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Acoustic and Hearing Aid Circuit Variables Affecting the Tolerability of Aided Impulsive-Type Sounds

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Abstract

The tolerability of aided impulsive-type stimuli was investigated in a group of 13 hearing-impaired listeners. Two linear circuits (one with a class A and one with a class D output stage) and one adaptive frequency response (AFR) circuit (with a class D output stage) were investigated. In a three-way paired-comparison task, subjects chose the hearing aid that was most tolerable when 75 dB sound pressure level (SPL) impulsive-type sounds were presented. Real-ear measurements of rms SPL, peak SPL, crest factor, and spectral distribution were made to determine which of these variables was most closely associated with behavioral tolerability scores. Results indicated significant differences across hearing aids for tolerability scores, rms sound pressure levels, and spectral peak frequencies. Highest tolerability scores were associated with the hearing aid that produced the lowest rms sound pressure levels in the ear canal (class D AFR). Significant correlations were found between tolerability and both rms SPL and peak SPL. Results are discussed in terms of circuit algorithm and in terms of the possible effects of hearing aid saturation.

Key Words: Hearing aids, impulsive-type stimuli, loudness discomfort

Impulsive-type sounds such as the slamming of a door are often objectionable to hearing aid wearers. If such a sound occurs and causes discomfort, the hearing aid user must decide whether to tolerate the stimulus, turn down the hearing aid, or remove the instrument from the ear. The decision will likely be based on the loudness experienced and on the frequency of occurrence of the objectionable sound. Those individuals who are frequently exposed to loud, impulsive-type sounds may eventually reject the hearing aid, even if the device provides benefit in other acoustic environments.

While it is clear that aided impulsive-type sounds can be found objectionable, the acoustic variables responsible for their objectionable nature are not well understood. Loudness discomfort could occur if high rms sound pressure levels (SPLs) develop in the ear canal upon amplification. It might also occur if peak SPLs or crest factors (the ratio between peak and rms SPL [Frye, 1987]) become high. The perception of impulsive sounds might also vary with the spectral composition of the stimulus, depending on the configuration of the hearing loss and the degree of loudness recruitment involved. The extent to which each of these variables contributes to the intolerable nature of impulsive-type sounds has not been adequately addressed. Preece and Wilson (1988) obtained equal loudness judgments of multitone stimuli differing in crest factor. They reported that stimuli presented at equal rms SPLs were found to be equally loud, regardless of crest factor. Similarly, Bentler et al (1990) measured loudness discomfort levels (LDLs) for stimuli similar to those used by Preece and Wilson and reported that LDLs were similar across stimuli, despite the fact that their stimuli differed greatly to crest factor. The results of these studies suggest that loudness is associated primarily with the rms SPL generated in the ear canal of the listener and is not associated with peak SPL or crest factor.

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The results reported by Preece and Wilson (1988) and by Bentler et al (1990) were based on the performance of normal-hearing listeners who were tested while wearing headphones. These studies were not intended to predict the real-world performance of hearing aids. Similar results, however, might be expected from hearing-impaired listeners exposed to impulsive-type sounds that have been amplified by hearing aids. The SPLs and crest factors generated in the ear canal by a hearing aid will be influenced by a variety of factors, including circuit algorithm, gain, and HFA SSPL90 (ANSI, 1987). A high-gain linear hearing aid with a relatively high HFA SSPL90 value could be expected to generate high peak and high rms SPLs in the ear canal. If an incoming stimulus does not saturate the aid, the crest factor produced in the ear canal might be similar to that of the stimulus unaided. In contrast, a comparable linear hearing aid with a lower HFA SSPL90 value could more easily saturate when driven by an impulsive stimulus. If peak SPL is limited by peak clipping while rms SPL remains high, the net effect of saturation might be to reduce crest factor. A third situation could occur with nonlinear hearing aids, such as those with adaptive frequency response (AFR) circuits. An AFR hearing aid might reduce the rms SPL of a stimulus, relative to that of a comparable linear aid. For AFR hearing aids with relatively high HFA SSPL90 values, crest factor would be determined by the extent to which rms SPL is reduced and on the time required for the circuit to respond to high-amplitude peaks. The relationship between hearing aid characteristics and the degree to which aided impulsive-type sounds are found tolerable to hearing-impaired listeners has considerable clinical importance but has received little attention in the literature.

The following investigation had two primary objectives. First, it was designed to determine which of several acoustic variables, including rms SPL, peak SPL, crest factor, and spectral distribution is most closely associated with the uncomfortable nature of impulse-type sounds. Second, it was intended to show whether tolerability is affected by circuit algorithm when hearing-impaired listeners are exposed to impulsive-type sounds. For the experiment, all acoustic measurements were based on stimuli recorded in the real ears of hearing-impaired subjects, and all behavioral data were based on stimuli heard through analog in-the-ear (ITE) hearing aids.

