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디지털 보청기 알고리즘 평가를 위한 감음신경성 난청의 모델링

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Modeling of Sensorineural Hearing Loss for the Evaluation of Digital Hearing Aid Algorithms

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요 약 : 디지털 보청기는 기존의 아날로그 보청기에 비하여 많은 장점이 있다. 디지털 신호처리 프로세서의 발달과 더불어 최근에 다양한 디지털 보청 알고리즘과 완전한 디지털 보청기가 선보였다. 디지털 보청기의 알고리즘을 개발하거나 디지털 보청기를 새로이 평가하려는 사람들에게 난청자를 대상으로 하는 임상연구는 필수적으로 거쳐야 하는 과정이다. 그러나 이러한 임상연구는 실제 난청자를 대상으로 하여야 하기 때문에 난청자와 검사자 간에 통상적으로 많은 시간과 노력이 필요하며 원활한 의사 소통이 때로는 어려울 수 있다. 왜냐하면 난청자들의 연령이 너무 어리거나 많아서 의사소통에 지장을 주거나 검사자가 필요로 하는 시간에 비슷한 난청 유형을 가진 대상자를 모으기 어렵다. 본고에서는 임상연구를 보조하여 디지털 보청기 또는 알고리즘이 개발되기까지 수행되어야 할 많은 임상연구의 결과를 예측하고 평가할 수 있는 디지털 난청 시뮬레이션 방법을 제안하고, 실제 환자의 데이터를 사용한 시뮬레이션과 그에 대한 임상 실험을 통하여 시스템의 성능을 평가하였다. 실험 결과, 정상인으로부터 모델링된 환자 데이터와 매우 유사한 측정 결과를 얻어냄으로써, 제안된 시스템이 목적하고자 하는 바를 이룰 수 있음을 검증하였다. 또한 난청 시뮬레이터의 목적인 디지털 보청기 알고리즘을 개발하기 위한 평가 툴로서, 개발 초기에 다양한 디지털 보청기용 알고리즘을 구현하여 실제 난청 시뮬레이터와 연계하여 실험함으로써 보청기 알고리즘의 평가 및 새로운 보청기 알고리즘을 개발하고 평가하거나 향후 난청자를 대상으로 하는 임상연구에서 사용할 수 있는 유용성을 입증하였다.

Abstract : Digital hearing aids offer many advantages over conventional analog hearing aids. With the advent of high speed digital signal processing chips, new digital techniques have been introduced to digital hearing aids. In addition, the evaluation of new ideas in hearing aids is necessarily accompanied by intensive subject-based clinical tests which requires much time and cost. In this paper, we present an objective method to evaluate and predict the performance of hearing aid systems without the help of such subject-based tests. In the hearing impairment simulation (HIS) algorithm, a sensorineural hearing impairment model is established from auditory test data of the impaired subject being simulated. Also, the nonlinear behavior of the loudness recruitment is defined using hearing loss functions generated from the measurements. To transform the natural input sound into the impaired one, a frequency sampling filter is designed. The filter is continuously refreshed with the level-dependent frequency response function provided by the impairment model. To assess the performance, the HIS algorithm was implemented in real-time using a floating-point DSP. Signals processed with the real-time system were presented to normal subjects and their auditory data modified by the system was measured. The sensorineural hearing impairment was simulated and tested. The threshold of hearing and the speech discrimination tests exhibited the efficiency of the system in its use for the hearing impairment simulation. Using the HIS system we evaluated three typical hearing aid algorithms.
Key words : Digital hearing aid, Sensorineural hearing loss, Hearing impairment simulation (HIS), Digital signal processing, Real-Time system

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INTRODUCTION

With the loss of hearing, a person is restricted from his or her normal social activity. In order to compensate for this kind of handicap, many researches have been conducted. Most of them have been focused on developing aid devices [1]. Recent developments in digital technology have offered new possibilities of noticeable advances of hearing aids. Using digital technology, it is possible to equip hearing aids with powerful features, such as nonlinear amplification, noise reduction, and enhanced fitting algorithms, that are often difficult to implement with analog circuits [1,2,3,4].

Although various ideas and methods have been suggested so far, there still exist many problems to be solved to meet the urgent requirements of the hearing impaired. The most important issue that must be considered in developing stage of a hearing aid system would be how to effectively reflect the "feeling" experienced by the impaired listeners. In addition, evaluation of new ideas in hearing aids is necessarily accompanied by intensive clinical tests on impaired subjects. However, subject-based clinical tests require much time and cost. Moreover, their potential problems are that 1) it is difficult for a patient to be involved continuously throughout the entire session, and 2) unreliable response from too old or too young aged patients. So, it is strongly required to develop a way to evaluate and predict the performance of the hearing aid system without help of subject-based tests. Rutledge [5] modeled impaired listener's transfer function using neural network and developed an objective measure from the function. And Chabries's approach [6] involved the modeling of an impaired listener's parameters based on a filterbank system when developing hearing aid. Rutledge's neural network is hard to manipulate auditory parameters, and Chabries's filterbank method is too complex to implement in real-time system.

