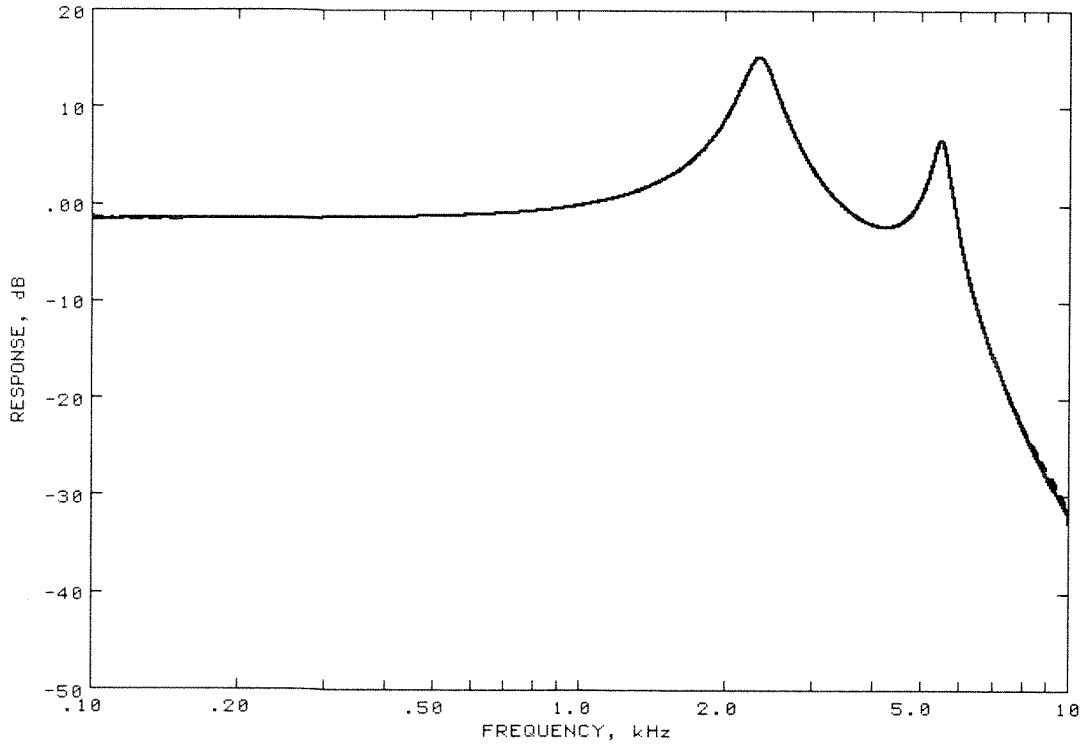

Figure 9.
Coupler pressure transfer function for the simulated Knowles ED-1913 receiver terminated with a tube 0.6 cm long and having a radius of 0.06 cm leading to the 2 cm^3 coupler. The solid line is the target frequency response, the dashed line the simulation.

