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A new portable digital hearing aid referred to as CLAIDHA (Compensate for Loudness by Analyzing Input-signal Digital Hearing Aid), which employs frequency-dependent amplitude compression based on narrow-band loudness compensation, was clinically evaluated in 159 subjects with hearing loss. The results of speech tests revealed better intelligibility compared with the subject's own hearing aids; the advantage of using CLAIDHA in daily life was also indicated by the results of a questionnaire completed by the subjects. In about 64% of the subjects with a flat, gradually sloping type of hearing loss, CLAIDHA was satisfactorily adopted for daily use. However, in the subjects with a steeply sloping type of hearing loss and subjects with losses mainly at high and low frequencies, with near-normal mid-frequency hearing, this loudness compensation scheme seems to be slightly less effective.

Key words: CLAIDHA, digital hearing aid, loudness compensation, sensorineural hearing loss.

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Introduction

Loudness function in ears with sensorineural hearing loss is typically characterized by 'loudness recruitment'. With loudness recruitment, the dynamic range between the hearing threshold and the level at which sounds become uncomfortably loud (uncomfortable loudness level, UCL) becomes narrower, and the perceived loudness of sounds often increases more rapidly than normal with increasing sound level above the absolute threshold. One possible way to compensate for loudness recruitment is to use a hearing aid which compresses the dynamic range of the input sounds (Villchur, 1973; Yund et al., 1987; Farassopoulos et al., 1989; Kollmeier, 1991; Moore, 1989; Moore et al., 1992).

Loudness functions, which describe the growth of loudness as a function of sound pressure level, as well as hearing thresholds, usually change as a function of frequency and differ among hearing-impaired people. For example, with high-frequency hearing loss caused by presbycusis, the loudness functions may be nearly normal at low frequencies while those at high frequencies may be far steeper. Linear amplification with frequency response shaping will not result in appropriate relative loudness of different frequency components for all input sound levels. Therefore, frequency-dependent amplitude compression based on compensation for the change of the loudness functions in each narrow frequency band has been proposed (Villchur, 1973; Kollmeier, 1991; Asano et al., 1991a, b, c; Suzuki & Sone, 1993; Kiessling et al., 1996). Yund et al. (1987) reported that an 8-channel compression hearing aid was effective in some sensorineurally impaired listeners. However, if the input speech signals are processed with many band pass filters having narrow band-widths and the gain of each band is controlled independently of other bands, severe spectral distortion or spectral flattening can occur and the output may be less beneficial than expected (Lippmann et al., 1981; Bustamante & Braida, 1987; Plomp, 1988). On the other hand, if the input signal is processed with only one or two bandpass filters, the system cannot deal effectively with large threshold changes of subjects with strongly frequency-dependent hearing impairment. To avoid this problem, Moore & Glasberg proposed making the widths of the bands broader than those of the 'auditory filters' of the impaired listener (Moore & Glasberg, 1986; Moore, 1989; Moore et al., 1992).

Our idea, on the other hand, is to divide the signal processing into frequency analysis and filtering for the spectral shaping of the signal. This enables the system to determine the gain at a certain frequency, taking not only the frequency component at that frequency but also the neighbouring frequency components into account, resulting in much less spectral flattening. Moreover, since a high order frequency-sampling-type digital filter is applied to signal processing, it can follow large threshold changes as a function of frequency (Asano et al., 1991a, b, c). For spectral shaping, the loudness compensation principle rather than the simple compression scheme is applied, i.e. the gain-frequency characteristic is temporarily changed so that the loudness perceived by the impaired listener in each narrow band becomes equal to that perceived by normal listeners for any moment and for any frequency band (Suzuki & Sone, 1993). Due to this principle, the loudness balance among frequencies can be restored. This is achieved by having the gain-frequency characteristic of a digital filter determined by 'loudness compensation functions (LCFs)', which describe the relation between the loudness perceived by a specific impaired listener and that perceived by (average) normal listeners.

We call the above algorithm CLAIDHA (Compensate for Loudness by Analyzing the Input-signal, Digital Hearing Aid). Several prototypes of real-time digital hearing aid systems realizing the algorithm with a single DSP (Digital Signal Processor) chip have been produced (Sone et al., 1995). In this clinical study, the outcome of the algorithm was evaluated by applying a fourth system, CLAIDHA IV, which was approved for sale by the Ministry of Health and Welfare of Japan in 1995 under the name of Cleartone® (Ono Sokki Ltd., Japan).

Material and Methods

1. Algorithm and System of the Aid

A block diagram of the CLAIDHA algorithm is shown in Fig. 1. Detailed explanations of the algorithm and system have been presented previously (Asano et al. 1991a, b, c; Suzuki & Sone, 1993; Sone et al., 1995). In brief, the input signal is divided into two paths. In one path, the signal is time-windowed into short-time blocks (128 samples, i.e., 8 ms at a sampling frequency of 16 kHz) and the frequency spectrum is calculated by 32-point FFT (Fast Fourier Transform). The frequency spectra are then averaged over four time-blocks. From the average frequency spectrum, band powers of five one-octave bands (centered at 250, 500, 1000, 2000 and 4000 Hz) are then calculated. The optimum gains at the centre frequencies of the one-octave bands for that time period are determined by referring to the LCFs, which are derived from loudness functions individually measured for a specific impaired listener and those for average normal listeners. Then the gains are smoothly interpolated at 24 equally spaced frequencies (linear scale).