We present a simulation tool for the hearing impairment of sensorineural hearing loss. In our hearing impairment simulation (HIS) tool, the impairment is modeled using the data from auditory tests conducted for an impaired patient with sensorineural hearing loss. Then, a frequency sampling filter is designed with which input signals are modified according to the gain function obtained from the modeled impairment. The selection of gains in the function for the filter design is controlled by the energy values in 20 frequency bands. Thus, each input signal is nonlinearly processed as a function of its frequency and intensity. The key feature of the HIS tool is its ability to simulate the feeling experienced by the impaired listener and make it possible to perform subjective tests without hearing impaired patients. Its best application would be in the evaluation of the hearing aids or algorithms for hearing aids. To test the effectiveness of the proposed HIS tool, it is implemented in a real-time system employing a floating-point digital signal processor (Motorola DSP96002). The processed signals are presented to normal subjects and their responses are then investigated. The test results of normal listeners are compared with those of impaired subjects.

We implemented three typical hearing aid algorithms, such as an 5-channel amplitude compression filterbank method, a linear gain filterbank method, and the nonlinear wide dynamic range compression using a single filter algorithm [7], on the DSP board.

MATERIALS AND METHOD

1. The Hearing Impairment Simulation (HIS) Algorithm

Hearing loss causes degradation in loudness perception, which can be modeled by matching the output of the normal ear to that of the impaired ear. By focusing on the auditory function rather than on the characteristics of the input audio signal, the resultant signal processing will be effective independent of the auditory stimulus. However, the structure of the modeling system will limit the complexity of the algorithm that can be implemented in a hearing aid system.

The audible thresholds of the normal listener are much lower than the range where most of speech signal components exist. However, as a hearing loss process proceeds, the audible thresholds begin to rise, so that it becomes increasingly difficult for the impaired person to hear weak level sounds. In the conductive hearing loss caused by problems in the outer ear and/or the middle ear, the sound level is reduced

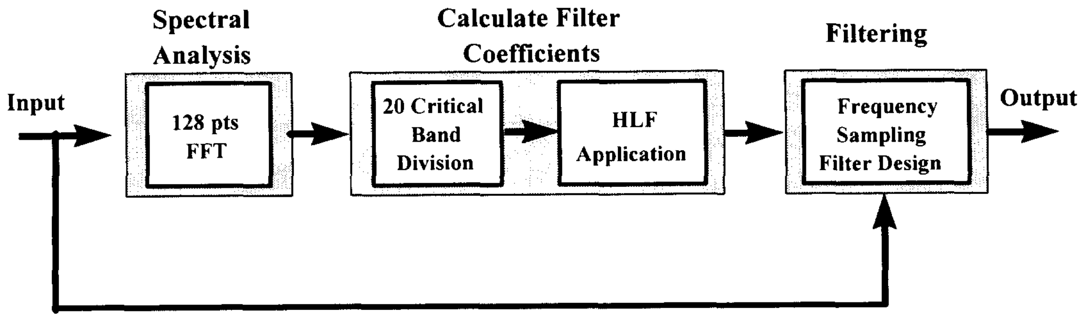

Fig. 1. Block diagram of Hearing Impairment Simulation(HIS) System
FFT : fast fourier transform, HLF : hearing loss function

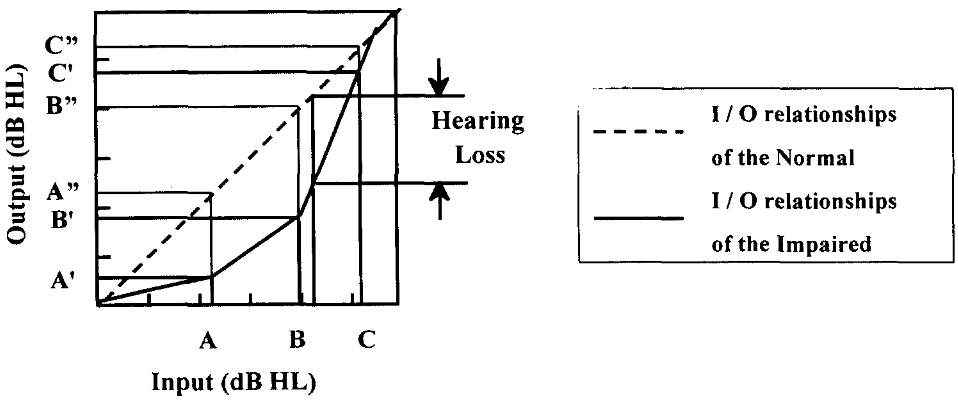

Fig. 2. Hearing Loss Function(HLF)
ABC : input level, A′B′C′ : perceived level of sensorineural hearing loss, A″B″C″ : perceived level of normal hearing, A : TH, B : MCL, C : UCL

uniformly throughout the audible frequency region. Thus, in terms of the hearing impairment modeling, this type can be feasibly modeled with linear electronic circuits. But, for sensorineural hearing loss, the problem is more complex because it is caused by problems in the cochlea or in the auditory nervous system. In this case, nonlinear modification of the audible range occurs a phenomenon known as loudness recruitment [8]. Thus, sensorineural hearing impairment is mainly characterized by two parameters: elevation of the threshold of hearing, and nonlinear mapping between the sound pressure level of natural acoustical signals and the perceived loudness.