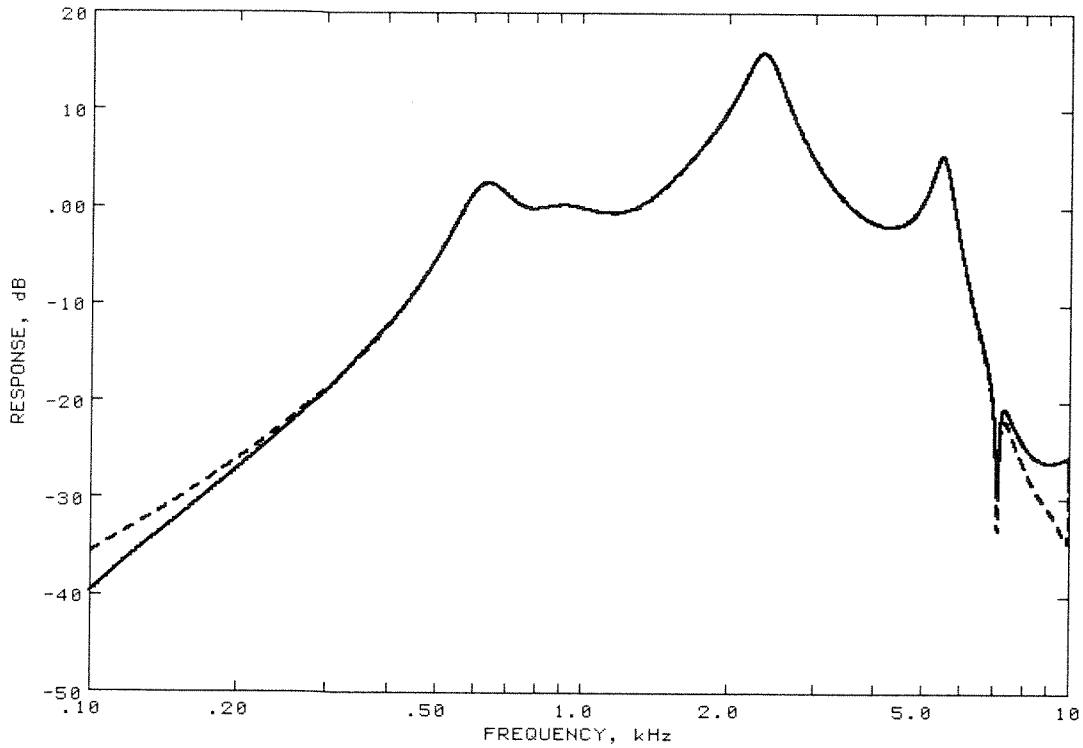

Figure 10.
Pressure transfer function for the simulated Knowles ED-1913 receiver terminated with the tube, vent, and ear canal specified in **Table 1.** Pressure is measured at the tube opening into the ear canal. The solid line is target frequency response, dashed line the simulation.

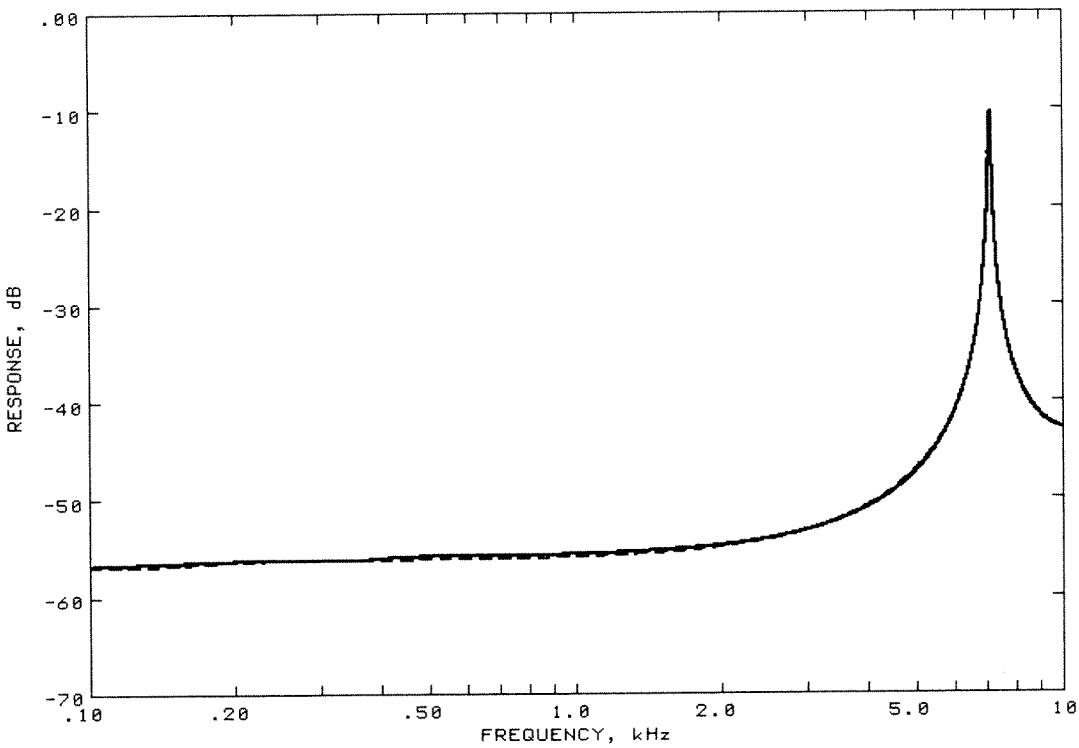

Figure 11.
Pressure transfer function for the feedback path through the vent and back to the microphone for the vent specified in **Table 1.** The solid line is the target response, the dashed line the simulation.

