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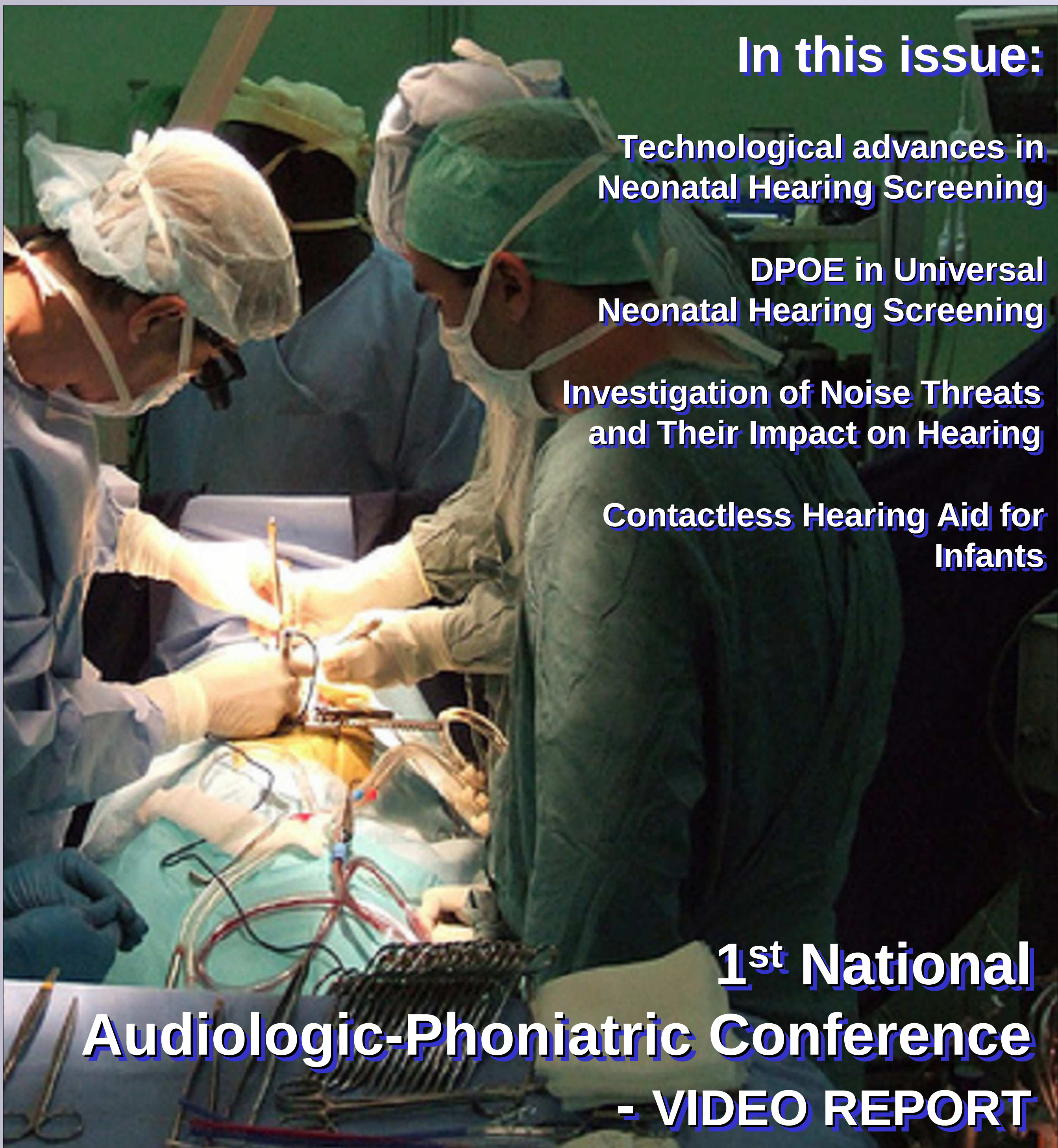

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Despite the fast development of telemedicine as scientific and practical discipline it is still considered a novelty in the global scale. That is because telemedicine is strictly interdisciplinary in character, combining medicine, computer science, telecommunications, multimedia technology, biomedical engineering, psychology, electronics as well as certain branches of physics (e.g. acoustics and optics) and mathematics (data analysis techniques).

We extend an invitation to all interested specialists in the above domains of medicine, science and technology. This Journal provides a forum for exchanging ideas between researchers, medical doctors and engineers, and will serve again future developments of international co-operation in the domains of: general telemedicine, telepathology, distance learning and advanced multimedia communication services. That is necessary because the question still stands: how should health care organisations, technologies and research respond in order to maximise the positive potentials of citizens participation in eHealth technologies? This question will be answered principally by those engaged in the process of developing telemedicine and telecommunication technologies who will be willing to contribute to our Journal of the International Telemedicine Academy.

Editorial Staff

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Designed by P. Szczuko

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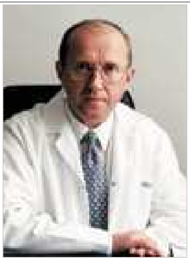

Henryk SKARŻYNSKI
Prof., M.D., Ph.D.

Institute of Physiology and Pathology of Hearing
Kajetany

Editorial

Ladies and Gentlemen,
Dear Friends,

On behalf of the Organizing Committee I would like to thank all guests and participants of the 1st Audiologic-Phoniatric Conference in Warsaw for their participation.

The Boards of Audiologic and Phoniatric Sections of the Polish Society of Otorhinolaryngologists and Head and Neck Surgeons took a decision on joint scientific meeting as a natural consequence of the fact that a basic specialty – Audiology and Phoniatry came into being several years ago.

The team of the Institute of Physiology and Pathology of Hearing has been honored to organize the first general scientific conference in this area, to which over 110 papers had been submitted.

During the conference, there had been held plenary sessions assisted with a group of 20 eminent and respected scientists, Polish and foreign, representing various areas of audiology and phoniatry, accompanied by lecture, paper, poster and satellite sessions, as well as workshops.

I am convinced that in the rich program of the conference you had found interesting issues, and it was also a great opportunity to discuss the recent personal achievements. I hope that the joint meeting was a great opportunity to exchange information on the recent achievements in both medical fields, as well as a chance of getting to know each other better.

I would like to express my gratitude to the main co-organizers of the Conference and to the medical companies for participating in that event.

On behalf of the Organizing Committee
Prof. Henryk Skarżyński, M.D., Ph.D.