In order to model this type of hearing impairment, the modeling process is based on a system operating on the full frequency bands and dynamic range. A nonlinear dynamic processing which transforms input according to the level and frequency of the input is required.

2. Signal Processing in the HIS

The processing flow of the HIS algorithm is similar to the nonlinear multiband loudness correction hearing aid [2,3]. The HIS algorithm is divided into three major processing steps as shown in Fig. 1: spectral analysis, computation of hearing loss gains, and designing of the FIR filter.

In order to have an accurate model for the impaired mapping associated with the loudness recruitment, the critical band (CB) processing [9] is employed. To estimate the energy of the input signal at each CB, the input signal sampled at 16kHz is segmented with Blackman window without overlap. The 16kHz sampling rate covers roughly 20 CBs. Each segment is 128 samples corresponding to 8 ms time period. Successive short-term spectra are then calcu-

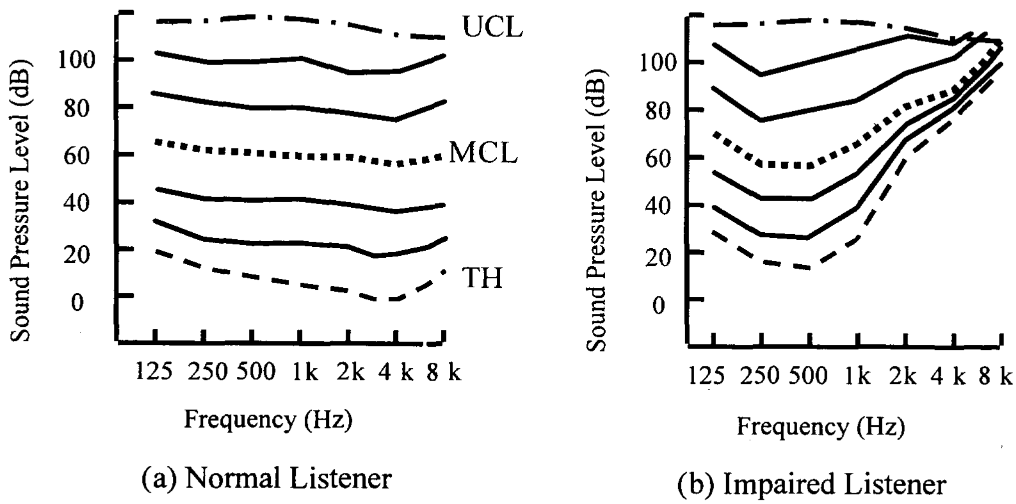

Fig. 3. Graphic showing equal loudness curve and MCL of normal(a) and impaired listeners(b)

lated using a 128-point FFT. In the algorithm, the short-term energy rather than the magnitude is computed in each CB to reflect psychoacoustical features.

The Blackman window was chosen for the input segmentation, because it is known to have a much lower sidelobe level than Hamming or Hanning windows frequently used for speech analysis [3]. The sidelobe level is particularly important in high frequency bands due to the low signal levels. Sometimes, the low-level inputs in high frequency bands are masked by the high-level inputs in low frequency bands, which in turn affects the loudness correction of the HIS system in those bands.

To define the impaired mapping between the sound pressure level of the natural acoustical signals and the perceived loudness of the impaired auditory system, hearing loss functions (HLFs) are generated from the measurements. These functions describe how much gain is needed for every frequency band and the level to restore normal loudness perception. The HLF's are generated from three auditory parameters: threshold of hearing (TH), most comfortable level (MCL), and uncomfortable level (UCL) which represents the upper limit of hearing range. These three parameters are obtained from the audiogram of a patient. Fig. 2 illustrates the HLF and auditory parameters used to model the hearing impairment. In the figure, A′, B′, C′ and A″, B″, C″ represent THs, MCLs, and UCLs for the hearing impaired and the normal listener's, respectively.

Simulating a phenomenon of loudness recruitment modeling may require all the information on equal loudness curves of the impaired and the normal subject's (Fig. 3). However, it is impractical to measure the loudness for every input level at every frequency. Furthermore, loudness itself is based on subjective feelings, so it is hard to obtain accurate data solely from the measurements. Fortunately, it is known that the MCL represents the characteristics of equal loudness curves and that it can be measured relatively precisely, especially in most cases of sensorineural-type hearing loss [6]. From this point of view, the MCL measurements are used as a reference parameter in loudness level that prescribes a proper mapping from audible range of the normal listener's into that of the impaired listener's.

Based on these aspects, the energy estimates in the 20 CBs are applied to HLFs and the hearing loss gains are computed. However, in the implementation stage, we computed the gains only at five center frequencies of octave bands, *i.e.*, 250 Hz, 500 Hz, 1 kHz, 2 kHz, 4 kHz, and then interpolated them linearly while all spectral components in a CB have the same gain. This approach is chosen because the hearing loss is often measured at a discrete number of octave frequencies, rather than CB frequencies.