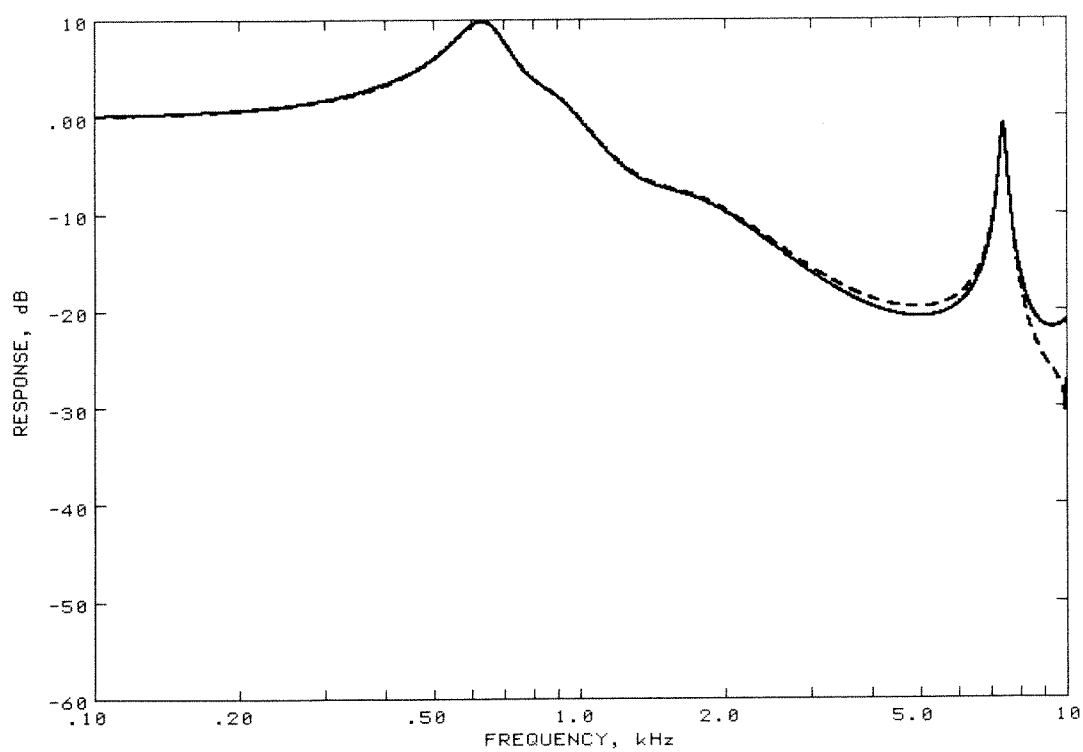

Figure 12.
Pressure transfer function for the direct signal path through the vent as specified in **Table 1.** The solid line is the target response, the dashed line the simulation.

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Table 2.
Parameters of the linear hearing aid without clipping.

Microphone:	Knowles EA-1842
Receiver:	Knowles ED-1913
Amplifier	
Type:	Current Source
Gain:	32 dB
Clipping level:	120 dB SPL
Crossover frequency:	2 kHz
Low-frequency channel	
Gain:	0 dB
Compression ratio:	1:1
Compression threshold:	Does not apply
Attack time:	Does not apply
Release time:	Does not apply
High-frequency channel	
Gain:	0 dB
Compression ratio:	1:1
Compression threshold:	Does not apply
Attack time:	Does not apply
Release time:	Does not apply

The major features of the curve are the two response peaks at about 2.2 and 5.4 kHz due to the receiver, and the low-frequency slope from the microphone.

The frequency response of the vented hearing aid is given by the solid line of **Figure 13**. The gain of the hearing aid amplifier is still 32 dB. At low frequencies, the resonance of the vent with the ear canal has resulted in a peak at about 600 Hz. Below the resonance peak, the gain is reduced due to the high-pass filtering provided by the vent frequencies below 200 Hz, the direct signal path through the vent starts to dominate the response and the slope of the curve is reduced. The gain at 100 Hz is therefore about 0 dB since most of the signal energy in this frequency region comes from the direct path.

The vent also has a strong effect at high frequencies. The feedback has increased the Q of both receiver resonances, resulting in peaks that are higher by several dB. The exact nature of the feedback effects, however, depends on the phase as well as the magnitude of the frequency response. Changing the phase response of the hearing aid processing will thus alter the details of the feedback effects in terms of which peak is more strongly affected and the maximum gain for which the system remains stable. An additional effect that is visible in the frequency response of **Figure 13** is the notch in the receiver output at 7 kHz due to the vent resonance.

Tone-burst response

In addition to modeling the behavior of a linear system with feedback, the time-domain simulation can also be used to model the behavior of a nonlinear system. Two common nonlinearities that occur in hearing aids are clipping distortion (due to amplifier saturation), and amplitude compression. The temporal response of a simulated hearing aid to a 1-kHz tone burst having a root mean square (RMS) level of 85 dB SPL was computed to illustrate the behavior of a hearing aid when nonlinear processing is present in the instrument.

The reference condition is the vented linear hearing aid specified in **Table 2** and used for the frequency-response of **Figure 13** (solid curve). The first 400 samples (20 ms) of the response of the linear instrument to the 1-kHz tone burst are shown in **Figure 14**. The amplitude scale is arbitrary, but is the same for all three figures in this section. The initial portion of the response shows the transients due to the sudden application of the sinusoid. There is a small amount of overshoot combined with components at other frequencies excited by the rising edge of the tone-burst. The envelope of the system response then oscillates slightly before reaching steady-state output at about 200 samples (10 ms).

The response to the same tone-burst was then repeated for the same hearing aid, but with amplifier clipping at an input-referred peak level of 85 SPL. The amplifier will thus clip the peaks of the input sinusoid. The hearing aid parameters are given in **Table 3**. This condition is still referred to in the industry as "linear" when discussing hearing aid processing, despite the presence of amplifier clipping or saturation, since no intentional nonlinear processing such as automatic gain control (AGC) has been designed into the instrument.

The tone-burst response is shown in **Figure 15**, where one can see that the clipping distortion has modified both the envelope of the transient response and the waveform shape of the steady-state response. The initial transient, especially between samples 20 (1 ms) and 80 (4 ms), shows a greater amount of high-frequency energy. The overshoot in this region, however, is similar to **Figure 14** since the clipping occurs before the receiver response and the transients, even when clipped, are still shaped by the receiver transfer function as modified by the vent and feedback. The steady-state response is at a slightly lower level than for the system without clipping. The distorted waveform shape clearly shows the presence of harmonic distortion. Thus, the effects of clipping are evident in the simulated temporal response.