Technological advances in Neonatal Hearing Screening

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Key words: Automated Otoacoustic Emissions, AOAE, Automated Auditory Brainstem response. AABR, Neonatal Hearing Screening, Auditory State Steady Response, hearing threshold

1. Introduction

In 2006 Otoacoustic Emissions (OAEs) celebrate a life span of 28 years (after the first OAE publication by David Kemp in 1978). The most significant contribution of OAEs is in the area of Neonatal Hearing Screening (NHS). Within the last decade, numerous new objectives have been presented in the NHS area such as the quality of the automated OAE responses, the estimation of a hearing threshold etc. To respond to the clinical demands of these objectives, several new methodologies have been introduced in the clinical practice the last few years and the aim of this technical note is to provide information on these latest technological trends.

2. Automated Auditory Brainstem Responses

In the early 2002, the first 4rth generation OAE devices appeared in the market and provided the possibility to numerous clinical realities to integrate information from automated OAE (A-OAE) and automated ABR (A-ABR) recordings. The combined screening protocols (A-OAE + A-ABR) targeted the identification of auditory neuropathy cases most prevalent in the NICU environment. Nevertheless, the presence of portable ABR equipment provided the possibility to conduct studies in real screening environments (and not in various simulations in ideal ambient conditions) where the hearing threshold was assessed with both portable and clinical equipment. A pilot study conducted by our group (Giorba et al, 2006) in the context of a regional project in Emilia-Romagna (Project CHEAP) have suggested that the portable ABR and OAE technologies are converging in terms of time requirements. The data collected in the above study has suggested : (i) the average time for a AOAE responses is clearly less than 10 s in a cooperative subject, and less that 120 s (2 min) in non-cooperative subjects. (ii) test times of A-ABR in cooperative subjects were less than 120 s, while uncooperative subjects were tested within 10 min (per ear). While it takes some minimum expertise to properly handle and position the OAE probe, the ABR electrode placement presents more complications especially in cases where the subject shows high electrode impedance. In the latter case the AABR testing is difficult to complete and the test times are unavoidably longer.

Theoretically a 2-stage approach (ie A-OAE + A-ABR) eliminates the risk of not identifying infants with Auditory Neuropathy and assures that the screening sensitivity is high . Contrary to these hypotheses recent data from an American study (White et al, 2005) suggest that this is not the case. The study assessed information from 86634 infants and for the infants who were screened for hearing loss using a typical 2-stage OAE/A-ABR protocol, approximately 23% of those with permanent hearing loss at 8–12 months of age would have passed the A-ABR. The data suggest that stringent criteria should be incorporated in the final evaluation of the current OAE and ABR automated devices.

3. Auditory Steady State Responses in Neonatal Screening

Both OAE and ABR technologies utilize as stimuli electrical clicks and the acquired information is clearly more related to the audiometric frequencies of 1.0 and 2.0 kHz. Within this context, there has been a speculation of whether other measurements technologies could be used in a fast hearing assessment of neonates, children and adults. A group of similar electrophysiological measurements to OAEs and AABR includes electro-cochleography (EcoG), and Middle latency (ML) and Steady State Responses (SSR). From this group the latter category has shown interesting characteristics due to fact that by alternating the modulation frequency (i.e. increasing it) of the stimuli one can get responses from the Auditory cortex (low modulation frequencies around 40 Hz) or from the Brainstem (Cone-Wesson et al; 2002; Dimitrijevic et al , 2002: John and Picton, 2002). The SSR protocol has already passed to an automated one (ASSR) and for the last two years numerous publications have been devoted to the threshold estimation via the ASSR technique. The ASSR protocols have been greatly optimized, (Gorga et al, 2004) and the SSR responses are detected in the frequency domain by robust probabilistic algorithms.

In 2002 Conne-Wesson et al proposed the use of ASSR as a hearing screening tool, with the objective that ASSR could substitute the A-ABR. A few reports have been available since (Stueve and O'Rourke, 2003; Luts et al, 2004; Swanepoel et al; 2004) indicating a good agreement between ASSR and A-ABR at 2.0 kHz and various differences at 0.5, 1.0 and 4.0 kHz. Most studies recommended the use of the SSR technique in the clinic but the point of substituting the A-ABR with ASSR is not supported yet by the available data. The factors which affect the A-ABR (ambient noise and electrode impedance) interfere with the ASSR recordings as well. In order to resolve these issues Vivosonic has presented a new line of devices using preamplifiers at the level of the scalp-electrodes (called amplitrodes) which suppress the level of ambient noise and provide very clean A-ABR and ASSR traces. It is to be seen how these electrodes will be intergraded in the normal clinical reality since the pre-amplifiers require electrical energy which translates into changing batteries every x tests. In the context of neonatal screening, an ASSR screening protocol might target initially a few frequency points (ie 1.0 & 2.0 kHz or 2.0 & 4.0 kHz) which show immunity to ambient noise (Figure 1 & Figure 2) . Nevertheless the ASSR protocol requires significant optimizations before becoming a member of the neonatal hearing screening battery of tests.

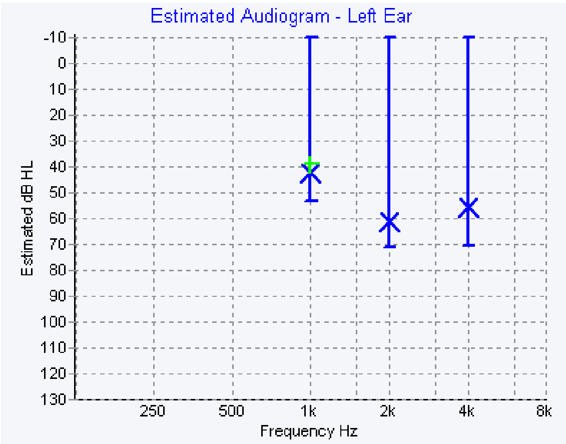

Figure 1. ASSR response from a well baby who was crying using the AUDERA device from VIASYS. The lowest tested frequency was not available due to noise. The length of the testing procedure was 22 min (14 min longer than the successfully completed A-ABR test). Despite the theoretical noise immunity at 2.0 and 4.0 kHz the size of the error bars indicate that the measurements are too variable to be considered. The “x” symbols indicate the mean threshold level of the measurements.

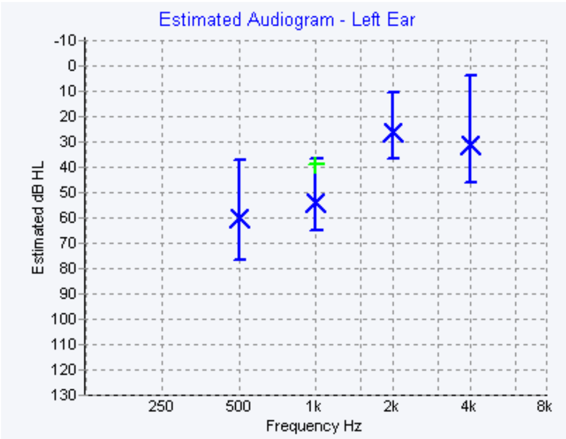

Figure 2. ASSR response from a well baby using the AUDERA device. The length of the test was also longer that the A-ABR (16 vs 7 min). The A-ABR suggested a REFER probably due to conductive complications suggested by the AASR outcome. In this case the 2.0 and 4.0 kHz frequencies show good noise immunity (suggested by the small size of the error bars). The “x” symbols indicate the mean threshold level of the measurements.