In the other path, the input signal within the block is processed with the gains, which are realized by a 48th-order frequency-sampling digital filter, which is equivalent to a 24-channel bandpass filter bank. Filter coefficients are calculated for every block. However, if the coefficients are updated immediately at the block boundary, the processed signal becomes discontinuous and serious distortion can be heard. Therefore, the coefficients are gradually updated. When the gain should increase, compared with that of the previous block, the change is

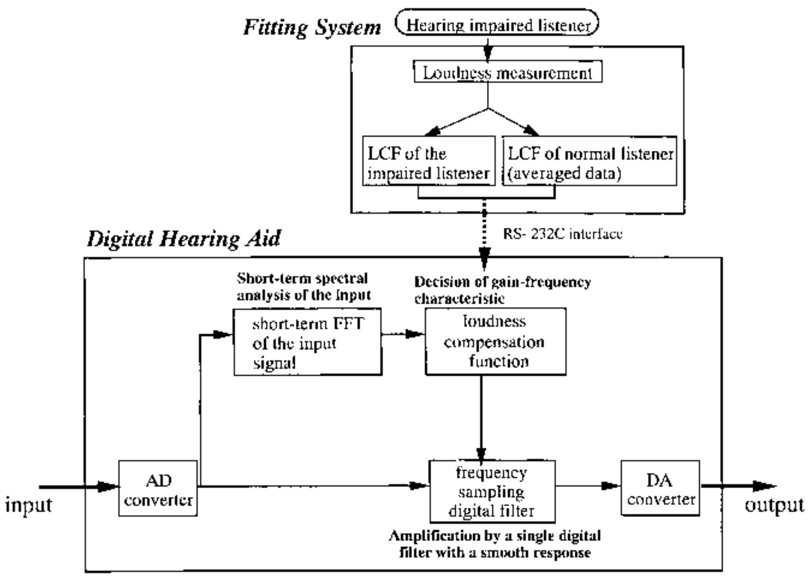

Fig. 1. Block diagram of the CLAIDHA and its fitting system. See text for details.

relatively slow (equivalent to a 10.4 ms release time). On the other hand, when the gain should decrease, the coefficients are changed rapidly (equivalent to a 1.24 ms attack time) to prevent the output signal from becoming too loud.

The fitting system consists of a host personal computer and hardware equipped with 16-bit A/D and D/A converters, a digital signal processor (Motorola DSP-96000, USA) and a dynamic-type earphone (HES100, Hoshiden, Japan). From the fitting system, the data of the calculated LCFs are transferred to the CLAIDHA hearing aid system using an RS-232C interface (Fig. 1). The portable digital hearing aid system ‘CLAIDHA IV’, 70 × 130 × 24 mm in size, uses a Motorola DSP-56166 with a 128-kbyte ROM.

2. Subjects and the Selection of Fitting Side

One-hundred-and-fifty-nine subjects (118 males and 41 females, with a mean age of 73.6 years) who were fitted with the CLAIDHA in our clinic were evaluated in the present study. Most of them suffered from presbycusis, and had complaints about the hearing aids which they had been using (‘Own Aid’). Pure-tone audiometry, tympanometry and speech audiometry were conducted after otoscopic examinations. All of the subjects were fitted monaurally. The fitted side was usually the one with the better hearing threshold. When the thresholds were symmetrical, the CLAIDHA was fitted to the ear with better speech intelligibility, as shown by speech audiometry with Japanese nonsense monosyllables. There were two exceptions, however, in which the fitting was conducted in the ear with the worse hearing threshold: for one subject, the contralateral ear had relatively better hearing (average hearing level of 33 dB HL), and thus the subject requested that the ear with worse hearing be fitted; in another subject, the worse ear showed a flat type of loss (an average hearing level of 54 dB HL), whereas the contralateral side had a low frequency hearing loss (an average hearing level of 45 dB HL).

One-hundred-and-thirty-four of the 159 subjects had used hearing aids before being fitted with the CLAIDHA. In 128 of these subjects, the CLAIDHA was fitted in the same ear as that in which their own aids were being used. At the beginning of this trial, 108 subjects were using their own aids in their daily lives while the other 20 subjects did not use their aids at all. Six subjects were fitted with the CLAIDHA in the ear contralateral to that in which their own aid was being used. In four of these subjects, this was done because they had used their own aid in the ear with a hearing threshold at least 5 dB worse than that on the contralateral side. In another subject, this was done because the ear had a flat type of loss (average hearing level of 54 dB HL), whereas the contralateral side had a low frequency hearing loss (average hearing level of 45 dB HL). In the remaining subject, the contralateral ear was fitted at her strong request; her audiogram was almost symmetrical. Twenty-five subjects had never used a hearing aid before the application of CLAIDHA.

The previous hearing aids and CLAIDHA were evaluated electroacoustically in some of the subjects. Standard ear simulator measurements were conducted at ‘use-gain’ levels with the LH-11 (RION Co. Ltd., Japan).

3. Measurements of Loudness Function

Since LCFs were calculated based on the relationship between average loudness functions for normal listeners and that of the impaired subject, measurements of loudness functions were important in the procedure to fit the aids effectively. A category subdivision scaling method (Heller, 1985; Hellbrück 1991), which has been shown to have smaller variances than the conventional category-scaling method (Suzuki et al., 1995), was used to assess the loudness functions in the present study (see Fig. 2). The above-mentioned fitting system was used to determine the loudness functions. Prior to the measurement of loudness, the threshold and the UCL of each subject was estimated with 1/3-octave band noises at the centre frequency of each 1/1-octave band from 250 Hz to 4 kHz through the earphone, which was calibrated with an ear simulator (B&K 4157, Denmark) and sound level meter (B&K Type 2610, Denmark). The levels of the noise bands were selected to fall above the threshold but below the UCL. The noise bands were presented through the earphone, with both frequency and level being randomized. The noise had a steady-state duration of 1 sec and linear rise and fall times of 50 ms. In the first step, subjects were requested to select one of seven rough categories of loudness shown on the monitor: ‘inaudible’, ‘very soft’, ‘soft’, ‘medium’, ‘loud’, ‘very loud’, ‘too loud’. Then, except in cases when ‘too loud’ or ‘inaudible’ was selected, the same noise band was presented to the subjects again. In the second step, the selected category (‘soft’ in the example shown in Fig. 2) was magnified, and the listeners were asked to select 1 of 10 fine subcategories subdividing each rough (first step) category. These find subcategories corresponding to a series of numbers from 0 to 50.

In one session, each frequency and level combination was presented three times in almost all of the cases, the median fragment of which was used to construct functions relating perceived loudness to sound pressure level at each centre frequency.