To produce outputs modified according to the frequency gains obtained from the HLF, a frequency

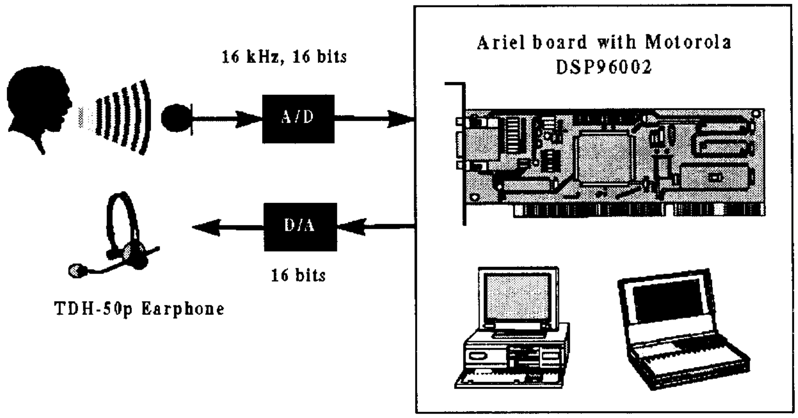

Fig. 4. The schematic diagram of the HIS system

sampling filter (FSF) is designed. The filter-bank approach can be considered to obtain the same output. However, processing each band independently in a multi-channel system using filter banks can cause spectral distortions among bands due to time-varying nonlinear characteristics of the filter gains [2,10], and may result in spectral flattening [3]. In addition, adding more processing bands for higher resolution increases the complexity of the filter-bank approach significantly [11].

Since the FSF obtains it's coefficients from the magnitude response function sampled uniformly in the frequency domain, it provides an effective and flexible method to implement loudness attenuation [7,12]. Furthermore, this approach enables one to avoid the problems associated with the filter-bank approach. The filter coefficients are updated at every input block and the input signal is finally modified via convolution operations.

3. Three Digital Hearing Aid Algorithms

Currently available three digital hearing aid algorithms such as the 5-channel amplitude compression filterbank method, the linear gain filterbank, and the nonlinear wide dynamic range compression (WDRC) algorithm were implemented in real-time on the Ariel's DSP board. Both the amplitude compression method and the linear gain method used 5-channel filterbank hearing aid algorithms. The amplitude compression filterbank method was implemented both using FIR and IIR filters. These two algorithms used the same gain function. The nonlinear WDRC algorithm determines the gain in each frequency band in such a way that the hearing perception of the impaired listener is restored close to that of normal listeners. The WDRC algorithm used the single filtered method using the frequency sampling filter which had same filter structure of the HIS system. A hearing compensation function is used instead of the HLF for the nonlinear WDRC algorithm. We also developed the hearing aid processor which employing the nonlinear WDRC algorithm. The chip is based on a general purpose 16-bit Samsung's DSP core which is able to perform maximum 33 MIPS at 33 MHZ clock.

4. Experimental System Setup

The schematic diagram of the system setup for the experiments is shown in Fig. 4. A floating-point digital signal processor (33 MHz Motorola DSP96002) was used to implement the algorithm in real-time. The processor is a part of an Ariel DSP board installed in a PC-bus slot with a 16-bit AD/DA CODEC unit.

The Grason-Stadler's SGI 61 audiometer was used to supply the input signal to the system. The input signal from the audiometer was amplified, anti-aliasing filtered and converted to digital numbers. The output signals of the HIS system were also anti-imaging filtered and presented to the subject under test via a Telephony's TDH-50p headphone. The HIS system was calibrated to have 0 dB input-output gain, so that the attenuation by the system could be directly observed at the output of the system. The DSP board was controlled by the PC and its status was moni-

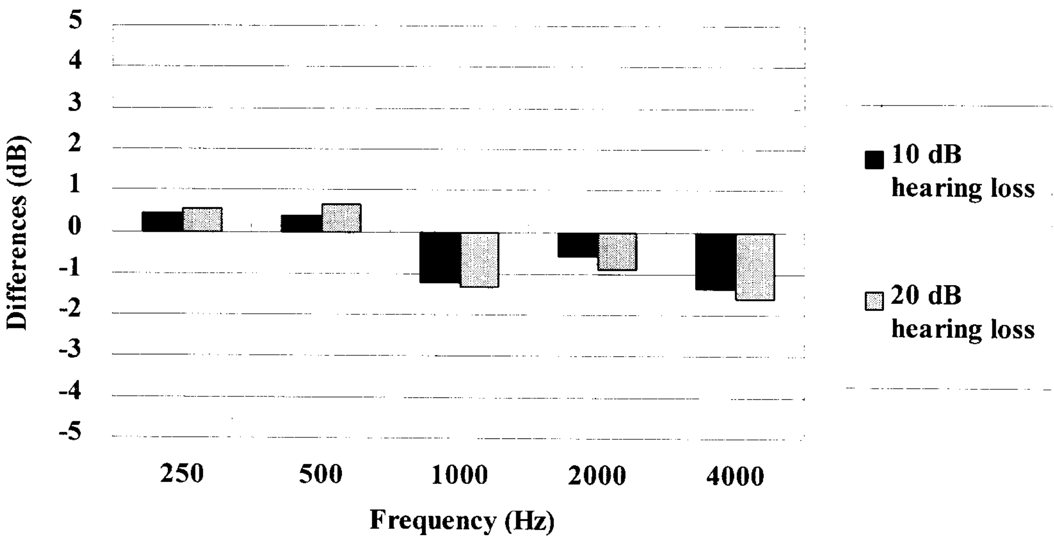

Fig. 5. Difference between modeled signal level and the output level of the audiometer

tored on the PC screen. The total processing time of the HIS algorithm was about 52% of the maximum processing time in the real-time DSP board.