METHOD

Subjects

Thirteen hearing-impaired subjects participated in the investigation. Subjects were categorized into four audiometric groups, representing gradually sloping (group 1, n = 4), steeply sloping (group 2, n = 3), moderate-flat (group 3, n = 4), and precipitous losses (n = 2). Audiometric thresholds for these subjects are shown in Table 1.

Hearing Aids

Three hearing aids were investigated. Specifications (ANSI, 1987) for each of these hearing aids are shown in Table 2. The first hearing aid (class D AFR) consisted of an adaptive high-pass filter circuit with a class D output stage. This circuit was designed to reduce low-frequency gain as input to the hearing aid increased and was similar in design to the Argosy Manhattan II™. The AFR hearing aid had a HFA SSPL90 of 118 dB SPL. The second hearing aid (class A linear) consisted of a linear circuit with a class A output stage. The HFA SSPL90 for the class A linear aid was 104 dB SPL. The third hearing aid (class D linear) consisted of a linear circuit with a high-powered class D output stage and is known commercially as the Linear Plus P™. This hearing aid had a HFA SSPL90 of 122 dB SPL. HFA full-on gain for the three hearing aids ranged from 39 (class A linear) to 46 (class D linear) dB.

Prior to testing, the gain and frequency response of each hearing aid were adjusted to the mean 2-cc coupler NAL target frequency response associated with the group to which each subject belonged. Two-cc coupler target gain was derived by applying real-ear to 2-cc coupler corrections (Burnett and Beck, 1987) to the real-ear target gain generated by the NAL formula (Byrne and Dillon, 1986). Once the hearing aids were adjusted, potentiometer settings and volume control positions were maintained throughout the course of the experiment.

Maximum output SPL for the class D AFR and the class D linear hearing aids varied with gain and thus were not adequately described by HFA SSPL90 (which is measured with the instrument full on). For this reason, saturation SPLs were obtained in a Zwislocki ear simulator on KEMAR with each of the three hearing aids adjusted as used during the experiment. These

Table 1 Audiometric Thresholds (dB HL) of Test Subjects from Each of Four Subject Groups

	Frequency (Hz)					
Group	*250*	*500*	*1000*	*2000*	*4000*	*8000*
Group 1 (Gradual Slope)						
S1	25	35	45	55	55	60
S2	25	35	45	50	45	60
S3	35	40	45	40	65	75
S4	30	45	50	55	55	60
Group 2 (Steep Slope)						
S5	15	20	45	60	60	85
S6	20	20	40	60	80	80
S7	10	10	25	30	75	65
Group 3 (Flat)						
S8	50	60	55	50	55	65
S9	55	55	40	50	55	60
S10	65	65	75	70	80	80
S11	50	45	55	65	105	105
Group 4 (Precipitous)						
S12	10	5	5	20	85	70
S13	10	10	10	35	70	75

measurements were made with a 100 dB SPL speech-shaped noise input (a level that was verified to ensure saturation), using a probe microphone (Acoustimed) routed to a signal analyzer (Bruel and Kjaer 2032). Saturation SPLs for each of the four subject groups are shown in Table 3, which shows that at use gain settings, overall mean saturation SPLs for the three hearing aids were similar, ranging from 110 dB SPL (class A linear) to 113 dB SPL (class D linear). Thus, while relatively small variations in saturation SPLs, HFA full-on gain, and frequency response existed across hearing aids, circuit algorithm (AFR vs linear) and output stage (class D vs class A) were the primary differences between hearing aids.

Apparatus

A block diagram of the experimental apparatus is shown in Figure 1. Two of the three hearing aids were randomly selected and coupled to two Zwislocki ear simulators that were mounted in the left and right ears of KEMAR. The manikin was positioned in the center of a sound suite (Tracoustics) and faced a loudspeaker (Realistic Minimus 3.5) at a distance of 24 inches. The output of each hearing aid was transduced by a 1/2-inch coupler microphone (Larsen-Davis 2559). Both stimulus channels were routed with a mixer/switching console (Realistic 32-1200B) to an insert earphone (Etymotic Research ER-2) that was placed within one ear of the subject.

Table 2 2-cc Coupler Data* for Hearing Aids

ANSI Specification	*Class D AFR*	*Class A Linear*	*Class D Linear*
Peak SSPL90 (dB SPL)	125	107	127
HFA SSPL90 (dB SPL)	118	104	122
HFA FOG (dB)	45	39	46
THD: 0.5 kHz (%)	1	0	2
THD: 0.8 kHz (%)	1	1	1
THD: 1.6 kHz (%)	0	0	0
F1 (Hz)	200	200	200
F2 (Hz)	6300	6500	6500

*As per ANSI S3.22-1987.

