EVALUATION OF A HEARING COMPENSATION ALGORITHM

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ABSTRACT

A new hearing compensation algorithm based on a homomorphic multiplicative AGC (automatic gain control) is evaluated and compared against commercially available digitally programmable analog hearing aids. Both quantitative (speech recognition threshold and speech discrimination) and qualitative tests (estimation of perceived quality) were used in the evaluation. The new algorithm is shown to have made significant progress in restoring normal or near normal hearing for hearing impaired individuals.

1. Hearing Compensation

Hearing impairment may occur for any of a number of reasons including noise exposure, old age, disease, and trauma; or an individual may possess a congenital hearing disorder. The types of hearing impairment that result are also quite varied and include damaged inner and/or outer hairs cells within the cochlea, damaged or missing middle ear ossicles or tympanic membrane, and defective central neural pathways. The methods used in compensating for hearing loss, on the other hand, are few and have usually involved only simple amplification, or amplification with high frequency emphasis. Recently, some hearing aids have begun using loudness compression to more accurately compensate for some hearing losses.

A hearing compensation algorithm is evaluated which provides an accurate way to compensate for several types of hearing losses. The new hearing compensation algorithm is based on a homomorphic AGC (automatic gain control). It filters sound into multiple bands with an approximate bandwidth of one–third octave each. The sound in each of these bands is modified by an AGC of the form shown in figure 1. The outputs of all the AGCs are then recombined by simple summation. This approach shows significant performance gains based on intelligibility scores, SRT's (which are explained later), and also restores intelligibily in noise corrupted environments to near normal performance levels. The hearing aid is based on a mathematical model which describes fundamental processed in the ear and it provides precise multichannel loudness compression tailored the an individual's hearing loss [1].

The hearing compensation algorithm was implemented on a real-time DSP board with a 16 bit ADC and DAC and a sample rate of 21.333 kHz. The sound was filtered into 12 one-third octave bands, the loudness of each band was compressed in such a way as to greatly increase the intensity of soft sounds but not loud sounds, and the bands were each recombined and the sound was played through a standard hearing aid receiver (speaker). This setup is hereafter referred to as the "new aid." The "new aid" is

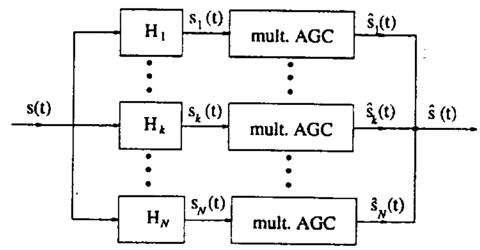

Figure 1: An N channel hearing compensation system. Each channel possessing a "critical bandwidth" bandpass filter, $H_k(f)$; a homomorphic multiplicative AGC with an exponentiation factor K_k determined for that band; and a sum of the AGC outputs.

compared against two other hearing aids that were deemed to be the best digitally controlled analog behind–the–ear aids commercially available. These two hearing aids are hereafter referred to as "aid A" and "aid B."

2. Evaluation Procedures

For each of eight subjects a routine audiological assessment was performed to determine type and severity of hearing impairment. Pure–tone hearing thresholds were used as a measure of the hearing impairment for each subject and bone conduction hearing thresholds were used to provide a measure of how much of the impairment was conductive (related to the transmission of sound to the inner ear), and how much of the impairment is sensori–neural (related to the transduction of the sound pressure wave into neural pulses and the subsequent transmission of those pulses). The eight subjects had a variety of types of hearing impairments including congenital, conductive, presbycusic, sensori–neural, and mixed conductive/sensori–neural.

The hearing aids were custom fit for each of the subjects. Fitting for "aid A" and "aid B" was performed by an audiologist trained by the respective companies in fitting both aids. Following the fitting of the hearing aids the subjects were placed in a sound booth and asked to listen to conversational speech. The hearing aids were adjusted as necessary to yield a comfortable setting for conversational speech.