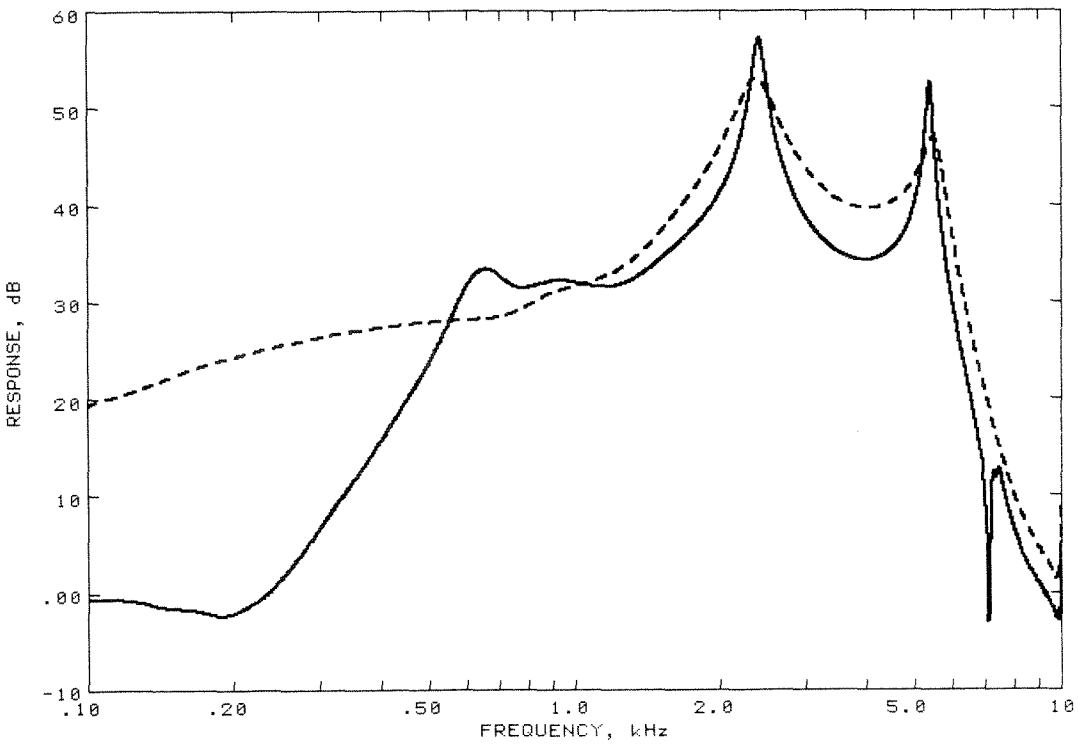

Figure 13.
Frequency response for the linear system with feedback as specified in **Table 2.** The solid line is the response for the vent open, the dashed line the response for the vent blocked.