Table 3 Saturation SPLs Obtained in a Zwislocki Ear Simulator (on KEMAR) with a 100 dB SPL Speech-Shaped Noise Input

SSPL (dB)	*Class D AFR*	*Class A Linear*	*Class D Linear*
Overall	111	110	113
Group 1	111	110	111
Group 2	111	110	115
Group 3	112	110	116
Group 4	110	110	110

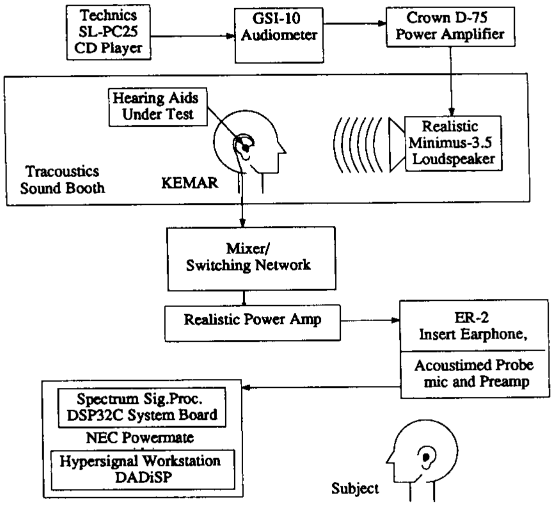

Figure 1 Block diagram of the experimental apparatus used to present test stimuli and to record and analyze real-ear data.

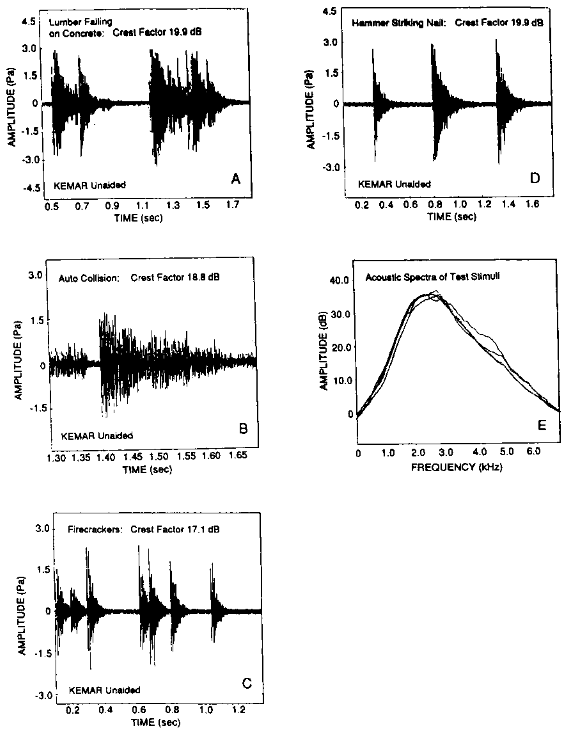

Figure 2 Examples of temporal waveforms of test stimuli, recorded unaided on KEMAR. *A*, Lumber falling on concrete. *B*, Automobile collision at the moment of impact. *C*, Seven firecracker bursts. *D*, Three strikes of a hammer to a nail. *E*, Acoustic spectra of test stimuli.

Stimuli

Four tracks, taken from the "Living Sound Effects" compact disks, (Microsound Products) were used. These consisted of the sounds of lumber falling onto a concrete floor, an automobile collision, exploding firecrackers, and a hammer striking a nail. Examples of the temporal waveforms of each of these stimuli, obtained unaided on KEMAR, are shown in Figure 2 A–D. Unaided crest factors for these stimuli ranged from 17.1 to 19.9 dB. Panel E shows the acoustic spectra of the four test stimuli. These spectra show frequency response peaks of approximately 2500 Hz. The low- and high-frequency slopes of these spectra were approximately −18 dB and −10 dB per octave, respectively, resulting in stimuli with very little energy below 1000 Hz or above 5000 Hz.

Stimuli were presented in the sound field at approximately 75 dB SPL, a level intended to produce saturation in one or more hearing aids without producing loudness discomfort. Input levels were determined in advance of the experiment, using the following method: each test item was presented at a fixed input level (in dB HL, using a Grason-Stadler GSI 10 audiometer) and transduced by the probe microphone, which was suspended at the approximate position of the hearing aid microphone. Each recording was digitized, and a 10,000-point portion of each waveform was extracted for analysis. The extracted segments began at easily identified points of each waveform, corresponding to stimulus onset for a discrete event. The mean amplitude of the squared waveform was then calculated and compared with that of a calibration stimulus (90 dB SPL speech-shaped noise). The rms SPL was expressed in dB by calculating 20 times the logarithm of the square root of the ratio between the test and calibration amplitudes and adding a constant of 90 dB. Stimulus attenuation was then adjusted to ensure that testing levels were as close to 75 dB SPL as possible. During testing, input levels for the four stimuli ranged from 73 to 77 dB SPL.