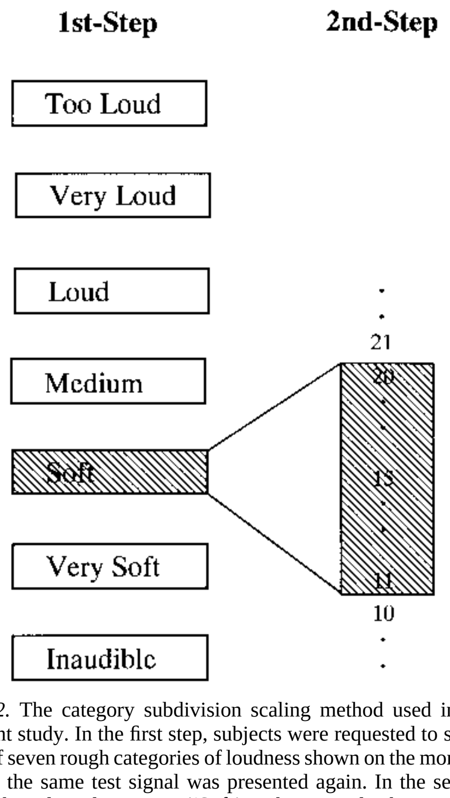

Fig. 2. The category subdivision scaling method used in the present study. In the first step, subjects were requested to select one of seven rough categories of loudness shown on the monitor. Then, the same test signal was presented again. In the second step, the selected category (‘Soft’ in the example shown in the Figure) was magnified, and the listeners selected one of ten fine categories subdividing each rough (the first step) category.

4. Fitting of the Aids

By comparing the measured loudness functions of the subjects with those of the normal listeners (average of 15 listeners in their twenties) at the five frequencies, LCFs were calculated so that the loudness perceived by the impaired listener for each one-octave band would be equal to that of the normal listener (Fig. 3A). Based on the *calculated* LCFs, the *basic* LCFs (i.e., LCFs actually used in the initial phase of the fitting procedure) were determined on the dB-dB plane by linear curve-fitting between 30 and 80 dB SPL input levels (Fig. 3B). After determining the LCFs for five frequencies (250, 500, 1000, 2000 and 4000 Hz), the gains were smoothly interpolated by using a spline function at equally spaced frequencies.

The LCFs were then transferred into the CLAIDHA (Fig. 1). Based on interviews with the subjects, as well as speech intelligibility results, if necessary the basic LCFs which were transferred to the CLAIDHA were modified (*modified* LCFs). For example, if the subject complained of low-level background noise, gains at low frequencies (250 and 500 Hz) were decreased (54 subjects) or compression thresholds at all frequencies were changed from 30 dB SPL to 40 or 50 dB SPL (for 32 subjects). Gains at higher frequencies (2000, 4000 Hz) were increased (four subjects) or decreased (six subjects), when speech sounded ‘unclear’ or too ‘shrill’, respectively. Some of the patients complained of total loudness of the output; five subjects preferred the LCFs which were linearly decreased more than 10 dB from the basic LCFs, and nine subjects preferred those which were linearly increased more than 10 dB from the basic LCFs.

5. Measurements of Speech Intelligibility

Speech discrimination tests were conducted to evaluate the benefits of CLAIDHA after initial fitting: some were tested a few days after the initial fitting. Each trial of the test employed a ‘67-S’ Japanese monosyllabic identification test (developed by the Japan Audiological Society), consisting of 20 randomized syllables. The speech signal was generated through a 16-bit digital-to-analog converter system, low-pass filtered at 21.7 kHz, attenuated (CLAIDHA fitting system, Ono-sokki, Japan), and delivered to the subjects by a loudspeaker (RAMSA ws-A80, Technics, Japan) in a dead sound-attenuating room. The subject was seated 2 m from and directly facing the loudspeaker with the external meatus of the contralateral ear occluded by an earmold. Sound levels were calibrated at the position corresponding to the centre of the subject’s head.

In a given test session, a subject was tested in three conditions: (a) with CLAIDHA, (b) with Own Aid and (c) Unaided. In cases when Own Aid was not used in the ear fitted with CLAIDHA, only conditions CLAIDHA and Unaided, were tested. In principle, the condition CLAIDHA was tested first, so as to avoid any advantage for the CLAIDHA condition due to learning of the speech signals. The volume controls of the subjects’ own aids were set at the volumes normally employed.

In the aided conditions, speech signals were usually presented starting at a level of 40 dB SPL, and the level was increased up to 80 dB SPL in 10 dB steps. In most tests of the Unaided condition, the signals were presented at levels from 50 to 90 dB SPL in 10 dB steps.

In 67 of all subjects, speech intelligibility tests were conducted only after the initial fitting because they could not return to our clinic or were unwilling to continue the evaluation. For the remaining 92 subjects, conditions CLAIDHA were tested again together with evaluation by questionnaires.

6. Questionnaire

The effectiveness of the aids was also evaluated with a questionnaire about experiences in everyday life (Walden et al., 1984). Each subject was asked to give a rating to each statement given below by circling a number between −3 and 3 (rating scale of 7). In question 2, circling ‘−3’, ‘−2’, ‘−1’, ‘0’, ‘1’, ‘2’ and ‘3’ meant ‘much worse’, ‘worse’, ‘slightly worse’, ‘same’, ‘slightly better’, ‘better’, and ‘much better’, respectively (a different scale from Own Aid). In the other questions, circling ‘−3’, ‘−2’, ‘−1’, ‘0’, ‘1’, ‘2’ and ‘3’ meant ‘very bad’, ‘bad’, ‘a