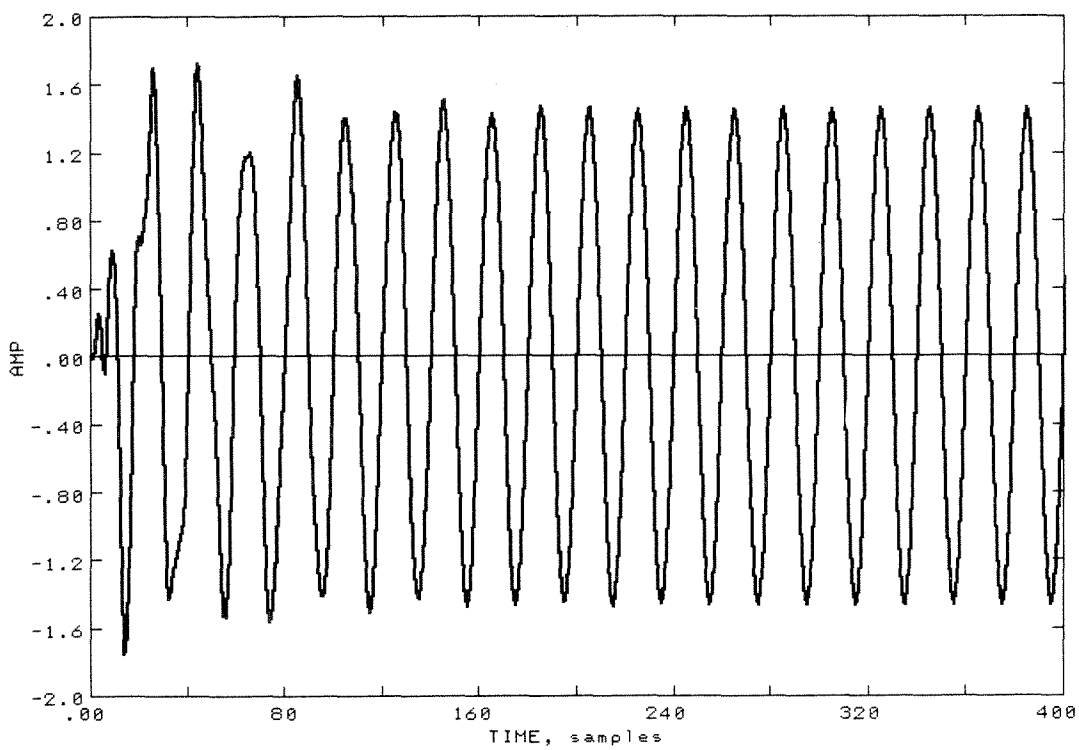

Figure 14.
Response of the simulated linear hearing aid of **Table 2** to an 85 dB SPL tone burst at 1 kHz. The sampling rate is 20 kHz.

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Table 3.
Parameters of the linear hearing aid with clipping.

Microphone:	Knowles EA-1842
Receiver:	Knowles ED-1913
Amplifier	
Type:	Current Source
Gain:	32 dB
Clipping level:	85 dB SPL
Crossover frequency:	2 kHz
Low-frequency channel	
Gain:	0 dB
Compression ratio:	1:1
Compression threshold:	Does not apply
Attack time:	Does not apply
Release time:	Does not apply
High-frequency channel	
Gain:	0 dB
Compression ratio:	1:1
Compression threshold:	Does not apply
Attack time:	Does not apply
Release time:	Does not apply

Table 4.
Parameters of the two-channel AGC-I hearing aid

Microphone:	Knowles EA-1842
Receiver:	Knowles ED-1913
Amplifier	
Type:	Current Source
Gain:	32 dB
Clipping level:	85 dB SPL
Crossover frequency:	2 kHz
Low-frequency channel	
Gain:	0 dB
Compression ratio:	4:1
Compression threshold:	75 dB SPL
Attack time:	1 ms
Release time:	50 ms
High-frequency channel	
Gain:	0 dB
Compression ratio:	2:1
Compression threshold:	65 dB SPL
Attack time:	1 ms
Release time:	50 ms

The response of a two-channel compression hearing aid to the 85 dB SPL tone-burst is shown in **Figure 16**. The hearing aid parameters are given in **Table 4**; the gains in the two channels are the same but the compression characteristics are different. The frequency response of the compression hearing aid will be identical to that of the previous examples for an input below the compression thresholds, but will differ for an input such as the tone burst which is above threshold. The initial transient for the compression instrument is different than for the previous examples primarily because the fast compression attack has limited the amplitude of the 1 kHz tone. Thus, the transient components due to the receiver response are more readily visible for this case. The steady-state portion of the response shows the decreased amplitude of the tone due to the compression, with no apparent distortion for the pure-tone excitation.

Intermodulation distortion

A test signal consisting of two tones was used to illustrate the ability of the time-domain model to simulate intermodulation distortion effects. The test signal consisted of a pair of sinusoids at 1.0 and 1.2 kHz, each at an RMS level of 82 dB SPL. The resultant signal had an RMS level of 85 dB SPL and a crest factor of 6 dB. The simulation sampling rate was again 20 kHz. The output spectrum computed from 4,096 samples of the steady-state response of the vented linear hearing aid specified in **Table 2** is shown in **Figure 17**. The dB scale is arbitrary, but is the same for **Figure 17, Figure 18**, and **Figure 19**. The peaks due to the two sinusoids are quite apparent, and there is essentially no distortion. Introducing amplifier clipping at a level of 85 dB SPL, as specified in **Table 3**, results in the spectrum of **Figure 18**. There are a large number of distortion products generated by the clipping, with major peaks found in the vicinity of the third and fifth harmonics of the excitation sinusoids. The distortion spectrum is shaped by the receiver frequency response with its peaks at 2.2 and 5.4 kHz, since the receiver follows the amplifier with its distortion, so distortion products in the vicinity of the receiver peaks are given additional emphasis. Adding the compression specified in **Table 4** results in the output spectrum of **Figure 19**. The compression processing results in a small amount of intermodulation distortion, typically about 1 to 2 percent, but this level of distortion is substantially lower than the results of amplifier saturation for the same input level and clipping threshold.