Procedure

Subjects were seated outside of the test suite, in a relatively quiet (45–55 dBA) environment. Nontest ears were occluded with E-A-R plugs (Cabot). In addition to the insert earphone, the probe microphone was placed within the test ear of the subject. Insertion depth was 20 mm, measured from the tragal notch. Each subject was given the following written instructions:

> Sounds such as the slamming of a door can sometimes be found objectionable to some-

one wearing a hearing aid. The sounds that you are about to hear should not be uncomfortably loud, but you may notice that a sound heard through one hearing aid seems more tolerable (less objectionable) than the same sound heard through another hearing aid. Your task is simply to identify which of two sounds is more tolerable to you, one heard through hearing aid #1 or the same sound heard through hearing aid #2. For each sound listed on the response form, please identify which of the two hearing aids you feel is more tolerable. If *any* sound is uncomfortably loud, please indicate this to me and I will discontinue the experiment.

Each subject listened to one stimulus first through one hearing aid (on KEMAR's left ear) then immediately through the second hearing aid (on KEMAR's right ear). Switching between hearing aids was performed by the investigator. The subject then marked a box on the response form, and the procedure continued until each of the four stimuli had been compared. At this time, one of the hearing aids on KEMAR was replaced with the third hearing aid, and the entire comparison process was repeated. All hearing aids were compared in this fashion (i.e., two at a time, using a tournament strategy similar to that reported by Neuman et al [1987]). During the course of testing, no complaints of loudness discomfort were reported. Upon the completion of testing, each hearing aid was awarded four tolerability scores, one score for each stimulus. Each score represented the number of times a given hearing aid was found more tolerable than the comparison hearing aid. For example, if hearing aid #1 was found more tolerable than either hearing aids #2 or #3, a score of 2 was awarded. A score of 1 was awarded if a hearing aid was more tolerable than one but not both comparison hearing aids, and a score of 0 was awarded if the hearing aid was found less tolerable than both comparison hearing aids.

Acoustic Analysis

During the acquisition of behavioral data, probe microphone recordings were made of all stimuli processed through each of the hearing aids. These recordings were digitized and analyzed for rms SPL, peak SPL, crest factor, and spectral content. The rms SPLs were determined using the method described above. Peak SPLs, based on the highest amplitude sample of each waveform, and crest factors (the difference between peak and rms SPL) were also calculated. Spectral composition was determined by fast Fourier (FFT) analysis. From this analysis, the frequency corresponding to the spectral peak of each waveform was determined.

Statistical Analysis

Four dependent variables of this investigation, tolerability score, rms SPL, peak SPL, and spectral peak frequency, were statistically analyzed using a one-factor (hearing aid circuit) analysis of variance (ANOVA). Significant circuit effects were further examined using Newman-Keuls post-hoc testing. Pearson product-moment correlations between tolerability scores and each of the three acoustic variables were also calculated to determine whether behavioral results could be explained by the acoustic conditions that were present in the ear canals of listeners during testing. Crest factor was not included in these analyses, since crest factors were derived directly from peak and rms SPLs. Group effects were also not statistically analyzed, due to the small number of subjects in each subject group.

Temporal Envelope Analysis

To estimate the extent to which signal processing affected the temporal characteristics of the test items, an envelope difference index (EDI) was devised, using the method outlined in Figure 3: the original waveform (in this example, the automobile collision [panel A], obtained unaided on KEMAR) was first modified by taking its absolute value (panel B). This waveform was then digitally low-pass filtered to produce the temporal envelope (panel C). Since the amplitudes of aided envelopes (obtained on real ears) were generally different from the unaided envelopes (obtained on KEMAR), each data point within a given envelope was divided by the mean amplitude of the entire envelope. This scaled each envelope to a mean amplitude of 1.0 (panel D), which allowed the direct comparison of aided and unaided envelopes. Panel E shows one such comparison, in this case between the aided (class D linear circuit) and unaided envelopes of the automobile collision. The EDI was derived by first subtracting the aided envelope from the unaided envelope and taking the absolute values of the differences. The EDI score was then