No standards exist for evaluating the general performance of hearing aids. Tests were chosen which evaluate the hearing aid's ability to enhance intelligibility of speech in both quiet and noise. Another very important measure of how well a hearing aid performs it how well the wearer likes the aid. Therefore, each of the subjects was asked to rate the "quality" of the output sound for each of the aids when listening to music, speech, and silence (to check for internal noise). Each of the tests were performed first unaided and

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then with each of the three hearing aids chosen in random order. For each subject only one ear was evaluated,[1] the other ear was plugged using EAR plugs. The hearing tests are described below.

Speech Reception Threshold (SRT) was determined.[2] The SRT is defined as the lowest level at which the subject can correctly repeat two syllable words 50% of the time. The level is measured in dB HL; 0 dB HL is the approximate threshold of hearing for an average individual with normal hearing sensitivity.

Speech discrimination in quiet was measured. For this test the subject is asked to repeat monosyllabic words presented to the subject at various intensities. The number of words correctly repeated is counted or "scored." Each of the words used has three phonemes or sounds—for example, "cat" has three phonemes which are the sounds associated with "c," "a," and "t" respectively. In addition to number of words correctly repeated, the number of phonemes correctly repeated is counted. If the word is "cat" and the subject says "sat," then two out of the three phonemes are counted as correct. Eight isophonemic word lists developed at the Mayo Clinic and produced at Brigham Young University were used. Each word list contains ten words with three phonemes per word for a total of 30 phonemes. The word lists were randomized presented to the subjects at 10, 20, 30, and 40 dB HL. The percentage of the words and phonemes in correctly repeated in each list was calculated at each presentation level and for each listening condition.

Speech discrimination in multitalker babble was measured using a procedure identical to that for speech discrimination in quiet except that twelve talker babble[3] was presented at 40 dB HL and the word lists were presented at 40, 45, 50, and 60 dB HL (yielding 0, 5, 10, and 20 dB signal to noise ratios respectively).

Speech discrimination in speech spectrum noise was measured using the same procedure as for speech discrimination in speech babble noise except that speech spectrum noise at 40 dB HL was used in place of babble noise.

3. Clinical Test Results

Test results of the homomorphic multiplicative AGC algorithm for hearing impaired subjects provide the following results.

- The SRT improved over the unaided case for every aid. However, the average SRT with the homomorphic AGC hearing compensation algorithm was within "normal" limits[4] for an unimpaired listener. See figure 2.

[1]Testing was performed at the Brigham Young University Audiology Clinic under the direction of Dr. Richard Harris.

[2]SRT was determined using the 2 dB step procedure recommended by the American Speech–Language–Hearing Association.

[3]The babble consisted of 12 talkers speaking simultaneously and continuously.

[4]An SRT of up to 10 dB HL is usually considered in the normal range.

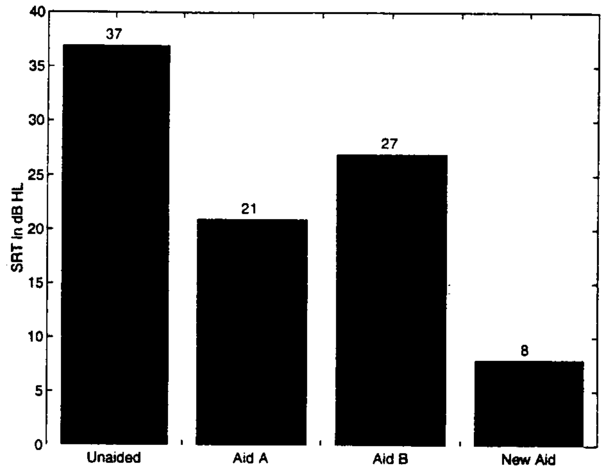

Figure 2: Comparison of the average SRT score between the new algorithm being tested and two other hearing aids labeled "A" and "B."

Speech Discrimination in Quiet Phonemic % Correct				
dB HL	Unaided	Aid A	Aid B	New Aid
10	0	4	0	14
20	0	17	9	71
30	12	58	32	95
40	33	90	66	97

Table 1: Comparison of phonemic scoring % in quiet. Table entries for each of the aided and the unaided conditions represent the average percentage of phonemes which were correctly identified by the subjects. Table entries for each of the aided and the unaided conditions represent the average percentage of phonemes which were correctly identified by the subjects.