Before we evaluated the digital hearing aid algorithms using HIS system, the HIS system was evaluated by comparing the hearing impaired patient's audiometry data with the simulated results. Speech audiometry tests were used to compare the digital hearing aid algorithms.

RESULTS

Three tests were conducted to evaluate the performance of the HIS system. The first test was to evaluate the numerical accuracy of the system. The HIS system and the audiometer were set to generate the same level of test signals and the output levels were measured for comparison. The second test was intended to simulate sensorineural hearing impairment. The audiological data of the impaired subject was given to the HIS system *a priori*, and the processed output was presented to normal subjects to measure their responses. The pure tone audiometry test (PTA : air-conduction audiometry) and the speech discrimination test (SDT) were performed to compare the HIS test results with the audiometry data of hearing impaired listeners. Finally, we evaluated three digital hearing aid algorithms using the HIS system and measured the SDT scores of them.

A. Objective Measurement of HIS Output

To assess the numerical accuracy of the HIS system, the conductive hearing loss i.e., a uniform hearing loss over the entire frequency was assumed. A 90 dB pure tone was applied to the input and results were obtained as RMS values measured at 5 frequencies; 250 Hz, 500 Hz, 1 kHz, 2 kHz, and 4 kHz. The results for two different cases, 10 dB and 20 dB hearing loss, are presented in Fig. 5. These results show that the HIS system successfully modifies the thresholds of hearing with an acceptable margin of error. The measured errors were smaller than 1.6 dB. Furthermore, the errors were considered insignificant when compared with the measurement error which may occurred during the subjective tests.

B. Simulation of Sensorineural Hearing Impairment

In the second test, the sensorineural hearing impairment was modeled using the HIS system. We chose two hearing impairments to simulate. The first hearing model was the moderate degree of sensorineural hearing loss of a 43-year old female subject (the right ear). The hearing was the flat form on the audiogram; 72 dB HL of the MCL and 80% of the SDT score. The first hearing impairment simulation was tested with 4 normal hearing listeners, aged between 20 and 32. The other simulation hearing model was the hearing of 64-year old male subject (the left ear) with the sensorineural hearing impairment. The

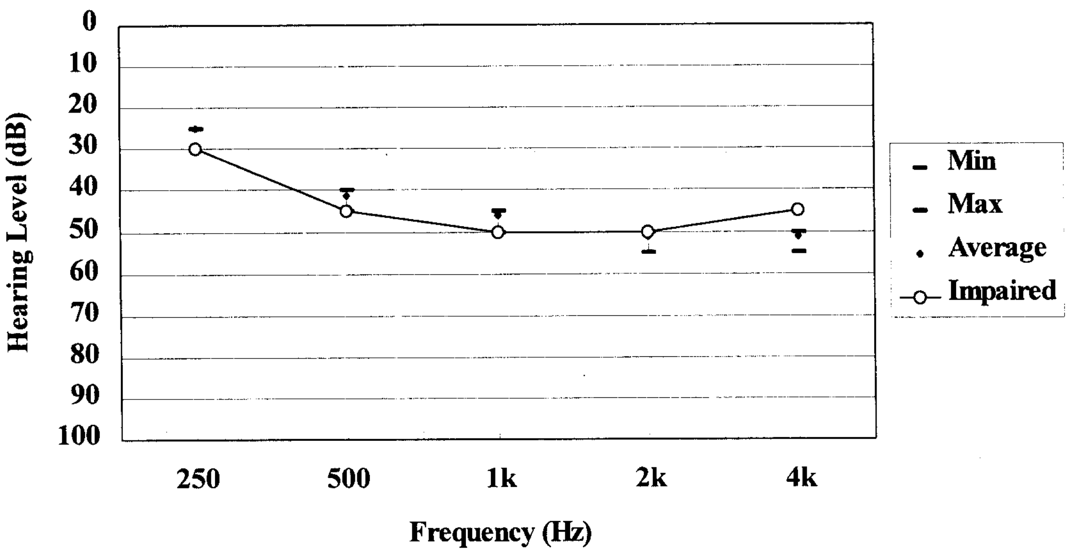

Fig. 6. Pure tone audiogram of impaired listener and average, minimal(Min), and maximum(Max) values of 4 HIS-processed subjects