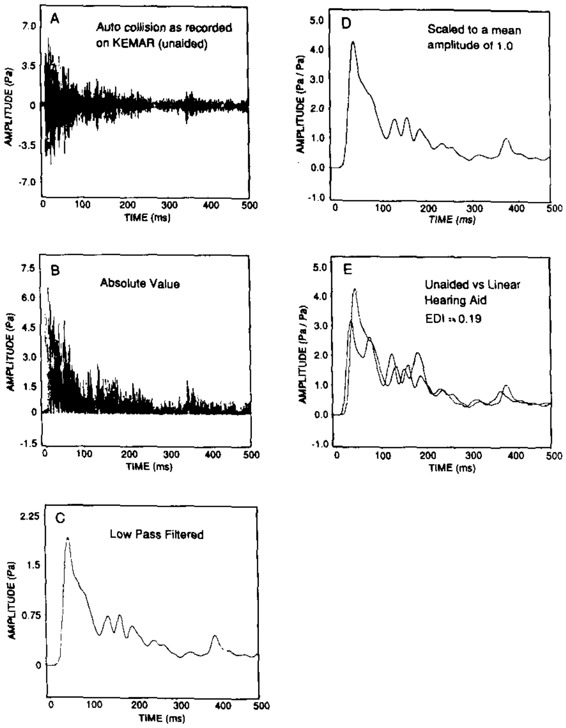

Figure 3 Method of EDI calculation. *A*, Temporal waveform of the automobile collision, as recorded unaided on KEMAR. *B*, Absolute value of the waveform in panel A. *C*, Temporal envelope of the stimulus, produced by low-pass filtering the waveform in panel B. *D*, Temporal envelope of the stimulus, scaled to a mean amplitude of 1.0. *E*, Comparison of the unaided envelope from panel D with an aided envelope, which was obtained by processing an aided waveform in the manner shown in panels B–D. Details regarding EDI calculations are provided in the text.

calculated as the mean of these absolute values and divided by 2, placing the EDI on a scale ranging from 0.00 (perfect correspondence between envelopes) to 1.00 (no correspondence between envelopes). EDI calculations such as these were made for all aided stimuli. Calculations were made for the first 2000 points (100 msec) of each waveform, representing stimulus onset, as well as for complete waveforms.

Envelope Difference Index and Hearing Aid Saturation

To determine whether acoustic or behavioral results could have been influenced by saturation-induced distortion, recordings of each of the four test stimuli were obtained on KEMAR, as processed by a saturated linear hearing aid. Saturation was produced by adjusting the class D linear hearing aid to produce 40 dB of noise gain (ANSI, 1992) and presenting

Table 4 Mean Tolerability Scores, RMS SPL, and Peak SPL Values Across Hearing Aids

	Class D AFR	*Class A Linear*	*Class D Linear*
Tolerability Score			
Overall	1.3 (0.8)	1.0 (0.7)	0.7 (0.9)
Group 1	1.3 (0.8)	1.0 (0.7)	0.8 (0.9)
Group 2	0.9 (0.9)	1.0 (0.9)	1.0 (0.9)
Group 3	1.3 (0.7)	0.9 (0.7)	0.6 (0.9)
Group 4	1.6 (0.7)	1.1 (0.6)	0.3 (0.5)
RMS SPL (dB)			
Overall	97.1 (3.8)	98.9 (4.7)	100.0 (3.5)
Group 1	96.6 (3.5)	99.1 (3.7)	99.6 (4.1)
Group 2	96.7 (3.7)	98.4 (4.7)	99.8 (2.4)
Group 3	97.7 (4.5)	99.7 (3.8)	100.5 (4.5)
Group 4	96.6 (1.6)	97.2 (3.2)	99.9 (4.2)
Peak SPL (dB)			
Overall	113.8 (3.6)	112.4 (1.0)	112.5 (0.7)
Group 1	112.6 (3.7)	113.1 (2.9)	112.2 (3.6)
Group 2	111.2 (4.1)	111.6 (3.0)	113.5 (3.8)
Group 3	115.1 (2.8)	112.7 (2.6)	112.6 (3.3)
Group 4	114.0 (3.6)	111.3 (3.8)	111.6 (4.1)

Standard deviations shown in parentheses.

each test stimulus at 95 dB SPL. These recordings were made to determine the EDI value associated with saturation. EDI calculations were made for the first 2000 points of each waveform, as well as for complete waveforms. By comparing real-ear EDIs with the EDI associated with saturation, it was possible to determine whether hearing aids were in saturation during testing.

RESULTS

Mean tolerability scores, obtained across and within groups, are shown in the upper panel of Table 4. Highest overall tolerability scores were associated with the class D AFR hearing aid (mean score of 1.3), followed by the class A linear (mean score of 1.0) and class D linear (mean score of 0.7) instruments. Subjects from three out of four subject groups rated the AFR hearing aid higher in tolerability than either of the two linear instruments. Subjects from three out of four groups also rated the class A linear hearing aid higher in tolerability than the class D linear hearing aid. Group 2 subjects rated the class D AFR, class A linear, and class D linear hearing aids with very similar tolerability scores of 0.9, 1.0, and 1.0, respectively.