Speech Discrimination in Multi-talker Babble Phonemic % Correct				
dB SNR	Unaided	Aid A	Aid B	New Aid
0	0	39	17	65
5	19	71	64	86
10	41	90	82	96
20	77	96	93	94

Table 2: Comparison of phonemic scoring % in speech babble noise presented at 40 dB HL in addition to the word lists. Words were presented at 40, 45, 50, and 60 dB HL for signal to noise ratios (SNRs) of 0, 5, 10, and 20 dB respectively. Table entries for each of the aided and the unaided conditions represent the average percentage of phonemes which were correctly identified by the subjects.

Speech Discrimination in Speech Spectrum Noise Phonemic % Correct				
dB SNR	Unaided	Aid A	Aid B	New Aid
0	7	9	12	17
5	18	45	48	61
10	43	73	69	75
20	72	91	95	93

Table 3: Comparison of phonemic scoring % in speech spectrum noise presented at 40 dB HL in addition to the word lists. Words were presented at 40, 45, 50, and 60 dB HL for signal to noise ratios (SNRs) of 0, 5, 10, and 20 dB respectively. Table entries for each of the aided and the unaided conditions represent the average percentage of phonemes which were correctly identified by the subjects.

Qualitative Scores			
Stimulus	Aid A	Aid B	New Aid
Speech	9.1	9.4	9.9
Music	8.9	8.6	9.1
Quiet	9.5	9.5	9.3

Table 4: For stimuli of speech, music, and silence, subjects were asked to rate the quality on a scale of 1 to 10 with 10 being the highest.

- Average performance in quiet for the hearing aid was also near normal and significantly better than both the other aids at nearly all levels.[5] See table 1.
- The performance of the new aid in speech babble noise was again near normal hearing and clearly superior to the other aids at 0 dB SNR and 5 dB SNR while performing only slightly better than or the same as the other aids at higher SNRs. All of the aids performed fairly well at the highest SNR (20 dB SNR with a speech level of 60 dB HL). Refer to table 2.
- In speech spectrum noise the new aid all aids performed about the same in restoring hearing. The scores for all aids were not far below the performance of normal hearing subjects for this type of noise. See table 3.
- For each of three types of stimuli, speech, music, and quiet, the subjects rated the subjective quality of each of the three aids on a scale of 1 to 10, with 10 being the highest. The "new aid" was preferred over both "aid A" and "aid B" for speech and music. The "new aid" had an average rating for quiet which was slightly lower than the others.[6] See table 4.

  Two of the subjects reported that the "new aid," based on the homomorphic AGC algorithm, was the best that they have ever heard. None of the subjects offered a similar report about "aid A" or "aid B." One of the subjects reported that speech was the "Best I can imagine it" and reported that music was "The best I've ever heard .... Really terrific, truly the finest of the three. Noticeably better."

[5] Differences of more than 10% should be considered significant while differences of less than 10% should be considered within expected test/retest variation.

[6] This was primarily due to one very low score given because of 60 cycle hum introduced at the microphone preamp during that subject's tests.

4. Discussion of Results

There are several key features which cause the new aid to perform well.

1. The implementation of the loudness compression is not based on a feedback type automatic gain control (AGC) but on a multiplicative homomorphic method which compensates more precisely for several types of hearing loss.
2. The loudness compression helps to squelch feedback so that higher gains may be applied for low level signals. Feedback is one of the major problems today in fitting hearing aids. Inherent feedback suppression is a feature of the homomorphic AGC algorithm.
3. Since the new aid utilizes so many channels (12 in this implementation) the compensation may be more precise and loud sounds in one frequency band do not cause a gain reduction in other bands. In this way the new aid performs similarly to masking in the human auditory system.

5. References

[1] D. C. Chabries, D. V. Anderson, T. G. Stockham, Jr., and R. W. Christiansen. Application of a human auditory model to loudness perception and hearing compensation. *ICASSP-95*, 1995.