CONCLUSIONS

This paper has presented a time-domain digital computer simulation of an ITE hearing aid. The simulation includes a microphone, two-channel compression process-

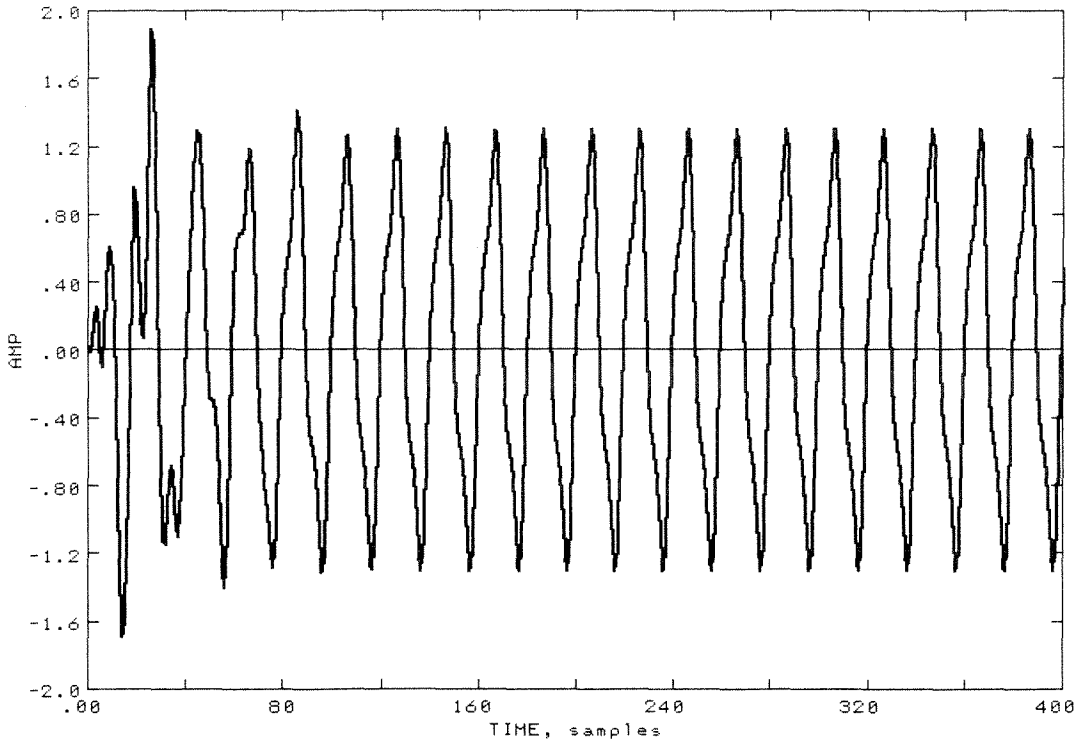

Figure 15.
Response of the simulated linear hearing aid with clipping, as specified in **Table 3**, to an 85 dB SPL tone burst at 1 kHz. The sampling rate is 20 kHz.

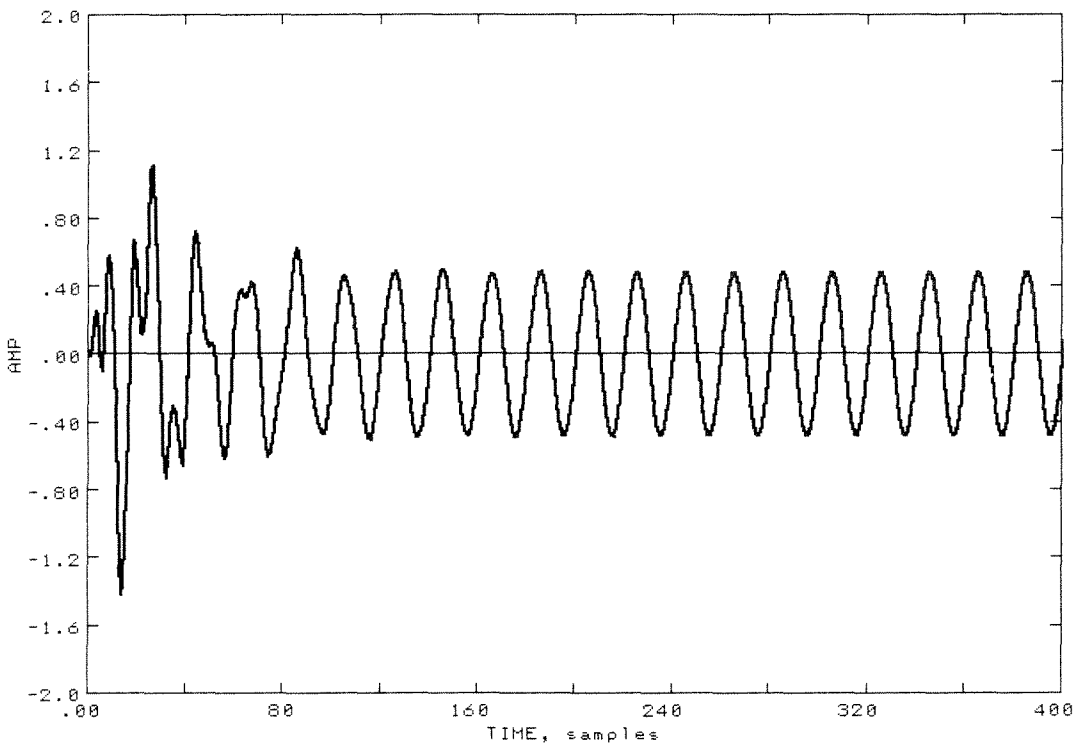

Figure 16.
Response of the simulated two-channel compression hearing aid of **Table 4** to an 85 dB SPL tone burst at 1 kHz. The sampling rate is 20 kHz.