4. Threshold estimation via DPOAE measurements

An interesting challenge for otoacoustic emissions has been the relationship between the amplitude of the OAE response and the hearing threshold (Whitehead et al 1995a; 1995b; Shera et al, 1999). For cases where no conductive losses are present there is a good agreement between OAEs and the hearing threshold. In such cases Input-Output distortion product OAE (DPOAE) protocols may offer more information (Whitehead et al, 1995a, Janssen et al, 1998; Dorn et al, 2001; Gorga et al, 2003b). Besides the relationship to pure-tone thresholds, DPOAE I/O-functions provide an estimate of the compression related to the outer hair cell amplifier. Data supporting this hypothesis are available from animal studies where the hearing of the animals was impaired with acute furosemide intoxication (Mills and Rubel, 1996) and human studies with subjects suffering from cochlear hearing loss (Janssen et al., 1998; Kummer et al., 1998; Boege and Janssen, 2002; Neely et al., 2003). In these studies the slope of the DPOAE I/O-function increased with increasing hearing loss revealing a loss of compression of the outer hair cell amplifiers. In this context by using numerous combinations of I/O DPOAE recordings one can obtain very precise information related to the status of the cochlear amplifier (Gorga et al, 2003a, 2003b). Recently, extrapolated DPOAE I/O-functions were constructed from neonates to estimate pure-tone threshold levels and the corresponding cochlear compression values (Janssen et al., 2003). The estimated hearing threshold was found to be increasing within the early postnatal period (average age: 3 days), predominantly at the higher frequencies, and to be normalized in a follow-up measurement (after four weeks). However, the slope of the DPOAE I/O-functions obtained in the first and second measurement was unchanged revealing normal cochlear compression. Consequently, these findings were interpreted as temporary conductive hearing losses due to the presence of amniotic fluid and/or Eustachian tube dysfunction. In this clinical scenario, especially during the first days of life, a hearing screening test may lead to false positive results due to a temporary conductive hearing loss. The use of the slope of DPOAE I/O-functions could be used as an index of conductive losses which might result in less false positives an in less time spent for audiological clinical diagnostics. According to the data of Jenssen et al (2003) the values of the DPOAE slope can discriminate and differentiate conductive from sensorineural hearing losses. In addition DPOAE I/O-functions have been reported to be correlated with loudness (Neely et al. 2003), so DPOAE I/O information would also offer the potentiality of assessing information to basic hearing aid fitting.

The research findings from Janssen et al (2003) and Gorga et al (2003a) have been commercialized in a device called Cochlea- Scan (Osvald et al, 2003) by Fischer-Zoth. Hearing threshold can be extrapolated up to values relative to 50 dB HL in the frequency range from 1.5 to 6 kHz. Figure 3 shows two data acquisition sequences. At present the Cochlea-Scan device offers a platform for a third generation OAE testing (TEOAEs, DPOAEs), I/O DPOAE estimation with hearing threshold extrapolation and Pure Tone Audiometry measurements.

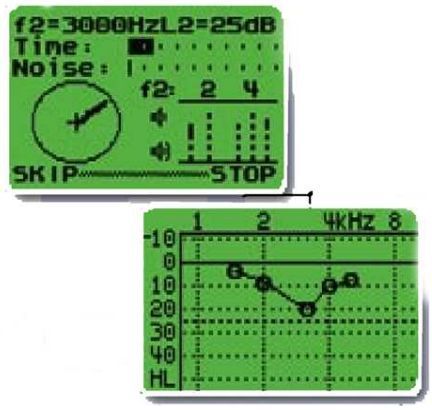

Figure 3. Cochlea-scan displays during the threshold estimation (top panel) and threshold extrapolation (bottom panel) from a well baby. The audiometric notch between 2 .0 to 4.0 kHz is probably enhanced by the standing waves in the external auditory meatus which influence the values of the employed DPOAE measurements.

Acknowledgements

The author would like to thank Maddalena Rossi and Giorgio Rossi for technical assistance with the CochleoScan and AUDERA equipment.

Appendix

The reader interested in additional information than the one presented might visit the OAE Portal (http://www.otoemissions.org) and the OAE Portal Forum.

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The use of Distortion Product Otoacoustic Emissions in a Universal Neonatal Hearing Screening (UNHS)program

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Abstract: We have conducted a comparison of three DPOAE protocols, testing cubic $2F_1$ - F_2 distortion products, in order to define the most feasible protocol for a universal hearing screening program. The protocols used asymmetrical stimulus intensities (L1 > L2) with a frequency ratio of 1.22, in the following format: : (P1), L_1= 60, L_2= 50 dB SPL; (P2), L_1= 65, L_2= 55 dB SPL; and (P3), L_1= 75, L_2= 65 dB SPL. Linear TEOAE responses evoked by click stimuli of 75 dB p.e. SPL were used as controls of the normal cochlear function. Five $2F_1$ - F_2 frequencies (ref. to F_2) 1.5, 2.0, 3.0, 4.0, 5.0 kHz were tested with a ILO-92 macro subroutine. The project included randomly selected recordings from 1200 well-baby nursery (WBN) infants (age 48 hr) and 50 very low birth weight NICU infants. Statistical analyses comparing the signal to noise ratios (S/N), at the predefined F_2 frequencies, indicated that the P1 and P2 DPOAE protocols perform similarly. Significant S/N differences were observed in the P3 to P2 and P3 to P1 data-set comparisons. DPOAE scoring criteria were estimated from the P3 data-set, using one-sided distribution-free tolerance boundaries. The scoring criteria for a "pass" were estimated as a minimum S/N of 6.0, 7.0 and 6.0 dB at 2.0, 3.0 and 4.0 kHz respectively. In terms of feasibility, the P3 protocol generated responses in 98% of the WBN and 76% of the NICU infants. All three DPOAE protocols demonstrated smaller time-recording requirements than the TEOAE standard. The false-positive ratio for the NICU infants was estimated as 8%.

Key words: OAEs, transient otoacoustic emissions, distortion product otoacoustic emissions, TEOAEs, DPOAEs, linear TEOAE protocol, well babies, NICU infants, universal hearing neonatal screening.