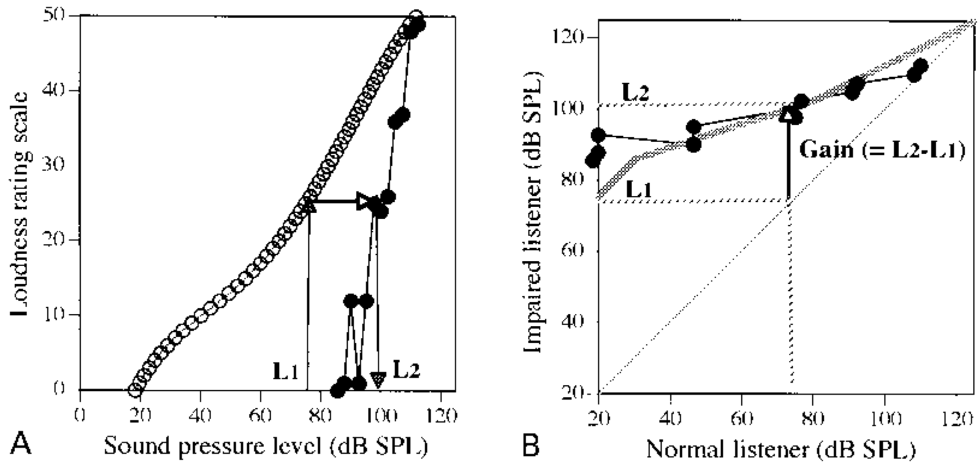

Fig. 3. An example of loudness function (left panel) and loudness compensation function (LCF) (right panel). In the left panel, loudness scale was plotted as a function of sound level. For the impaired listener, the level of L2 is needed for the subject to perceive the same loudness as perceived by a normal listener with level L1. In the right panel, the solid line with filled circles shows the *calculated* LCF and the dashed line the *basic* LCF actually used. To avoid overamplification of faint background noise, the input–output functions were linearly decreased below an input level of 30 dB SPL. Those above 80 dB SPL were determined as follows: the points at the input level of 80 dB SPL were linked to the points which indicated discomfort levels for both normal listener and impaired listener.

little bad', 'fair', 'a little good', 'good', and 'excellent', respectively (an absolute scale).

1. Overall evaluation.
2. Comparison with own aid if any (only the questionnaire for CLAIDHA).
3. Effectiveness in quiet situations. (a) Dialogue with one person, (b) watching TV alone, (c) group conversation.
4. Effectiveness in noisy situations. (a) Dialogue with one person, (b) watching TV alone, (c) group conversation, (d) conversation in street noise.
5. Noisiness/loudness of sounds. (a) Impulsive sound from china/glass being tapped or from sounds of footsteps. (b) Loudness and sound quality of own voice. (c) Total loudness of the sounds.
6. Ease of operation of the hearing aid.

The evaluation for Own Aid was usually done before the ear was fitted with the CLAIDHA: that for CLAIDHA was done from 1 to 20 weeks (a mean period of 26 days) after the initial fitting. The test period varied because subjects came to the clinic from various parts of Japan and it was difficult to keep the test period constant.

Results

1. Speech Intelligibility

The percent correct on the monosyllable speech tests with hearing aids at each of four input stimulus levels (50, 60, 70 and 80 dB SPL) is plotted as a function of scores for the unaided condition (Fig. 4, left column). At 50 and 60 dB SPL, speech intelligibility appears to be generally better in the aided conditions than in the unaided conditions, but not at 80 dB SPL. To compare the effect of CLAIDHA with that of Own Aid, the results for the CLAIDHA were plotted as a function of those for the Own Aid (Fig. 4, right column). Scores for CLAIDHA tended to be larger than those for Own Aid at lower input levels, but the differences became smaller at higher input levels.

The mean percentages correct at the input levels of 50, 60, 70 and 80 dB SPL for the three conditions are shown in Fig. 6A. In condition CLAIDHA, subjects achieved slightly higher scores than in the condition Own Aid for levels up to 70 dB SPL. An analysis of variance (ANOVA) on scores for the three conditions was conducted with factors subject, level (from 50 to 80 dB SPL) and condition. Condition was significant [$F_0 = 23.731$, $p < 0.01$] as was the interaction between level and condition [$F_0 = 20.004$, $p < 0.01$]. A paired *t*-test at each input level revealed that scores under the condition of CLAIDHA were significantly higher than those of Own Aid at 50 [$t = 4.131$, $p < 0.01$], 60 [$t = 4.505$, $p < 0.01$], and 70 [$t = 2.484$, $p < 0.01$], but not at 80 dB SPL.

For the 92 subjects for whom retesting for the CLAIDHA after daily use could be performed, the results for the retest were plotted as a function of the scores for CLAIDHA at the initial fitting (Fig. 5). Although more than half of the subjects showed better performance after daily use, these differences were not statistically significant at all input levels (paired *t*-test). The mean percent correct at the input levels of 50, 60, 70, and 80 dB SPL observed in the retest is shown in Fig. 6B as well as those for the other speech tests. The mean scores for the CLAIDHA after daily use were slightly higher than those at the initial fitting; these differences were not statistically significant at any input levels (paired *t*-test).

2. Questionnaire

Average results of the questionnaire for subjects who had used their own aids on the same ear as CLAIDHA are shown in Fig. 7. There was a clear trend for CLAIDHA to be preferred over Own Aid, except in terms of the ease of operation. These mean scores were subjected to Friedman two-way analysis of variance by ranks. The results indicated that there was a highly significant difference between the different aid conditions [$p < 0.001$]. To test the hypotheses that CLAIDHA gave better results than Own Aid, a Wilcoxon test was performed for each questionnaire statement. The results showed that the scores for CLAIDHA were significantly higher than those for Own Aid for all of the questions except for ease of operation of the aids. The mean score of ease of operation was higher for condition Own Aid, but the difference was not statistically significant.

Averaged scores, subgrouped by the type of audiogram, are shown for conditions CLAIDHA and Own Aid in Table I. Results of a Wilcoxon test for each questionnaire statement are also shown. For the flat and gradually sloping types, significantly better scores were obtained for CLAIDHA than for Own Aid for all the statements except ease of operation. On the other hand, in the subjects with other types of hearing loss, such as steeply sloping, and high and low tone loss, the number of better results for CLAIDHA was lower. When comparing the scores for conversational statements (3 and 4) among the different types of audiogram, greater improvements with CLAIDHA were observed in the subjects with flat and gradually sloping types of hearing loss than in the other types. The score for CLAIDHA for each column, shown in Table I, was subtracted from that for Own Aid (except for the case of comparison with own aid). The resulting scores for each audiogram type and each question of the questionnaire were subjected to