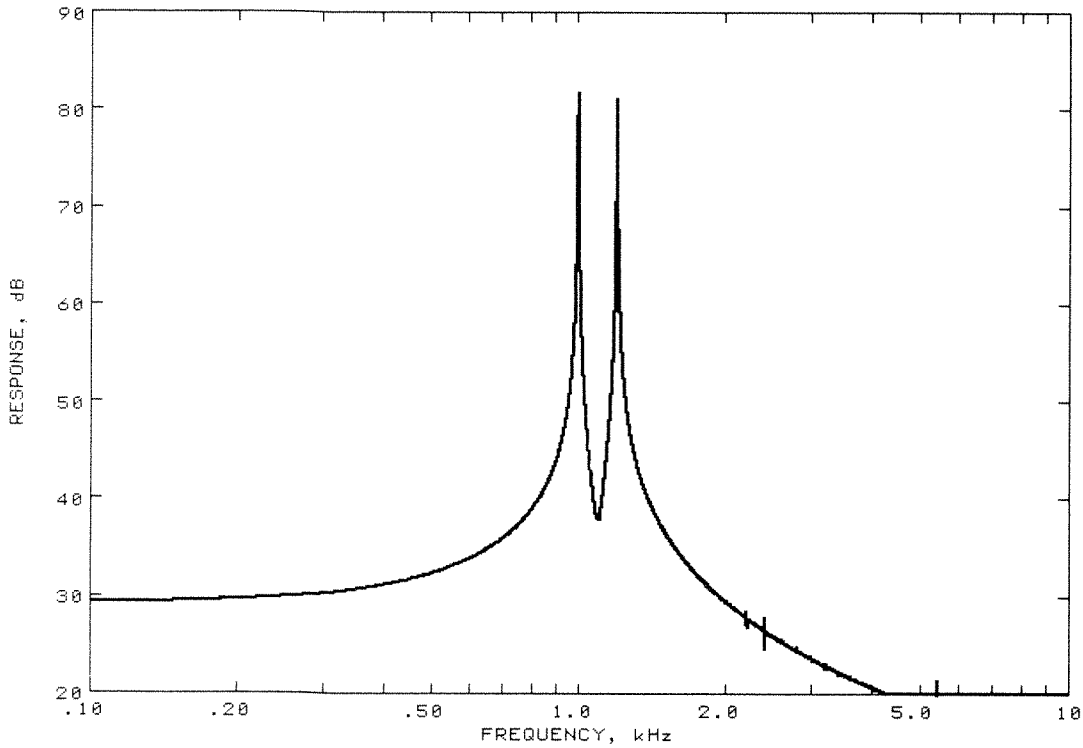

Figure 17.
Response of the simulated linear hearing aid of **Table 2** to a two-tone excitation at 85 dB SPL. The sampling rate is 20 kHz.

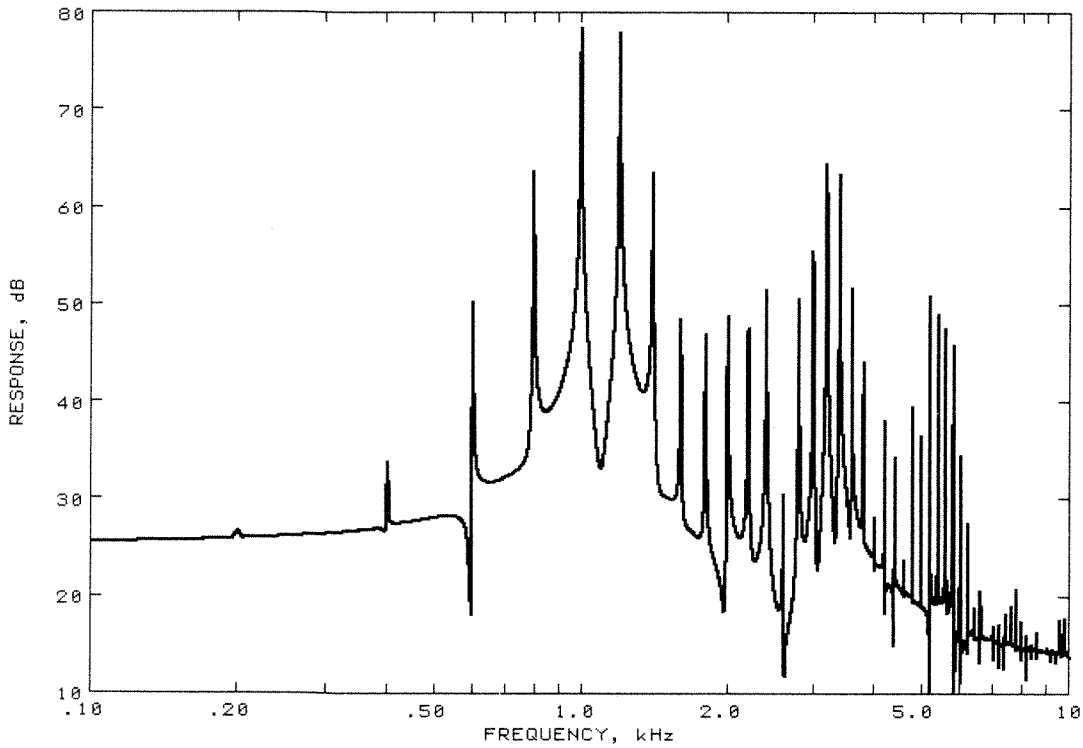

Figure 18.
Response of the simulated hearing aid with clipping, as specified in **Table 3**, to the two-tone excitation at 85 dB SPL. The sampling rate is 20 kHz.