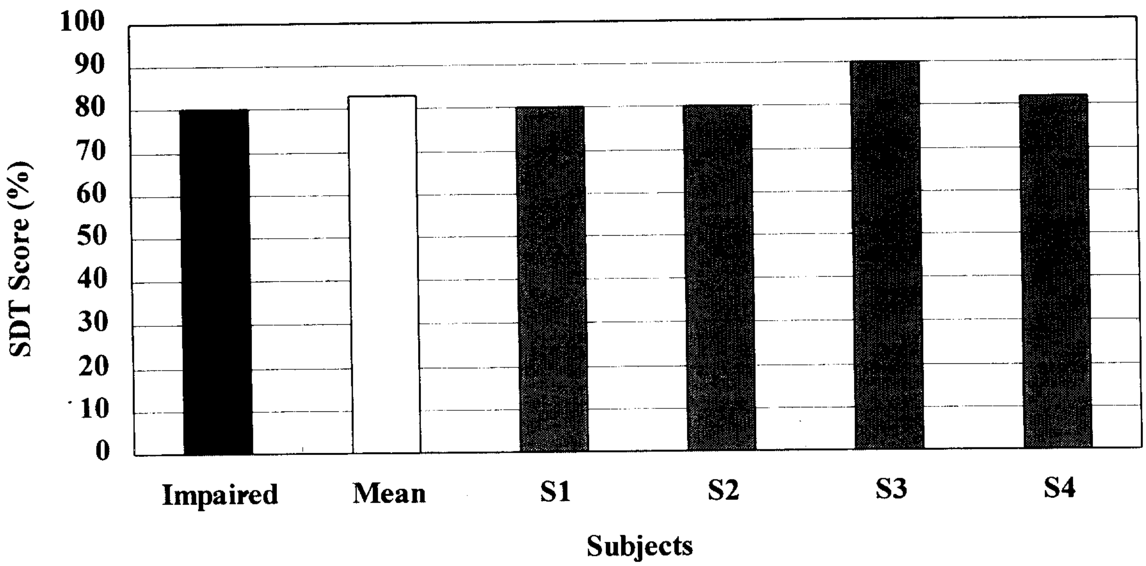

Fig. 7. SDT scores of HIS-processed listeners

target subject had a severe degree of high frequency hearing loss; 78 dB HL of the MCL and 88% of the SDT score. The HIS system was tested by 11 hearing listeners, aged between 21 to 33. The hearing thresholds of the 7 listeners were qualified as normal hearings (i.e., whose hearing thresholds were below 20 dB for whole frequency range and SDT scores were over 96%) and the other four subjects had high frequency hearing loss. Three subjects showed the C_5 (4kHz) dip and the other showed high frequency hearing loss above 8 kHz.

The thresholds of hearing were measured again with the HIS system. The results are presented in Fig. 6 and Fig. 8. As shown in the figures, the threshold levels of the tested subjects are different from those of the case without the HIS system. Although these results varied from person by person, the averaged thresholds exhibited similarity among the target model and the simulated ones.

The SDT scores were measured along with the threshold tests. The results are shown in Fig. 7 and Fig. 9. The bars with the label of 'Impaired' and

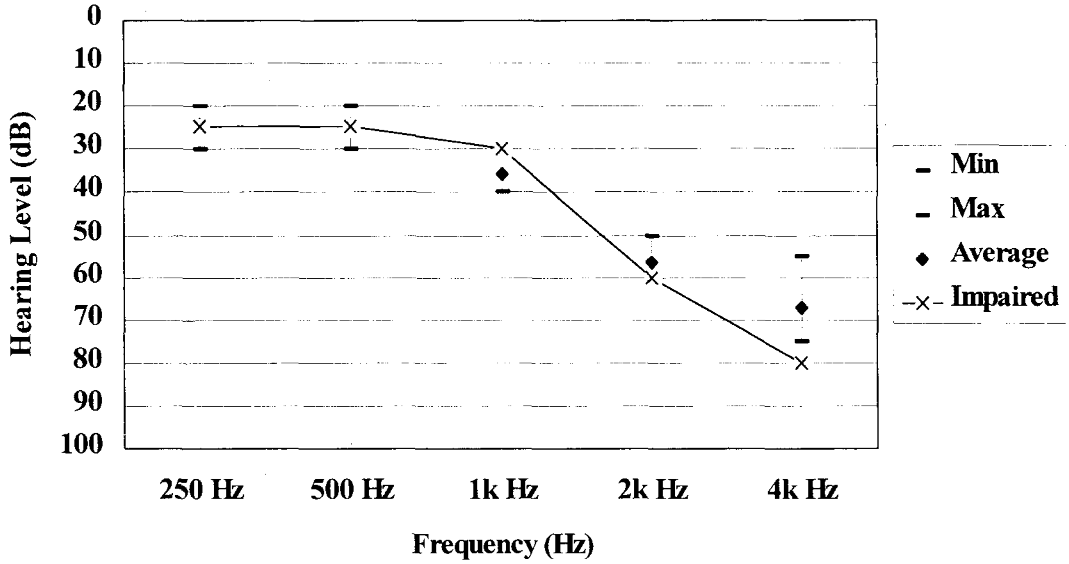

Fig. 8. Pure tone audiogram of impaired listener and average, minimal(Min), and maximum(Max) values of 7 HIS-processed subjects