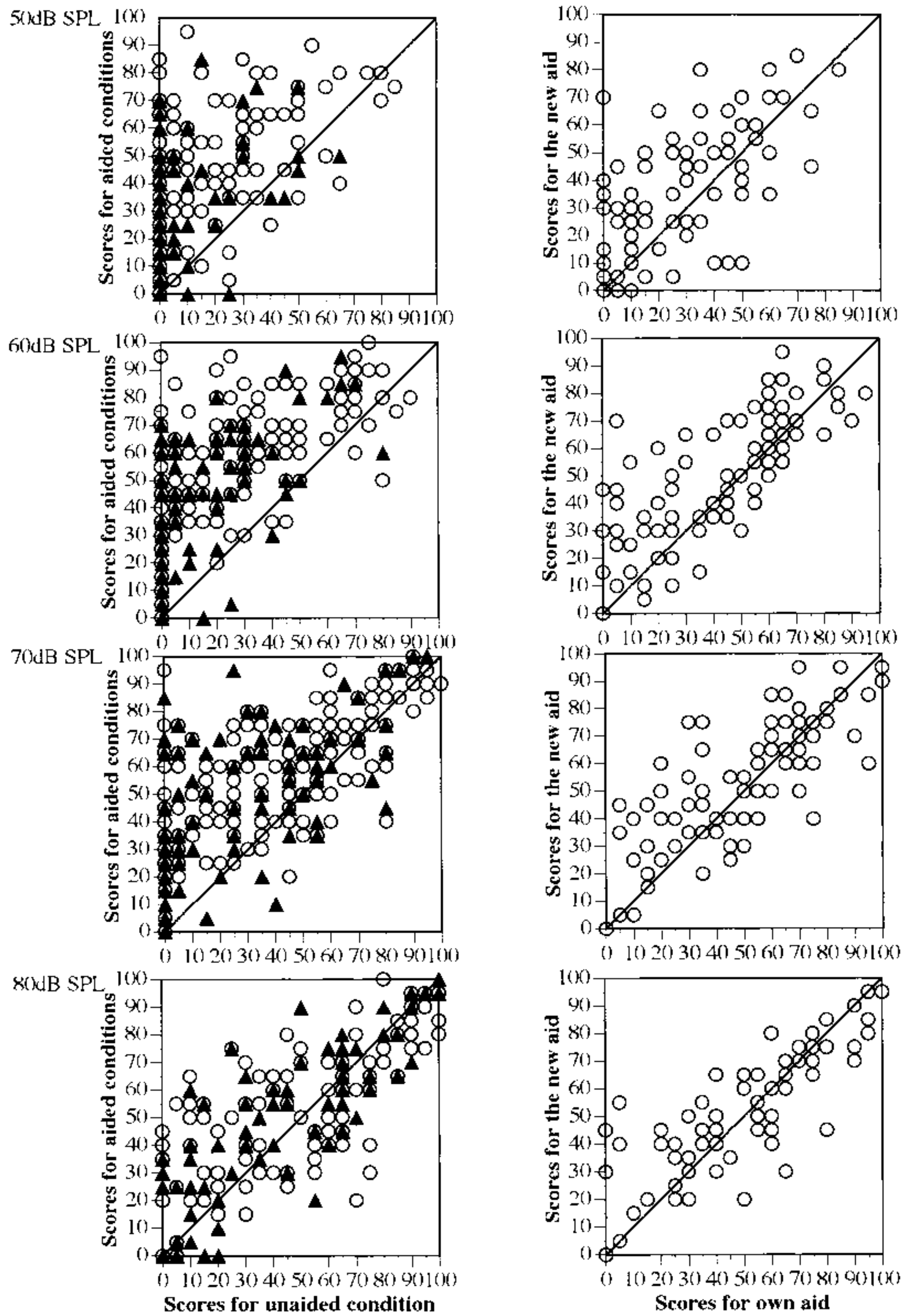

Fig. 4. Speech intelligibility at the levels of 50, 60, 70 and 80 dB SPL are shown from top to bottom. In the four figures on the left, the percent correct for aided condition (filled triangles: Own Aid, circles: CLAIDHA) is plotted as a function of that for unaided condition. In the four figures on the right, the score for CLAIDHA is plotted as a function of that for Own aid. See text for further details.

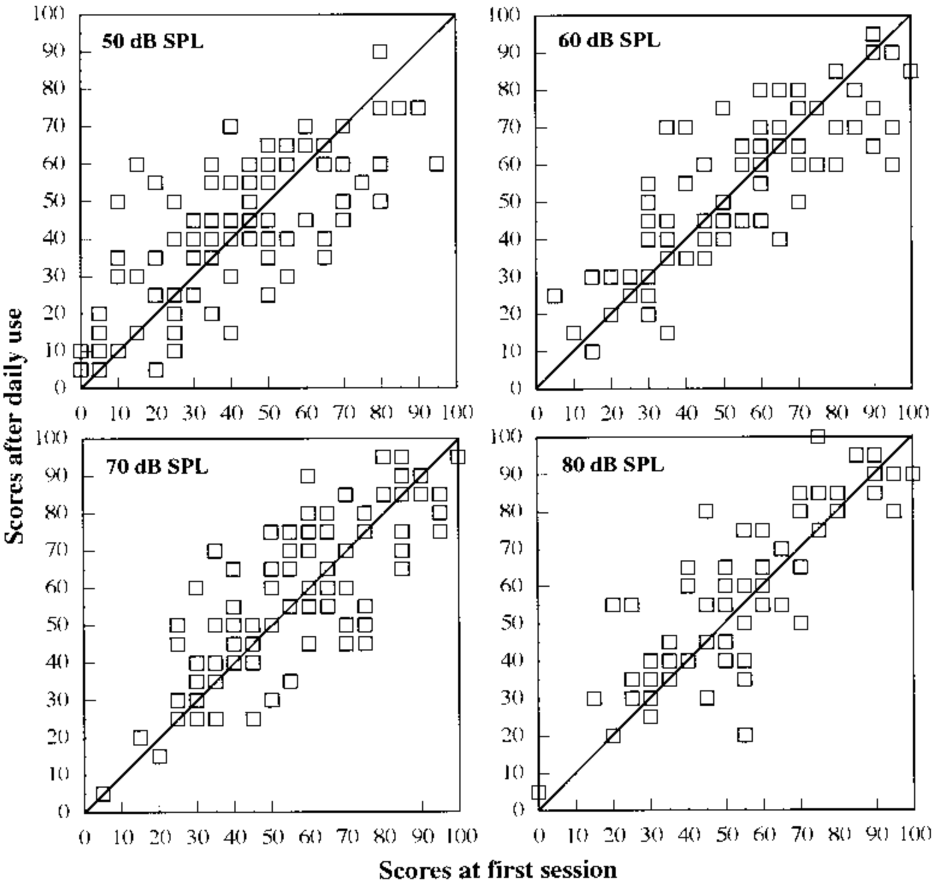

Fig. 5. Speech intelligibility at each level. The score for CLAIDHA after daily use is plotted as a function of that at the initial fitting.