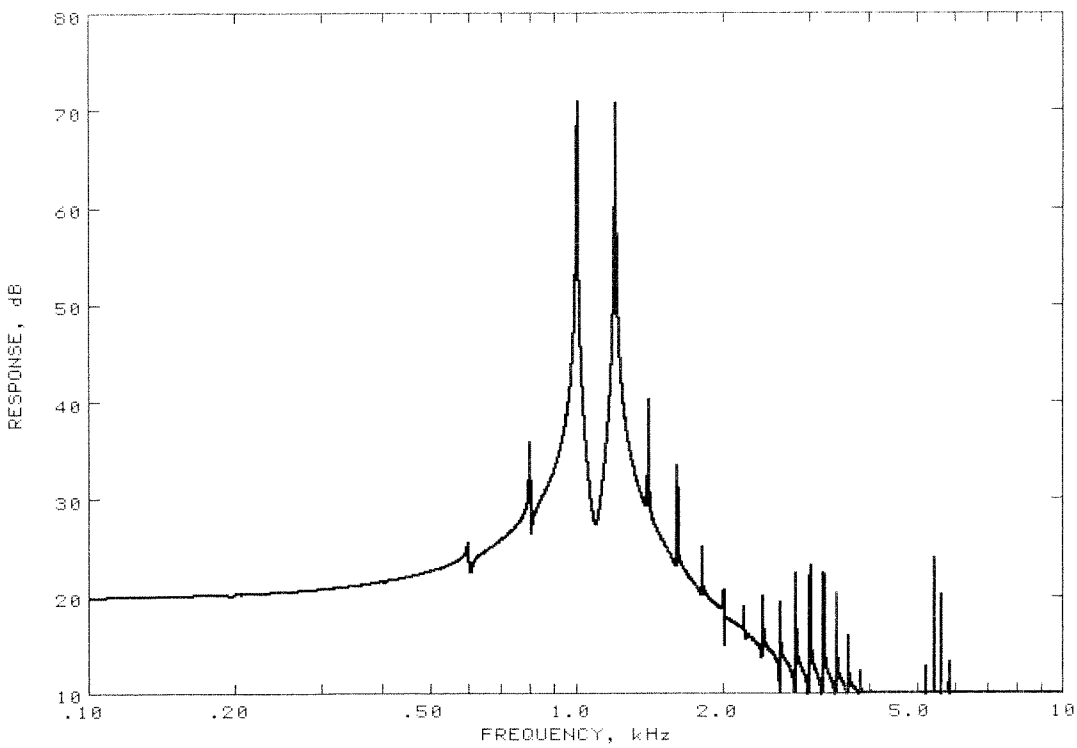

Figure 19.
Response of the simulated two-channel compression hearing aid of **Table 4** to the two-tone excitation at 85 dB SPL. The sampling rate is 20 kHz.

ing, an amplifier with clipping distortion, a receiver, an ear canal and ear drum, and a vent. The model allows for the simulation of acoustic feedback, and goes beyond existing frequency-domain models in also simulating nonlinear effects such as AGC-I and amplifier saturation. The model has been implemented using floating-point arithmetic, so quantization effects in the digital filters are not significant.

Because it is a digital transformation of an analog system, the simulation does not provide an exact reproduction of the analog hearing aid. This is apparent in the slight deviations in portions of the frequency response curves presented for some of the elements of the simulation. Greater accuracy can be obtained by using longer digital filters and a higher sampling rate, but this increases the computational burden without providing a corresponding increase in the amount of information produced. The simulation is thus a representation of a general class of hearing aids rather than a reproduction of a specific instrument.

The purpose of the time-domain simulation is to illustrate aspects of hearing aid behavior and to serve as a computer test bed for the development of new hearing aid processing algorithms and test approaches. The basic simulation can be modified to include more processing channels, different forms of compression, speech enhancement, adaptive filters for feedback suppression, or any other processing scheme to be developed or evaluated. Different portions of the simulation, such as the vent or a specific processing option, can be adjusted or turned on or off at will to give a convenient way of studying the effects of modifying the instrument or the acoustic environment. Because the simulation is equivalent to a self-contained, computer-controlled hearing aid, it can also be used to process stimuli for the development of test signals and processing techniques for the characterization of existing or future hearing aids. The time-domain simulation is therefore a powerful tool for the study of hearing aid behavior.

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