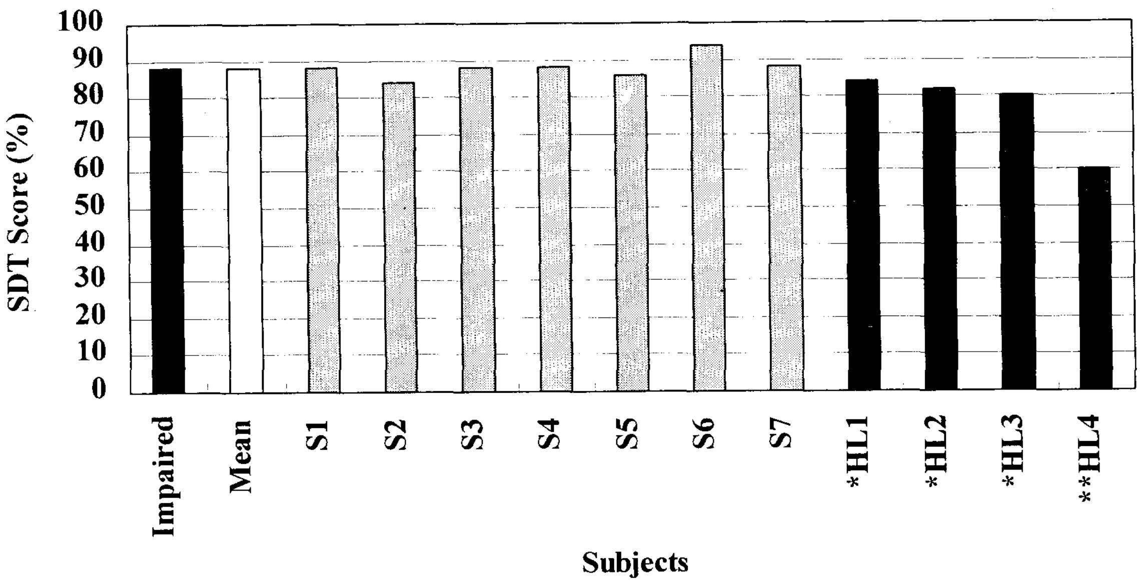

Fig. 9. SDT scores of impaired listener and HIS-processed listeners. The subjects are composed of 7 normal listeners and 4 impaired listeners with C_5 dip(* : 30-45 dB TH at 4kHz), and a high tone sensorineural hearing loss(* * : 10dB TH at 4kHz, 55dB TH at 8kHz). Mean : 88±3% (S1 to S7)

'Mean' indicate an SDT score of our subjects. The SDT scores again show that there exists a close resemblance among the target model and the simulated ones. Most of the subjects showed the SDT scores comparable to the discrimination score of the target impaired listener. It should be noted that the four subjects with high frequency hearing loss (who are marked by * and * * in Fig. 9) showed much lower the SDT scores. From the results shown in Fig. 6 to 9, it could be said that the present HIS system was successful in simulating sensorineural hearing impairment using the normal listeners.

C. Evaluation of Digital Hearing Aid Algorithms Using the HIS System

To evaluate the digital hearing aid algorithms using the HIS system the SDT score was measured. Fig. 11 and Fig. 12 shows the SDT results for simulation of two hearing impairments. The higher the score the better fitting was assumed. The simulated hearing

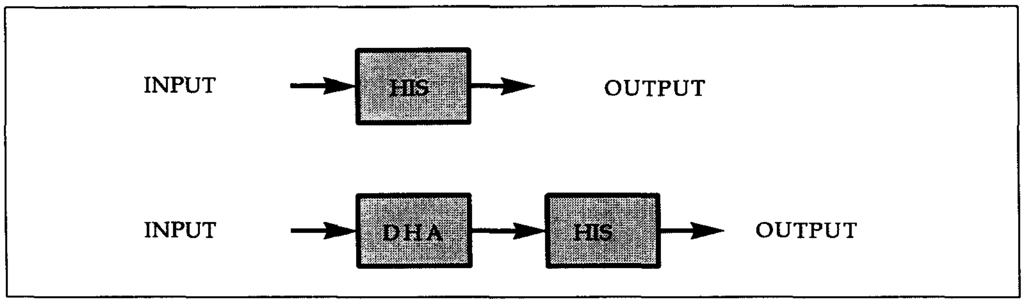

Fig. 10. Evaluation of the digital hearing aid algorithms using HIS system
[CASE 1] Hearing Impairment Simulation [CASE 2] Digital hearing aid algorithms + HIS