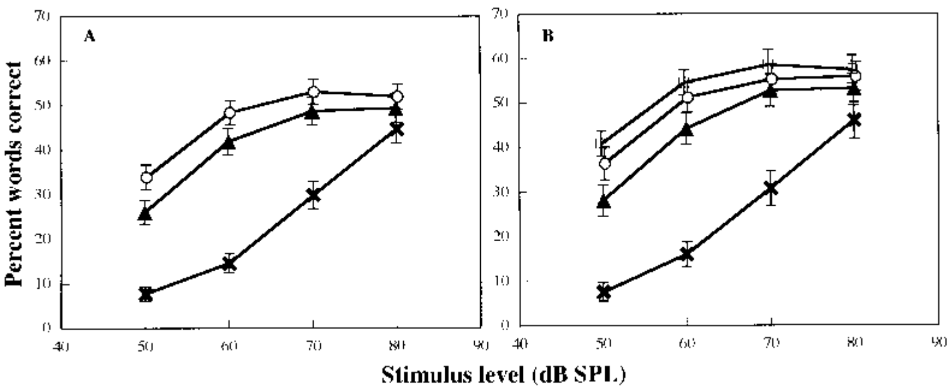

Fig. 6. Averaged data of speech intelligibility for different aided condition: Unaided (×), Own Aid (▲), CLAIDHA at the initial fitting (○) and that after daily use (□). The analyses were limited to the data of the 87 subjects who were tested in all three conditions (panel A). Forty-eight of these subjects were tested for the CLAIDHA after daily use and their averaged data is shown in panel B. Mean values of percent corrects for each aided condition are shown as a function of the RMS levels of speech. Error bars indicate ± standard errors (SE).

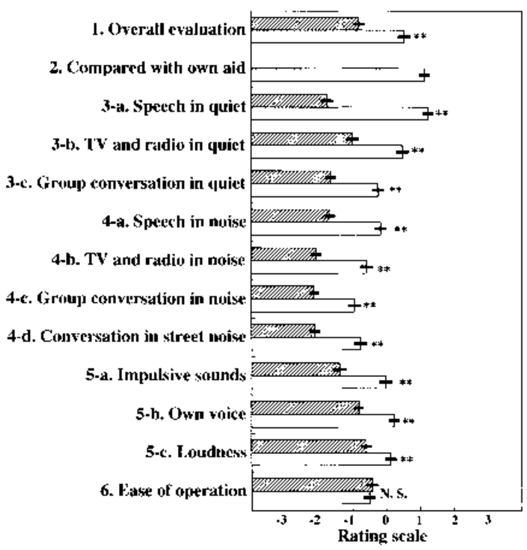

Fig. 7. Average results of the questionnaire. For question 2, the rating scale '–3', '–2', '–1', '0', '1', '2' and '3' meant 'much worse', 'worse', 'slightly worse', 'same', 'slightly better', 'better', and 'much better', respectively. For the other questions the rating scale of seven meant 'very bad', 'bad', 'a little bad', 'fair', 'a little good', 'good' and excellent', respectively. Among 128 subjects in which the subject's own aids were used in the same ear as that in which CLAIDHA was fitted, 12 could not fill in the questionnaire because they had not used the CLAIDHA in their daily lives; questionnaire results were averaged for the other 116 subjects. Error bars indicate ± standard errors (SE). In all items except 'ease of the operation', the scores for CLAIDHA were significantly higher than those for the subjects' own aids (Wilcoxon test $^{**}p < 0.01$).

Friedman two-way analysis of variance by ranks. These results indicated that there was a significant difference among the different audiogram types ($p < 0.001$).

3. Overall Outcome

After the clinical evaluation, otolaryngologists interviewed each subject and decided whether CLAIDHA was suitable or not. The prescription of CLAIDHA for each subject was decided based on both the subjective and objective usefulness of the aid. The size (i.e., weight, dimensions, etc.) and/or price were sometimes important reasons not to use CLAIDHA; the number of such subjects cannot be given separately because the final prescription was based on the comprehensive judgment arrived at through the interview. The numbers of subjects who bought and continued using CLAIDHA (applicable category) are shown in Table II. The subjects are subdivided based on past histories with regard to their experiences with hearing aids and the type of hearing loss of the ears in which CLAIDHA was fitted. The overall percentage of 'Applicable' subjects was 57.2. CLAIDHA tended to be more applicable in the subjects with flat-type and gradually sloping type hearing loss (64.3%) than in other types of hearing loss (48.6%).

Discussion and Conclusions

Comparison of CLAIDHA with Other Aids

Full-digital hearing aids, in which all signal processing and control are conducted digitally, have been tested in the laboratory for more than a decade (Levitt et al., 1986; Levitt, 1987; Engebretson et al., 1987; Kollmeier, 1991; Asano et al., 1991a, b; Hohmann & Kollmeier, 1995). However, it has been difficult to develop a system small enough to wear and sufficiently low in power consumption; only a few attempts have so far been made to evaluate these aids clinically both in the laboratory and in daily life (Lunner et al., 1997; Arlinger et al., 1998). We have developed a portable digital hearing aid system (CLAIDHA IV) based on the loudness compensation principle and have clinically evaluated it as detailed in this article.