1. Introduction

The etiology of deafness arising during childhood age is various and often unknown. Nevertheless the prevalence of hearing impairment among the children is high (5/1000) (Brackett et al, 1993). Severe genetic or congenital hearing loss is represented in about 1-2/1000 of well-babies (White et al, 1993) and in 4-5% (Mason and Herrmann, 1998; White et al, 1995) of newborns which exhibit one or more audiological risk factor (White et al, 1993; Parving, 1984;Joint Committee on Infant Hearing Screening, 1990). It has been shown that approximately 50% of children identified with sensorineural hearing loss (SNHL) do not exhibit any risk factors at birth (Mauk and Beherens, 1991; Vohr and Maxon, 1996). Within this context, a hearing screening focused only on the “at risk group” will detect no more than half of the deaf children. To increase the detection of hearing loss at the youngest possible age , it is necessary to implement an optimized universal hearing screening program, testing both well baby nursery (WBN) and neonatal intensive care unit (NICU) infants.

The leading technology for neonatal hearing screening is currently based on otoacoustic emission protocols. Transiently evoked otoacoustic emissions (TEOAEs) are the choice of the majority of European and US screening programs (Mencher et al, 2001; Welzl-Muller and Stephan, 2001: Psarommatis et al, 2001; Torrico Roman et al, 2001; Zehnder et al, 2000;White et al, 1993; Norton et, 2000;Gravel et al, 2000). The TEOAE protocols have a high penetrance in the hearing screening studies, despite the fact that the distortion product otoacoustic emissions (DPOAEs) show a good frequency specificity (Brown and Kemp, 1984; Martin et al, 1987) and a very good noise-immunity (Nelson and Kimberly, 1992; Gorga et al; 1997; Hatzopoulos et al, 2001), in the 2- 4 kHz frequency range where most screening programs evaluate the hearing function .

The main goal of this study was to test the hypothesis of whether a DPOAE screening protocol can be used efficiently, or as efficiently as a TEOAE protocol, in a Universal Neonatal Hearing Screening program (UNHS). To attain this objective we have: (1) used a large sample size of WBN infants; (2) a sample of NICU infants (very low birth weight cases); and (3) we have tested three different DPOAE protocols (using a preset number of five tested frequencies) in order to find the DPOAE protocol offering the best quality of responses (high signal to noise ratios) and the best test-feasibility. Due to time restrictions in the NICU environment, the DPOAE protocol-comparison and scoring-criteria estimation was performed only on the data from the WBN infants. The best DPOAE protocol was applied on the data from the NICU infants to attain a test-feasibility estimate.

2. Materials and Methods

The WBN screening program at Ferrara University (and in the region of Emilia-Romagna) uses a three-stage protocol. To test WBN infants, TEOAEs are used in stages 1 and 2 (re-tests are repeated within 14 days from birth). An Auditory Brainstem Response (ABR) diagnostic evaluation is used in stage 3 (within 3 months from birth). Neonates without reproducible TEOAEs (see scoring criteria in the protocols section) in either ear, after stage 2 are referred to stage 3. The objective of the program is to provide effective clinical assistance (intervention) to hearing impaired subjects within the first 6 months of life (Yoshinaga-Itano, 1995).

Different screening procedures are used for the Neonatal Intensive Care Unit infants. Although TEOAEs are routinely used in stages 1 and 2, the re-testing is conducted within a 3 day interval, for a maximum of 6 re-test sessions. For stage 3, as in the case of the WBN infants a diagnostic ABR evaluation is used.

2.1 Subjects

A group of 1200, randomly selected, neonates (mean age 40.5 ±1.8 weeks) was tested during the second day of life during natural sleep and after feeding, in a quiet room in the well-baby nursery . The neonates were considered normal if they had a gestational age > than 37 weeks, a birth weight appropriated for their gestational age (AGA) and an Apgar score between 7 to 10 (1rst and 5^{th} min). The normal cochlear function of each subject was evaluated by TEOAE scoring criteria (mentioned in the section below) and then the DPOAE responses were acquired.

A group of 50, randomly selected, NICU infants characterized by very low birth weight (mean age 33 weeks ± 2.3, mean weight 1.200 ± 250 g) was tested also with TEOAEs in a silent room to verify the presence of emissions and the normality of the cochlear function. The evaluation of these responses was based on scoring criteria derived from well-babies.

2.2 Equipment and employed protocols

The OAE recordings were collected with a portable ILO292 equipment running the ILO software version 5.60. In previous publications (Hatzopoulos et al, 1999;Hatzopoulos et al, 2000a;Hatzopoulos et al, 2000b) we have presented evidence suggesting that a linear TEOAE protocol outperforms its nonlinear counterparts (QuickScreen included) in terms of signal to noise (S/N) ratios, correlation and noise level values. The linear TEOAE recordings were elicited using stimuli of 72-75 dB p.e. SPL with an acceptable noise level set at 52.0 dB p.e. SPL. Each recording was the average of a minimum of 50 sweeps (the max allowable number was 80). The TEOAEs were post-windowed by a 3.5-12.5 ms windowing function and in order to suppress muscular and respiration artifacts, the low TEOAE frequencies were removed by setting the bandwidth of the ILO recording to 1.5-5.0 kHz. An ear was considered normal when the S/N ratios from the corresponding linear TEOAE response, at 2.0, 3.0 and 4.0 kHz, (scoring criteria) were higher or equal to 7.0, 10.0, and 9.0 dB respectively (Hatzopoulos et al, 1999) . For a "pass" condition both ears should have satisfied the TEOAE scoring criteria.

The DPOAEs were elicited, using three asymmetrical protocols ($L_1 > L_2$) which were named as follows: (P1), L_1= 60, L_2= 50 dB SPL; (P2), L_1= 65, L_2= 55 dB SPL; and (P3), L_1= 75, L_2= 65 dB SPL. The rationale for the choice of these protocols was the following: Regarding the issue of asymmetrical and not equal intensity protocols, data from animal studies have indicated that a 10-15 dB SPL difference in the DPOAE stimulus primaries generates the highest amplitude cubic DPOAE responses (Lonsbury-Martin et al, 1993; Whitehead et al, 1998). Although this advantage decreases above 65 dB SPL (Gaskill and Brown, 1990), we have used a high level asymmetrical protocol to be able to compare the S/N ratios of the three protocols at the same frequencies (symmetrical protocols are referenced to the geometric mean and not to F_2). Regarding the issue of the selected stimulus intensities : A number of studies have shown that at lower DPOAE stimulus intensities it is easier to assess possible cochlear insults (Mills et al, 1993; Quinonez, 1999; Shera et Guinan, 1999). For that purpose we have implemented in the present study the P1 protocol (60-50 dB SPL). It should be noted that a protocol presenting lower stimulus intensities (i.e 50-40 dB SPL) might have been more efficient in detecting cochlear insults, but considering the noise levels in the WBN and the NICU environment such a protocol would have produced low test-efficiency results. The protocol P2 is considered the default option of many DPOAE instruments, although there is no evidence in the literature supporting such a choice. Finally, the high stimulus intensity protocol P3 was designed to overcome the ambient and subject noise problems in the neonatal hearing testing areas. Similar high stimulus protocols have been suggested or implemented in previous studies (Norton et al, 2000; Hatzopoulos et al, 2001).