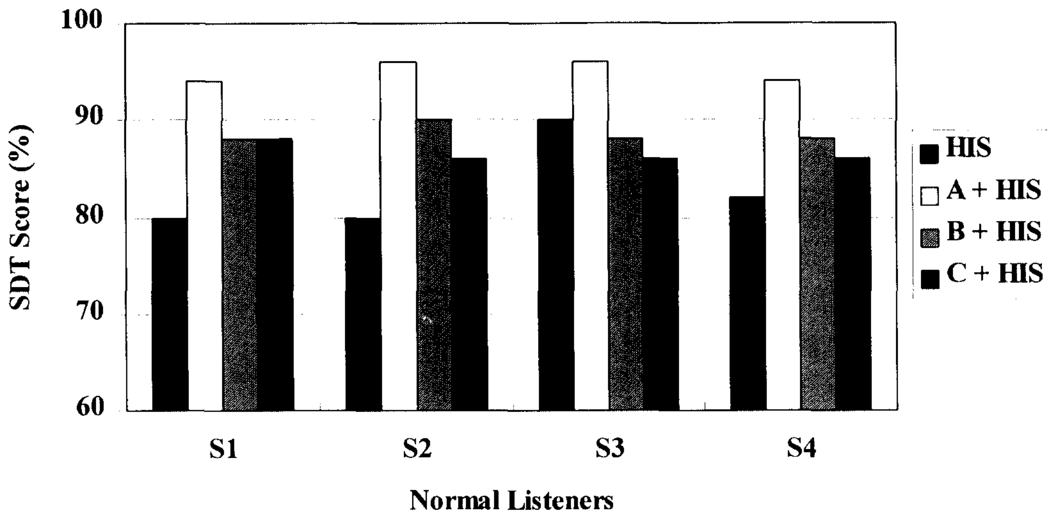

Fig. 11. SDT scores of the HIS system(Modeling in Fig. 7) with the digital hearing aid algorithms, A : Nonlinear WDRC using a single filter algorithm, B : 5-channel amplitude compression filterbank method, C : 5-channel linear gain filterbank method

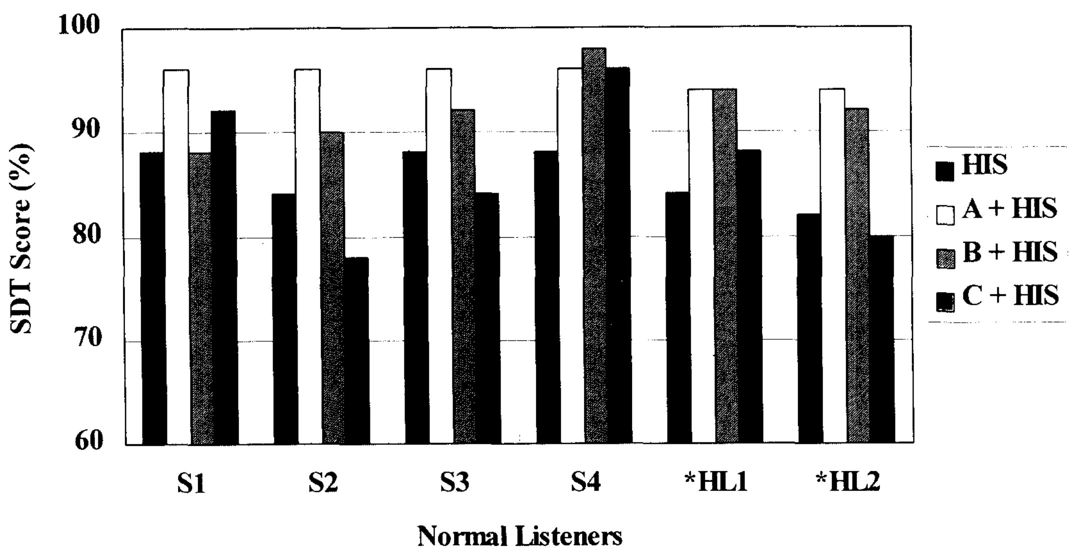

Fig. 12. SDT scores of the HIS system(Modeling in Fig. 9) with the digital hearing aid algorithms, A : Nonlinear WDRC using a single filter algorithm, B : 5-channel amplitude compression filterbank method (FIR filters), C : 5-channel linear gain filterbank method(IIR filters)

data of this evaluation test were the same as ones used for simulating the sensorineural hearing impairments. Among the three digital hearing aid algorithms the nonlinear wide dynamic range compression algorithm (A in Fig. 11 and 12) showed the best score in terms of the ability to understand speech.

CONCLUSION

A hearing impairment simulation tool was developed and its performance was evaluated using a real-time system. To modify the natural input sound into an impaired one, a frequency sampling filter was designed, whose frequency response is continuously adjusted according to hearing loss functions. Because the selection of the gains for the FIR filter design was controlled by the energy values in 20 critical bands, the signal is nonlinearly processed as a function of its frequency and intensity. The experimental results of the HIS system showed that the HIS system implemented the hearing loss model with a tolerate margin of error. Furthermore, subjective tests conducted with normal subjects confirmed the effectiveness of the HIS system in simulating the sensorineural hearing impairment, which was clearly indicated in the measurement results of both the thresholds of hearing and the SDT.

So far, we have shown that the HIS system developed in this study can simulate sensorineural hearing impairment and it is quite worthy in using the system for providing the feeling of the hearing impairment to normal subjects. Moreover, the system can be used for more practical issues such as evaluating conventional hearing aids and developing new fitting methods for the digital hearing aids. More work regarding these issues is currently under way.

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