One of the difficulties in evaluating new hearing aid systems is the choice of an appropriate control and the conditions to be used for comparison (Moore et al., 1992). We previously used the same hardware to realize the CLAIDHA system as well as linear hearing aid and showed some benefits of using CLAIDHA over linear amplification in a laboratory experiment (Asano et al., 1991a, b; Suzuki & Sone, 1993). Since the main purpose of this study, on the other hand, was to evaluate the CLAIDHA system in use in daily life, we compared the CLAIDHA with the subject's own aids. In such a comparison, one can never rule out the possibility that some factors other than the effects of loudness compensation (for example, frequency characteristics) might be responsible for the differences of the results. As reported by Lippmann et al. (1981), increases in speech scores with compression may be caused by the use of inferior linear reference systems. Another problem is that subjects might select smaller gains than those required to provide optimum speech intelligibility to avoid unpleasantly loud sounds (Leijon, 1989). But considering that most of the subjects' own aids had been fitted in professional fitting centres or by medical doctors, the present comparison between CLAIDHA and the subjects' own aids is meaningful at least clinically. Moreover, as the gain-frequency characteristics of the subjects' own aids, POGO (McCandless, 1983) and the simple half-gain rule (Lybarger, 1944), often with some low-cut characteristics, are mainly used at professional

Table I. *Mean questionnaire score for each statement and each audiogram type for CLAIDHA and Own Aid (Wilcoxon test; **p < 0.001, *p <0.05, N.S.; not significant). The definitions of each type of audiogram are as follows: (A) the flat type was defined as a hearing loss in which hearing levels for each frequency did not differ by more than 10 dB throughout the speech frequencies (from 250 to 4000 Hz), (B) the gradually sloping type was a loss pattern that sloped from low frequencies to high frequencies with a loss less than 20 dB between each neighbouring octave-spaced speech frequency, (C) the steeply sloping type was a high tone loss that was not included in the gradually sloping type, (D) the high and low tone loss was a pattern showing normal thresholds in the middle frequency range (between 500 and 2000 Hz), with a sloping curve to the lower and higher frequencies, (E) others (low tone loss and conductive hearing loss, which comprised relatively fewer cases, were included)*

	Aid	A) Flat type	B) Gradually sloping	C) Steeply sloping	D) High and low tone loss	E) Others
1. Overall evaluation	Own Aid	−1.28	−0.73	−0.71	−0.58	−1.33
	CLAIDHA	0.77**	0.90**	0.00 (n.s.)	0.67*	0.05**
2. Comparison with Own Aid	CLAIDHA	1.50	1.25	0.46	1.27	0.75
3-a. Speech in quiet	Own Aid	−0.52	0.10	0.42	−0.31	−0.05
	CLAIDHA	1.50**	1.41**	1.00 (N.S.)	1.08*	0.86*
3-b. TV and radio in quiet	Own Aid	−1.84	−0.36	−0.91	−1.08	−1.17
	CLAIDHA	0.50**	1.19**	0.43 (N.S.)	−0.25 (N.S.)	−0.21*
3-c. Group conversation in quiet	Own Aid	−1.96	−1.55	−1.50	−1.92	−1.75
	CLAIDHA	−0.24**	0.15*	0.57**	−0.77*	−0.79*
4-a. Speech in noise	Own Aid	−1.89	−1.44	−1.58	−1.92	−1.78
	CLAIDHA	−0.08**	0.00**	−0.50*	−0.39**	−0.25**
4-b. TV and radio in noise	Own Aid	−2.46	−1.90	−2.17	−2.23	−2.11
	CLAIDHA	−0.50**	0.07**	−0.75*	−1.42*	−1.31 (N.S.)
4-c. Group conversation in noise	Own Aid	−2.35	−2.16	−2.25	−2.31	−2.24
	CLAIDHA	−1.05**	−0.62**	−1.00*	−1.39*	−1.29*
4-d. Conversation in street noise	Own Aid	−2.33	−1.97	−2.55	−2.23	−2.11
	CLAIDHA	−0.68**	−0.52**	−1.23*	−1.56 (N.S.)	−0.78*
5-a. Impulsive sounds	Own Aid	−1.36	−1.66	−1.67	−1.00	−1.00
	CLAIDHA	0.00*	0.09**	−0.21 (N.S.)	−0.46 (N.S.)	0.25*
5-b. Own voice	Own Aid	−0.86	−0.72	−1.00	−0.67	−0.44
	CLAIDHA	0.23*	0.15**	0.21*	0.09 (N.S.)	0.43*
5-c. Loudness	Own Aid	−1.00	−0.27	−0.64	−0.55	−0.6
	CLAIDHA	0.36**	0.06*	−0.43 (N.S.)	0.00 (N.S.)	−0.10 (N.S.)
6. Ease of operation	Own Aid	−1.11	−0.09	−0.50	−0.08	−0.44
	CLAIDHA	−0.19 (N.S.)	−0.41 (N.S.)	−1.00 (N.S.)	−0.77 (N.S.)	−0.85 (N.S.)

Table II. *Numbers (and percentages) of subjects in the 'applicable' categories, classified according to the type of hearing loss and prior experience with aids*

	A) Flat type	B) Gradually sloping	C) Steeply sloping	D) High and low tone loss	E) Others	Total
Used own aid	15/27 (55.6%)	20/32 (62.5%)	5/13 (38.5%)	7/13 (53.8%)	10/23 (43.5%)	57/108 (52.8%)
Had experience but did not use own aid	4/5 (80.0%)	4/5 (80.0%)	3/4 (75.0%)	1/3 (33.3%)	2/3 (66.7%)	14/20 (70.0%)
Used own aid on the contralateral side	2/2 (100.0%)	1/1 (100.0%)	2/3 (66.7%)	0/0 (0.0%)	0/0 (0.0%)	5/6 (83.3%)
Did not have own aid	4/8 (50.0%)	6/7 (85.7%)	3/5 (60.0%)	1/3 (33.3%)	1/2 (50.0%)	15/25 (60.0%)
Total	25/42 (59.5%)	31/45 (68.9%)	13/25 (52.0%)	9/19 (47.4%)	13/28 (46.4%)	91/159 (57.2%)

fitting centres in Japan. This was confirmed by inspecting the gain-frequency characteristics of the Own Aids for the sampled population of subjects.