Mean real-ear rms SPLs generated by each of the three hearing aids are shown in the

Table 5 Mean Crest Factors and Spectral Peak Frequencies across Hearing Aids

	KEMAR		*Class D AFR*		*Class A Linear*		*Class D Linear*	
Crest Factor (dB)								
Overall	18.9	(1.3)	16.7	(2.3)	13.5	(1.7)	12.5	(1.6)
Group 1	—		16.0	(2.5)	14.0	(1.4)	12.6	(1.7)
Group 2	—		14.5	(1.3)	13.2	(2.6)	13.7	(1.4)
Group 3	—		17.4	(2.4)	13.0	(1.4)	12.1	(1.5)
Group 4	—		17.4	(0.5)	14.1	(1.2)	11.7	(0.8)
Spectral Peak (Hz)								
Overall	2720	(245)	2930	(290)	2732	(685)	2440	(591)
Group 1	—		2920	(198)	2059	(667)	2264	(337)
Group 2	—		3787	(663)	3635	(532)	3398	(874)
Group 3	—		2748	(153)	1857	(541)	2170	(259)
Group 4	—		3071	(220)	3742	(438)	2730	(309)

Standard deviations shown in parentheses.

middle panel of Table 4. Overall, the class D AFR hearing aid generated the lowest real-ear rms levels (97.1 dB SPL), followed by the class A linear (98.9 dB) and class D linear (100.0 dB) instruments. This pattern of results was found for all subject groups; rms SPLs associated with the AFR hearing aid were for all groups lower than those associated with either linear instrument, and rms SPLs associated with the class A linear hearing aid were for all groups lower than those associated with class D linear instrument. For individual hearing aids, rms levels were similar across groups, despite the fact that each group was associated with a unique NAL target prescription.

Mean real-ear peak SPLs are shown in the lower panel of Table 4. Overall, peak levels were similar across hearing aids, ranging from 113.8 dB SPL (class D AFR) to 112.4 dB SPL (class A linear). Mean peak SPLs were similar to the corresponding saturation SPLs shown in Table 3, suggesting that saturation may have occurred at the moment of peak SPL. For individual hearing aids, mean peak SPLs varied by no more than 4 dB across groups.

Mean real-ear crest factors are shown in the upper panel of Table 5. Mean crest factors obtained unaided on KEMAR are also shown. KEMAR data represent a mean based on a single recording of each of the four test stimuli. Each of the three hearing aids produced mean crest factors that were lower than those obtained on KEMAR. The class D AFR hearing aid produced the highest mean real-ear crest factor (16.7 dB), followed by the class A linear (13.5 dB) and the class D linear (12.5 dB) hearing aids. The relatively high crest factor associated with the AFR hearing aid reflects relatively high peak levels, combined with relatively low rms levels. In contrast, the class D linear hearing aid produced peak and rms levels that were *both* fairly high, yielding relatively low crest factors. The mean crest factor associated with the class A hearing aid fell between those of the two class D hearing aids, as was also true of the mean rms SPL associated with this hearing aid. This pattern of results was repeated for all subject groups except group 2. Crest factors for group 2 fell within a very narrow range, but were highest for class D AFR hearing aid (14.5 dB), followed by the class D linear (13.7 dB) and class A linear (13.2 dB) instruments.

Spectral analysis results are shown in the lower panel of Table 5. The AFR hearing aid produced the highest overall mean peak frequency (2930 Hz), followed by the class A linear (2732 Hz) and class D linear (2440 Hz) instruments. The mean spectral peak frequency obtained unaided on KEMAR was 2720 Hz. Group data revealed that spectral peak frequencies were higher for groups 2 and 4 than for groups 1 and 3. Subjects in groups 2 and 4 had audiometric configurations that generated NAL targets requiring minimal low frequency gain. Thus, the results most likely reflect frequency response adjustments made prior to testing, rather than circuit algorithm.

A summary of statistical results is shown in Table 6. Analysis of variance revealed significant differences between hearing aids for tolerability scores ($p = .002$), rms SPLs ($p = .015$), and spectral peak frequencies ($p = .005$). Peak SPLs did not statistically differ across hearing aids. Post-hoc analysis of significant main effects

Table 6 Summary of Statistical Analysis

ANOVA	*SS*	*MS*	*MS (within)*	*F (2,153)*	*P*
Tolerability Score	8.58	4.29	0.63	6.78	.002
RMS SPL	161.96	80.98	18.83	4.30	.015
Peak SPL	67.64	33.82	21.66	1.56	.211
Spectral Peak	$8.4^{*} 10^{6}$	$4.2^{*} 10^{6}$	$7.5^{*} 10^{5}$	5.59	.005