The DPOAE protocols used a frequency-ratio of 1.22. Five frequencies of the cubic distortion product $2F_1 – F_2$ were tested with an ILO macro routine at 1.5, 2.0, 3.0, 4.0, and 5.0 kHz (referenced to F_2). The advantage of using the ILO macro was that the program could collect data at the F_2 frequency with the worst S/N ratio when the S/N ratios at other tested frequencies were higher than 10 dB. At each frequency a minimum of 32 averages was collected, with an acceptable noise level set at –10 dB SPL. It should be noted that the tested frequency of 5 kHz is not a common audiometric frequency, but it was chosen over the frequency of 6.0 kHz, because the frequency response of the ILO probe above 5.0 kHz is not flat.

2.3 Statistical Analyses

The optimization of the best DPOAE protocol was conducted only with data from the WBN infants. The NICU environment does not favor the recording of long sequences of OAE responses and it was decided to optimize the DPOAE protocol with the WBN infants and then to apply it on the NICU population.

In terms of performance-optimization, we have considered as best the DPOAE protocol generating responses with the highest signal to noise ratio at 2.0, 3.0, 4.0 and 5.0 kHz. To evaluate which protocol generates the best S/N ratios Hotelling’s T^2 test was applied. This multivariate procedure compares all the DPOAE variables (creating a test-vector) between two given setups, i.e. between P1 and P2 or P2 and P3 . Additional details on this procedure can be found in a previous publication (Hatzopoulos et al, 1999). Due to time restrictions two DPOAE protocols and the standard linear TEOAE protocol were tested at a time . We have used 600 WBN infants to test the P1 and P2 and 600 infants to test the P3 and the P2 DPOAE protocols. To avoid any biasing errors the protocol order (which protocol was applied first) was randomized. For 46 subjects it was not possible to record TEOAE responses (14 agitated subjects) or responses from a second DPOAE protocol (32 subjects), therefore these cases were excluded from the study. The actual datasets used in the study refer to 581 cases for P1 and P2 comparisons and 573 cases for the P3-P1 protocol comparison.

The definition of the normative DPOAE scoring criteria was conducted by using a free distribution (Hatzopoulos et al, 1999; Hatzopoulos et al, 2000b). The reason for using a free distribution method is that the DPOAE variables (S/N ratios at 2.0, 3.0, 4,0 kHz) are not normally distributed. In this context, it should be noted that inferences about population means (as in the case of Hotelling’s test) tend to be robust, if the sample size is moderate to large since the Central Limit Theorem guarantees that the distribution of means is closer to normal than the distribution of the original data. The scoring criteria were estimated from the best DPOAE protocol according the results of Hotteling’s T^2 test. The scoring criteria provide us with a minimum estimate of normal performance, which is the lower tolerance bound of the estimated tolerance interval (for every tested frequency). The statistical premises of this calculation can be expressed with the following statement: we are 99% confident that the calculated tolerance interval (at " x" tested DOPAE frequency) contains at least 95% of the S/N ratio values of the entire WBN population. Prior to the estimation of the DPOAE scoring criteria the DPOAE responses were evaluated with an apriori criterion, according to which a response was considered "pass" when the S/N ratios at 2.0, 3.0 and 4.0 kHz were higher or equal to 3 dB and the DPOAE amplitude at each tested frequency was > 0 dB SPL.

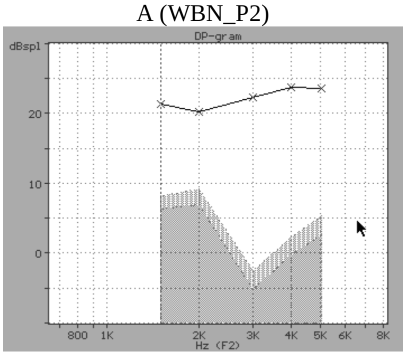
For all analysis a mainframe SAS statistical package was used.

3. Results

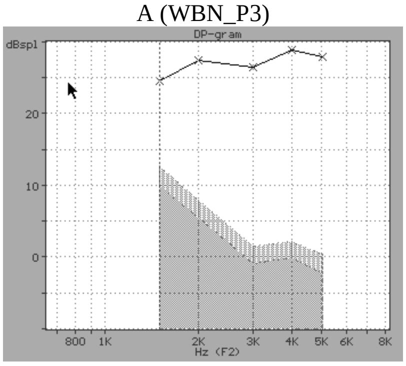
3.1 Description of the Data

Typical DPOAE responses from WBN and NICU infants, evaluated as "pass" are shown in Figure 1. The descriptive statistics from the TEOAE and DPOAE data-sets are presented in Table 1 (WBN) and Table 2 (NICU) . As it was expected the S/N ratios at the lower frequencies of 1.0 and 1.5 kHz show the highest variability, which is probably caused by respiration and muscular artifacts.

A (WBN_P2)



A (WBN_P3)



B (NICU_P3)

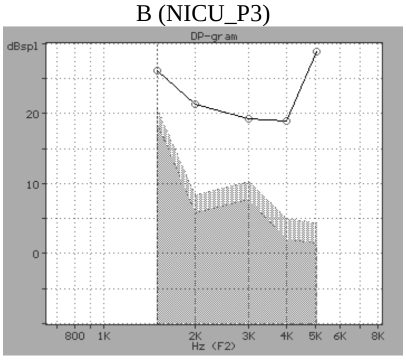


Figure 1. DP-gram data : (A) From a WMN infant (case 1322). The left panel shows the responses evoked by the P2 protocol (65-55 dB SPL) and the right panel the responses from the P3 protocol (75-65 dB SPL). In the latter there is a significant increase of the DPOAE amplitude level and a change in the shape of the noise level. By using a 75-65 protocol a DPOAE response can be clearly observed even in cases where the ambient noise is as high as 52 dB SPL. (B) a response from NICU infant (case P0054_007), elicited by the P3 protocol. The highest noise level was observed at the 1.5 kHz frequency, suggesting that the seal between the external meatus and the ILO neonatal probe was not optimized. In all three graphs the horizontal axis shows frequency in kHz and the vertical axis shows amplitude of the S/N ratio in dB.

Table 1. Descriptive statistics of the TEOAE and DPOAE variables from the 1200 WBN infants. The shaded cells indicate the beginning of a new protocol. The data variables presented are organized in the following order: TEOAE, DPOAE-P1, DPOAE-P2 and DPOAE-P3 protocols. The cases included in these 4 data-sets were evaluated as normal (normal cochlear function) by independent TEOAE and DPOAE criteria.