On the other hand, in recent years, NAL-R procedures (Byrne & Dillon, 1986) have been the most popular formulae suitable for linear hearing aids (Killion, 1996; Kiessling et al., 1996). The NAL-R procedures, which suppose all frequency regions in amplified speech to be equally loud at a comfortable level (Van Tasell, 1993), provide relatively greater weighted gain in the low frequency region (Killion, 1996) than procedures such as POGO. Since the gain realized by CLAIDHA also seemed to be greater than that of the subjects' own aids in the low frequency region, the NAL-R procedures and CLAIDHA should show somewhat similar gain-frequency characteristics for a constant input level. Therefore, for purposes of compression with CLAIDHA, we are planning an experiment in which this hearing aid system will also be installed with a software program acting as a linear aid prescribed by the NAL-R procedures. Such a study may provide evidence as to whether the effects of CLAIDHA are attributable to the loudness compensation functions or to the gain-frequency characteristics, or to both.

Previous Work with Compression

As most of the subjects showed recruitment in their loudness functions to a greater or lesser degree, a kind of compression amplification was usually applied with the CLAIDHA. The beneficial effects of CLAIDHA on speech perception in a quiet environment, which were shown by the results of the speech test and the questionnaire in the present study, are basically consistent with recent evidence on hearing aids with compression systems; it is known that compression systems (single-band, two-band and multi-band systems) allow speech to be understood over a wide range of sound levels (Bustamante & Braida, 1987, Laurence et al., 1983; Lippmann et al., 1981; Moore, 1987; Moore & Glasberg, 1986, 1988; Moore et al., 1991, 1992; Biering-Sörensen et al., 1995). As compared with the high score for CLAIDHA seen in the questionnaire, those seen in the speech tests were relatively small and the practice effects of use in daily life were not statistically significant (Figs. 5 and 6B). The '67-S' Japanese monosyllabic identification test is less time-consuming and one of the most widely used tools for evaluating the effects of hearing aids in Japan; we clinically evaluated speech hearing ability using this method. Recent studies have revealed, however, that there is a significant discrepancy between the frequencies of monosyllables appearing in Japanese conversation and those in the '67-S' lists (Kodera & Hiraishi, 1998). Though several studies are being attempted to develop more effective tools for estimating the effects of hearing aids (Kodera et al., 1997), they are not yet generally available as clinical instruments.

In this study, as indicated by the results of the questionnaire, CLAIDHA did not show negative effects of speech in noisy environments. A multiband syllabic compression system with more than several channels (for example 8 or 16) was first advocated by Villchur (1973). Since then, its effects on speech as well as its interactions with characteristics of impaired hearing have been controversial (Villchur, 1973, 1989; Plomp, 1988). On the other hand, multiband compression systems with a small number of bands (two or three) can improve the ability to understand speech in a noisy environment (Laurence et al., 1983; Moore, 1987; Moore & Glasberg, 1986, 1988; Moore et al., 1985, 1991, 1992; Ringdahl et al., 1990). As for the CLAIDHA, its frequency-sampling-type digital filter can be regarded as a bandpass filter consisting of N/2 channels, where N is the order of the filter. In the present study, N was set at 48, and thus the filter acted as a 24-channel bandpass filter bank. However, in CLAIDHA, these channels do not process the input signal independently, but the gains in adjacent frequency bands are determined to exhibit smooth gain-frequency characteristics as described in the previous section. This might make it possible to avoid the defects of multiband compression systems consisting of many independently controlled channels, i.e., spectral distortion or flattening (Plomp, 1998). Furthermore, if these defects are overcome, a multiband system can deal more effectively with the changes of the loudness function as a function of frequency than systems with a few bands.

Different Effects Among Types of Audiogram

The advantages of CLAIDHA over the subjects' own aids differed among the different types of audiogram. As shown in Tables I and II, subjects with a steeply sloping type of hearing loss demonstrated fewer consistent benefits of CLAIDHA over their own aids than those with the flat or gradually sloping types. In such cases, hearing at higher frequencies may be so damaged that no useful information can be extracted from the higher frequency region (Moore & Glasberg, 1997); attempts to amplify this frequency region can be counterproductive (Ching et al., 1997). If so, the compensation of loudness at these higher frequency bands cannot improve analysis of sound and extraction of cues from the speech nor does it make any difference compared with high-frequency emphasis in linear amplification. This idea can also be used when considering the relatively poor results in subjects with low tone loss. A person with a low-frequency hearing loss may lack neurons with low characteristic frequencies (CF) and may detect low frequencies via stimulation of neurons with medium to high characteristic frequencies (Moore, 1991; Turner et al., 1983; Thoronton & Abbas, 1980). The compensation of loudness at these low frequencies might adversely affect extraction of the speech cues due to the effect of masking, which spreads upward to higher frequencies.

Modification of Fitting Parameters

One of the advantages of the CLAIDHA system is that the gain-frequency response can be determined very flexibly as a function of frequency and sound level of the input signal. This flexibility can introduce greater complexity in fitting the aid and in determining the optimum fitting parameters. The basic LCFs, which were measured using 1/3 octave noise bands, were preferred to their own aids by many subjects. Approximately 33% of the subjects preferred the gain-frequency characteristics which are decreased from the basic LCFs at lower frequencies. These results agree with those of Kiessling et al. (1996), who stated that complete loudness compensation in the low frequency channel should be avoided so as not to create too much of an upward spread of masking.

For some of the subjects, the basic LCFs needed to be modified before they would accept the modified LCFs for daily use (see Material and Methods). Simple signals such as steady bands of noise are far removed from the speech and other complex signals that users wish to hear in daily life (Moore et al., 1992). It might be better to use time-variant speech stimuli for the measurement of the loudness function of the subjects as suggested by Moore et al. (1992) and to check whether or not the aided loudness function has been successfully compensated for in order to avoid inappropriate loudness perception due to the channel interaction between adjacent bands (Kiessling et al., 1996).

In spite of applying modifications of the LCFs due to unsatisfactory impressions of the basic LCFs, some subjects, in the end, preferred the basic LCFs once they had used the aid in daily life for several weeks and had become accustomed to using the CLAIDHA. Subjects who have suffered from hearing impairment for many years may need time to become accustomed to hearing aids and to using the cues they provide (Gatehouse, 1992). Several authors reported that learning to use a compression aid seems to be important (Yund et al., 1987; Laurence et al., 1983). Further studies are necessary to evaluate the practice effect with the CLAIDHA system and to develop more effective or clinically applicable fitting procedures.

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