Correlations with Tolerability	*R (x,y)*	*P*
RMS SPL	-.45	.000
Peak SPL	-.24	.002
Spectral Peak	.12	.103

revealed the AFR hearing aid to have significantly higher tolerability scores ($p < .05$) and significantly lower rms levels ($p < .05$) than the class D linear instrument. No significant differences were found between the class D AFR and class A linear hearing aids for either tolerability scores or rms levels. Post-hoc analysis of spectral peak frequencies revealed that the AFR hearing aid generated significantly higher spectral peak frequencies than either of the two linear instruments. The lower portion of the table shows that significant negative correlations with tolerability scores were found for both rms SPL ($p = .000$), and peak SPL ($p = .002$). These correlations indicate that higher tolerability scores were associated with lower rms and peak SPLs. The correlation between tolerability and spectral peak frequency was nonsignificant ($p = .103$).

Results of the temporal envelope analysis are summarized in Figure 4. Columns within the figure represent the mean EDIs obtained across groups for each of the three hearing aids. Results of the 2000-point (100 msec) and 10,000-point (500 msec) calculations are shown in panels A and B, respectively. The horizontal line within each panel represents the EDI associated with saturation for the analysis interval shown. EDIs for the 2000-point analysis were very similar across hearing aids, ranging from 0.16 (class D linear) to 0.19 (class D AFR). These values exceeded those associated with a saturated linear hearing aid (0.15), suggesting that each of the three hearing aids were, on the average, saturated during stimulus onset. This condition was not unexpected. When calculated over the entire waveform, mean EDIs were also very similar across hearing aids, ranging from 0.29 (class D linear) to 0.31 (class D AFR). These values, while higher than those derived from the 2000-point analysis, were lower than that associated with a saturated linear hearing aid (0.35). EDIs were higher for the 10,000-point analysis because the mean amplitude of 10,000-point waveforms was always lower than the mean amplitude of waveforms calculated over the first 2000 points, due to the natural decrease in stimulus amplitude over time. To calculate the EDI, 10,000-point waveforms were therefore scaled by values lower than those required for 2000-point waveforms, in order to achieve envelopes with a mean amplitude of 1.0 (see Method). Thus, the extent of temporal alteration is better reflected by the relationship between the real-ear EDI and the saturated EDI examined within, rather than across, a particular analysis interval. The data in Figure 4 suggest that as the natural temporal envelope of the unaided stimulus decayed in amplitude over time (i.e., beyond the initial temporal peak), each hearing aid processed the signal without saturating. This pattern of results occurred regardless of circuit algorithm. Thus, these data suggest that saturation occurred for all hearing aids, but only during the first few tenths of a second of each stimulus.

DISCUSSION

The results of this investigation suggest that the objectionable nature of impulsive-type sounds is associated with both the rms and the peak SPL present in the ear canal of the hearing aid wearer. In this respect, impulse noise appears to be no different than other types of high-level stimuli. These results are consistent with those reported by Bentler et al (1990) and by Preece and Wilson (1988), who tested normal-hearing listeners under headphones. The results further suggest the straightforward conclusion that hearing aids that prevent the occurrence of high SPLs in the ear

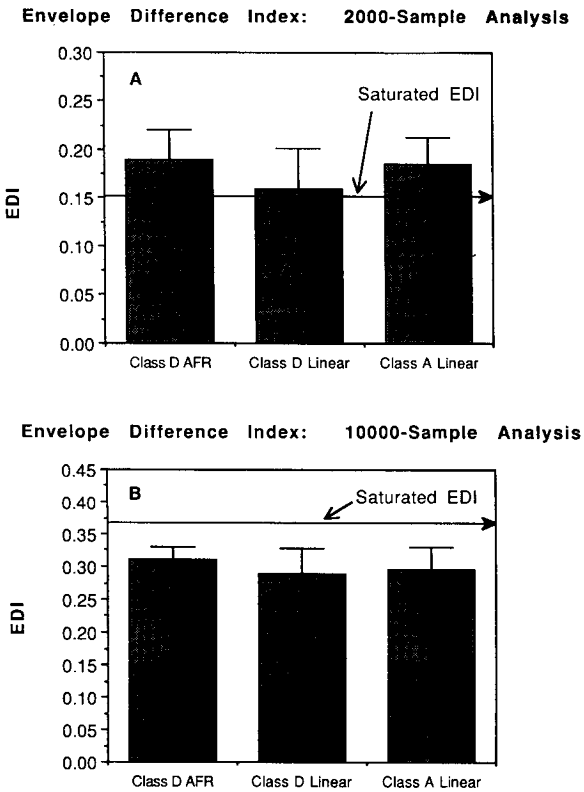

Figure 4 Mean envelope difference index values obtained across hearing aids. Panels A and B represent 2000-sample (beginning with stimulus onset) and 10,000-sample analysis intervals, respectively. The horizontal line within each panel represents the EDI associated with a saturated linear hearing aid. Error bars represent one standard deviation.

canal will be found more tolerable than those that do not. In this study, the AFR hearing aid was found the most tolerable overall, followed by the class A linear and class D linear hearing aids. These results followed the rms SPLs associated with these instruments. While a significant correlation between tolerability and peak SPL was also found, differences in peak SPL across hearing aids did not reach statistical significance in this investigation.