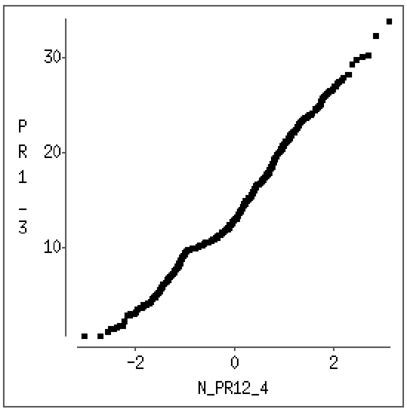
OAE Variables	Mean	Standard Deviation
TEOAE S/N 1.0 kHz (dB)	4.4	7.6
TEOAE S/N 2.0 kHz (dB)	16.9	4.9
TEOAE S/N 3.0 kHz (dB)	19.1	5.4
TEOAE S/N 4.0 kHz (dB)	19.1	5.9
TEOAE S/N 5.0 kHz (dB)	10.1	7.2
TEOAE Response (dB SPL)	21.6	5.1
TEOAE Noise (dB SPL)	3.9	2.2
TEOAE Time (sec)	54.3	35.3
DPOAE (P1) S/N 1.5 kHz (dB)	11.5	5.7
DPOAE (P1) S/N 2.0 kHz (dB)	14.8	5.8
DPOAE (P1) S/N 3.0. kHz (dB)	14.1	6.2
DPOAE (P1) S/N 4.0 kHz (dB)	17.5	7.3
DPOAE (P1) S/N 5.0 kHz (dB)	21.2	7.6
DPOAE (P1) Time (sec)	36.3	17.4
DPOAE (P2) S/N 1.5 kHz (dB)	11.6	6.4
DPOAE (P2) S/N 2.0 kHz (dB)	15.1	6.4
DPOAE (P2) S/N 3.0. kHz (dB)	14.9	6.4
DPOAE (P2) S/N 4.0 kHz (dB)	16.4	7.2
DPOAE (P2) S/N 5.0 kHz (dB)	21.2	7.8
DPOAE (P2) Time (sec)	35.3	15.9
DPOAE (P3) S/N 1.5 kHz (dB)	11.9	7.9
DPOAE (P3) S/N 2.0 kHz (dB)	17.9	8.1
DPOAE (P3) S/N 3.0. kHz (dB)	20.3	7.4
DPOAE (P3) S/N 4.0 kHz (dB)	20.8	8.4
DPOAE (P3) S/N 5.0 kHz (dB)	25.7	9.5
DPOAE (P3) Time (sec)	29.1	11.2

Table 2. Descriptive statistics of the TEOAE and DPOAE variables from the 50 NICU infants. The shaded cells indicate the beginning of a new protocol.

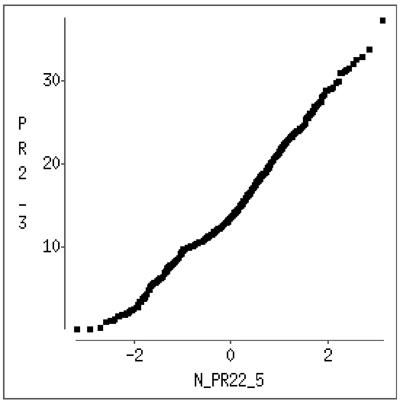
OAE Variables	Mean	Standard Deviation
TEOAE S/N 1.0 kHz (dB)	3.7	8.1
TEOAE S/N 2.0 kHz (dB)	9.8	6.9
TEOAE S/N 3.0 kHz (dB)	13.2	7.7
TEOAE S/N 4.0 kHz (dB)	13.5	7.5
TEOAE S/N 5.0 kHz (dB)	10.1	7.2
TEOAE Response (dB SPL)	16.9	8.1
TEOAE Noise (dB SPL)	5.5	3.6
DPOAE (P3) S/N 1.5 kHz (dB)	11.9	7.9
DPOAE (P3) S/N 2.0 kHz (dB)	12.8	8.0
DPOAE (P3) S/N 3.0. kHz (dB)	13.3	9.0
DPOAE (P3) S/N 4.0 kHz (dB)	14.6	8.3
DPOAE (P3) S/N 5.0 kHz (dB)	15.7	9.4

The distribution of the S/N ratios of DPOAE responses from the P1 and P2 protocols presented skewed patterns suggesting that the DPOAE variables were not normally distributed. The skewness was less pronounced in the P3 DPOAE data-set. Normalized quantile plots of various S/N ratios from the P1, P2 datasets presented evidence of bimodal distributions. Figure 2 shows normalized plots of the S/N ratio at 3.0 kHz from the P1, P2 and P3 datasets. The little "bump" in the graphs 2_P1 and 2_P2 of Figure 2 is an indication of a bimodal behavior.

P1



P2



P3

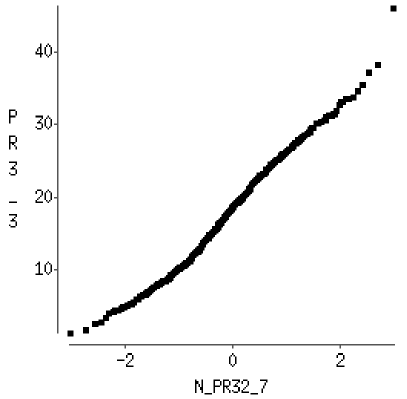


Figure 2. Normal quantile plots of the S/N DPOAE responses at 3.0 kHz elicited by P1, P2 and P3 DPOAE protocols. In the first two graphs a “bump” is shown, suggesting that the data have a bimodal distribution. The horizontal axis shows the normalized quantile values, while the vertical axis shows amplitude of the S/N ratio in dB.

3.2 Comparison of DPOAE protocols

The results from the protocol comparisons are shown in Table 3. The comparison between protocols P1 and P2 indicated that there were no significant S/N ratio differences at the frequencies 1.5 , 2.0 and 5.0 kHz. Significant differences were observed at the frequencies 3.0 and 4.0 kHz. The P1 protocol produced larger responses at 3.0 kHz while the P2 protocol produced larger responses at 4.0 kHz. Considering that the responses from these two tested protocols were quite similar, only an additional comparison was conducted (P3 vs. P2). For the latter, the data indicated that for all tested frequencies the S/N ratio differences were significant . Since the DPOAE responses from the P3 protocol presented the largest S/N ratios, the P3 protocol was considered as the one generating the best quality of DPOAE responses and the best candidate for a UNHS program.

Table 3. Results from the DPOAE protocol comparison using Hotelling's T2 test . The second column shows the differences between protocols P1 and P2 and the third column shows differences between protocols P3 and P2. Since the responses from P1 were found very similar to the responses of P2 no comparison between P3 and P1 was conducted.