The results presented in Tables 4 and 5 may in part be explained by saturation and by differences in circuit algorithm. At use gain settings, the saturation SPLs for the three hearing aids were similar to one another and relatively low (Table 3). When exposed to 75 dB SPL impulsive-type sounds with natural crest factors approaching 20 dB, the data suggest that each hearing aid was briefly saturated during stimulus onset. Saturation was suggested by the fact that (1) EDIs based on stimulus onset exceeded that associated with a saturated linear hearing aid; (2) peak SPLs measured across instruments reached the saturation SPLs; and (3) aided crest factors were lower than those obtained unaided. Beyond stimulus onset, however, it is likely that the high-pass filtering associated with the AFR algorithm reduced the rms SPL present in the ear canal, relative to the two linear instruments. This would account for the higher tolerability scores associated with this circuit. Differences in rms SPL across circuits were, however, small, probably due to the lack of low-frequency energy present in the test stimuli. The effects of high-pass filtering might have been greater had stimuli with more low-frequency energy been used.

One conclusion that could be drawn from these data would be that reducing rms (or peak) SPL either by adaptive high-pass filtering or by output limiting via peak clipping would be acceptable for preventing loudness discomfort for impulsive-type sounds. While this conclusion seems intuitively reasonable, other factors need to be considered. One such variable is the extent to which a hearing aid will saturate in response to a variety of naturally occurring sounds, including speech. Previous research (Fortune and Preves, 1992) has shown that linear hearing aids with low HFA SSPL90 values are associated with lower aided LDLs and lower impressions of sound quality for speech than comparable hearing aids with higher HFA SSPL90 values. Thus, reducing SPL with an output limiting potentiometer will not likely prevent loudness discomfort or improve sound quality. On the other hand, if AFR or perhaps other nonlinear algorithms successfully prevent high SPLs from developing in the ear canal *without saturating*, they may offer advantages over linear aids with low HFA SSPL90 values. In this study, EDI analysis showed little evidence of saturation beyond the first few tenths of a second of each stimulus for any of the three hearing aids. Despite expectations, the moderate 75 dB input level, combined with low prescriptive gain requirements, rarely produced saturation beyond stimulus onset. Higher input levels or gain were avoided, in order to prevent loudness discomfort during testing. Thus, these data do not adequately address the effects of saturation on tolerability; they do, however, indicate the significant effects that rms and peak SPL have on the tolerability of aided impulsive-type sounds.

This investigation represents an initial examination of the complex effects that hearing aids may have on the tolerability of impulsive-type sounds. Effects that have been reported here could vary with other circuit types, different circuit characteristics, or alternative prescriptive fitting formulas. Future investigations should systematically examine these variables and should continue to use real hearing aids worn by hearing-impaired listeners. By systematically pursuing these effects, it should become possible to reduce or eliminate the objectionable nature of aided impulsive-type sounds for the majority of hearing-impaired listeners.

REFERENCES

American National Standards Institute. (1987). *Specification of Hearing Aid Characteristics*. (ANSI S3.22-1987). New York: ANSI.

American National Standards Institute. (1992). *Testing Hearing Aids with a Broad-band Noise Signal*. (ANSI S3.42-1992). New York: ANSI.

Bentler R, Pavlovic C, Abbas P. (1990). A signal processing scheme for output limitation. *Acta Otolaryngol* Suppl 469:230–235.

Burnett ED, Beck LB. (1987). A correction for converting 2 cm^3 coupler responses to insertion responses for custom in-the-ear nondirectional hearing aids. *Ear Hear* 8(5): 89S–94S.

Byrne D, Dillon H. (1986). The national acoustic laboratories (NAL) new procedure for selecting the gain and frequency response of a hearing aid. *Ear Hear* 7(4):257–265.

Fortune T, Preves D. (1992). Hearing aid saturation and aided loudness discomfort. *J Speech Hear Res* 35(1):175–185.

Frye G. (1987). Crest factor and composite signals for hearing aid testing. *Hear J* 40(10):15–18.

Neuman C, Levitt H, Mills R, Schwander T. (1987). An evaluation of three adaptive hearing aid selection strategies. *J Acoust Soc Am* 82(6):1967–1976.

Preece J, Wilson R. (1988). Detection, loudness and discrimination of five-component tonal complexes differing in crest factor. *J Acoust Soc Am* 84:166–171.