DPOAE Variables	P1 vs. P2 CI= 99%	P3 Vs P2 CI= 99%
S/N 1.5 kHz	Ns	✡
S/N 2.0 kHz	Ns	✡
S/N 3.0 kHz	✡ (better P1)	✡
S/N 4.0 kHz	✡ (better P2)	✡
S/N 5.0 kHz	Ns	✡
P1 = 60-50 dB SPL, P2= 65-55 dB SPL, P3= 75-65 dB SPL ✡ = significant differences ns = no significant differences CI= Confidence Interval		

3.3 Estimation of scoring criteria

With the strict application of the apriori 3 dB criterion on the DPOAE responses of the P3 data-set, the number of available cases was reduced to 560 (573-13). The thirteen cases which were excluded (in order to avoid any outliers in the data distribution) presented S/N ratios well above 3 dB, but DPOAE amplitude levels between -0.5 and -3.0 dB SPL at the frequencies 3.0 or 4.0 kHz.

The results from the free distribution confidence interval estimation are shown in Table 4. According to these, a DPOAE response is considered a "pass" when both the left and right ear responses have S/N ratios at 2.0, 3.0 and 4.0 kHz higher than 6.0, 7.0 and 6.0 dB respectively. For this set of criteria there is no need to include additional rules regarding the DPOAE amplitude level at each tested frequency. In all 560 cases the DPOAE amplitudes at 2.0, 3.0 , 4.0 and 5.0 kHz were larger than 0 dB SPL. For comparison purposes Table 4 includes values of scoring criteria, computed via a free distribution (Hatzopoulos et al, 1999), from normal neonatal TEOAE recording elicited by a linear TEOAE protocol.

Table 4. Scoring criteria for DPOAE and TEOAE protocols for 2.0, 3.0 and 4.0 kHz. The table presents also data from a non-audiometric frequency at 5.0 kHz (shaded cells) , which might offer useful information about basal cochlear functionality. The TEOAE scoring criteria have larger values because the TEOAE click stimulus (on the average of 76 dB SPL) is larger than the intensity of the L_2 tone (mainly responsible for the generation of the cubic distortion product in an asymmetrical DPOAE protocol).

Frequencies (kHz)	P3 – DPOAE protocol (dB)	Linear TEOAE protocol (dB)
2.0	6.0	8.0
3.0	7.0	11.0
4.0	6.0	10.0
5.0	8.0	10.0

3.4 Feasibility of the DPOAE testing

The screening yields (feasibility index) of the 75-65 DPOAE protocol were estimated as follows: For the available WBN infants we have considered that the test was feasible in 560 cases (560 / 573 = 98%) which produced DPOAE responses with acceptable S/N ratios. The calculation of the feasibility estimate from the DPOAE responses of the NICU residents was more complicated. The reason for this complexity was that a number of NICU neonates did not produce a DPOAE response even when the TEOAE response was considered a "pass", and that in 6 cases no TEOAE response was present at the initial test. The main contributor to the lack of DPOAE responses was the size of the neonatal DPOAE probe, whose diameter was probably large for the meatus size of the preterm infants (note that the neonatal TEOAE probe is considerably thinner). To overcome this technical difficulty we established a rule by which we considered a DPOAE response as valid within 4 re-tests for which TEOAEs were present sequentially at least in 2 out of the 4 re-tests. After the application of this rule 38 infants (38 / 50 = 76%) were evaluated as "pass" cases according to the established DPOAE criteria.

For the well baby population no ABR testing is conducted once the case is considered a "pass". In contrary due to the small number of NICU residents (approximately 12% of the total number of births in our hospital, which translates to approx. 140 cases per year) it is possible to perform a diagnostic ABR independently of the outcome of the OAE testing. For the 38 cases tested with DPOAEs, a diagnostic ABR at 50 dB nHL indicated 3 cases with monolateral losses (i.e. "fail cases). The corrected age of these three infants at the time of the ABR test was greater than 40 weeks. The results from the first ABR were considered as non-conclusive, due to issues related to a possible non-maturation of the central pathways. A second ABR test after 8 weeks indicated that 2 of the 3 cases had normal ABR latency values. According to these data the false negative estimate of the DPOAE testing in the NICU was equal to 1/ 38 = 2.6%. From the 12 infants which did not satisfy the scoring criteria, ABR testing identified 3 bilateral and 5 monolateral hearing impairment cases. In this context, the false positive estimate of the DPOAEs was equal to 4 / 50 = 8%. For comparison purposes we present the data from the TEOAE recordings. From the 12 cases , where a DPOAE evaluation was not possible, 5 resulted as fail and 7 as partial pass since all these cases the S/N limit at 2.0 kHz was not satisfied. The ABR evaluation (at 50 dB nHL) has indicated that 4 of the partial pass cases presented monolateral losses. In this context ,it might be said that the false-negative ratio of the TEOAEs was 7 / 50 = 14%.

4. Discussion

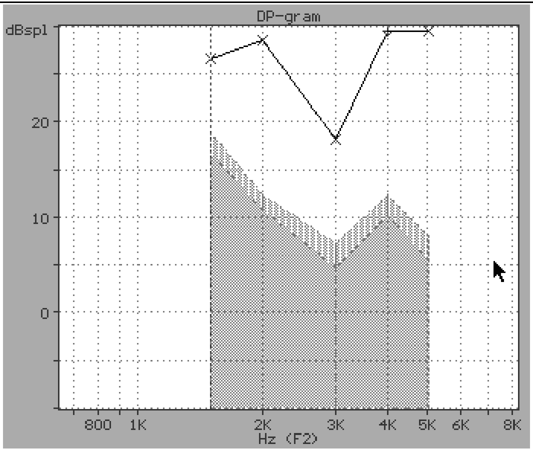
The objective of this study was the evaluation of the clinical performance of a DPOAE screening protocol , in the context of UNHS program. The presented data suggest that a DPOAE cochlear evaluation, at 3 pre-selected frequencies, has a good test feasibility (98%) for the WBN infants and an average feasibility (76%) for the NICU infants.

From the three proposed DPOAE protocols, the high stimulus intensity asymmetrical protocol (75-65) presented responses with the highest S/N ratios. The advantage of using high level primaries is that a cochlear response can be elicited even in a noisy environment. It might be argued that the high level primaries might elicit responses even from subjects with mild mid-frequency hearing losses, but it should be noted that the current TEOAE programs are using click stimuli (optimized in the mid-frequencies) as high as 85 dB SPL, 10 dB higher than the L_1 of the suggested protocol.

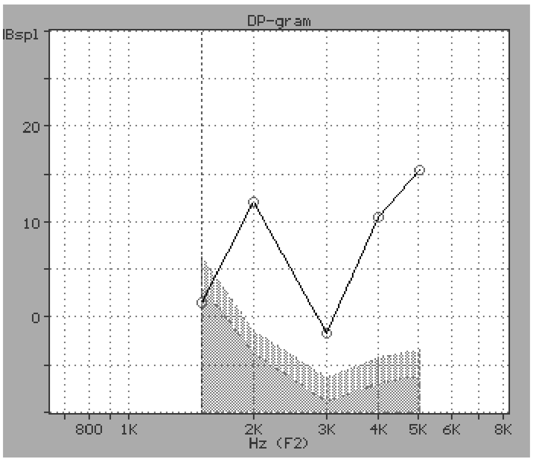
In terms of recording time requirements, significant differences were found between the tested WBN DPOAE and TEOAE (mean 54 ± 32 s) time estimates with the latter showing larger mean values. No significant differences were found between the recording times from the P1 (mean 36 ±17.2 s) and P2 (mean 35 ±15.9 s) protocols. As expected significant differences were found between the recording times of P3 (mean 29 ± 11.2 s) and P2 , P1. These recording time differences can be attributed to the fact that the DPOAE protocols stimulate in a more efficient manner the cochlea, therefore more robust DPOAE responses contribute to fewer necessary recording averages. For the NICU infants the recording time requirements are not an important issue, because several TEOAE or DPOAE re-tests are necessary in order to obtain an acceptable OAE response.

The characteristics of the DPOAE recordings of our study resemble the data reported by previous studies for WBN infants (Abdala, 1996; 2000;Quinonez, 1999; Norton et al, 2000; Gordts et al, 2000) and NICU infants (Smurzinsky, 1993; Gorga et al, 2000). In our study the majority of the WBN DPOAE responses had profiles similar to Figure 1. Approximately 10% of the responses of the P1 and P2 datasets and 5% of the responses of the P3 data-set, presented what we might call a DP-gram notch pattern. Similar observations have been presented in a previous study by Marco et al, where the DPOAEs were described as presenting peaks at 2.0 and 5.0 kHz and “valleys or notches” in between. Representative responses of the notch in the DP-gram are shown in Figure 3. We have postulated that this “notch” behavior around the DPOAE frequency of 3 kHz (referenced to F_2) might have been caused by a number of factors: (a) by an interaction between the DPOAE response (referenced to F_2) and a nearby peak of a spontaneous emission; (b) by an interaction between the DP cubic response and a standing wave in the external meatus; and (c) by a particular resonance of the ILO-92 probe. The fact that the “DP-gram notch” was observed in only a small percentage of the tested cases favors more the first two postulates. The hypothesis of an interaction between cubic DPOAE responses and a spontaneous emission is quite possible, but one should expect interactions in other frequencies as well (i.e. at 2.0 or 4.0 kHz). Within this context we consider more probable the standing wave hypothesis, which depends on the position of the ILO-probe and the dimensions of the external auditory meatus (note: The DP-gram notch usually becomes less profound when the probe is positioned closer to the tympanic membrane). Additional support for the first hypothesis was derived from the NICU preterm data-set. Approximately in 29% (11 cases) of infants tested with the P3 protocol, a notch pattern was visible in the DP-gram. In the preterm cases the DPOAE probe was positioned further away from the entrance of the auditory meatus, due to large diameter of the probe. Despite this confirmation, additional analyses are considered necessary in order to resolve the origin of this particular DP-gram behavior.

Regarding the bimodal structure of the data, we postulate that several factors might be responsible, such as : (a) a change of the number of active components, in the DPOAE response, to passive components which are less sensitive to the incoming vibrations. Such an change is taking place approximately around the stimulus level of 65 dB SPL (Mills et al, 1993; Whitehead et al, 1995) ; and (b) parameters related to the recording instrument (ILO-92). The data from this study do not offer a satisfactory explanation, therefore further studies are necessary to elucidate the issue of the bimodal structure of the DPOAE data.



A



B

Figure 3. DP-gram from a 75-65 protocol for : (A) a WBN infant (case 2246_L001) and (B) from a NICU infant (case P0031_007). The DP-gram “notch” at 3.0 kHz is considered as a technical artifact, caused by the interaction of the DPOAE response and a standing wave, and not an indication of a cochlear impairment. The horizontal axis shows frequency in kHz and the vertical axis the amplitude of the S/N ratio in dB. The "depth" of the notch is minimized by a repositioning of the DPOAE probe.

In terms of screening efficiency in the WBN environment, both TEOAE and DPOAE (P3) protocols perform equally well. This high DPOAE yield permits us to postulate that the DPOAEs might outperform the linear TEOAEs in noisy WBN environments, due to the fact that the energy delivery to the cochlea of the DPOAE protocol is more efficient. The S/N ratios at 5 kHz showed the highest values among the tested frequencies, thus it is conceivable that by using a DPOAE screening test with 4 frequencies, we might evaluate more accurately the basal cochlear partitions.

The screening efficiency of DPOAEs in the NICU infants was average and only 76% of the cases presented acceptable responses. The reported false-positive estimate was 8%, a value lower than the data reported by Barker et al, referring to DPOAE false-positive rates of 11% to 35%. Although these estimates are bound to change with the use of larger data-sets, it should be mentioned that some bias is introduced to the results by employing screening practices which are not fully optimized. For example we have identified that in the NICU screening a major component of the "fail" cases is related to technical problems (i.e. the DPOAE probe is too large in comparison to the auditory meatus). Analyses of variance of the NICU DPOAE responses have indicated that the majority of the "fail" responses were corresponding to an age between 30 and 32 weeks. Testing at a latte age, for example at 35th week, might provide better feasibility scores and lower false-positive rates.

In conclusion, the data of the study suggest that a high stimulus DPOAE protocol can be used in a UNHS program. Although the DPOAEs have significant lower time-recording requirements the data show that both TEOAEs and DPOAEs perform equally well on the WBN infants. This implies that hybrid programs can be designed using DPOAEs in the noisiest testing site (i.e stage 1 or stage 2) . For the NICU infants the data indicate that some technical problems must be resolved first (i.e. smaller probe sizes), before a proper evaluation of the OAE protocols is conducted. The data of this study suggest that a DPOAE protocol cannot evaluate properly all the NICU infants and it is strongly suggested to employ a clinical program combining multiple sessions of DPOAEs /TEOAEs and ABR.

Acknowledgements

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Investigation of Noise Threats and Their Impact on Hearing in Selected Schools

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Abstract: Noise measurements conducted in selected schools in Gdansk area are presented in this paper. The main aim of this research was to determine noise threats at schools. Some objective measurements of the acoustic climate were performed employing a noise monitoring station engineered at the Multimedia System Department, Gdansk University of Technology. Simultaneously, subjective noise annoyance examinations were carried out among pupils in chosen schools. The survey includes a noise analysis in places of residence, music preferences and preliminary hearing tests results taken after the exposure to noise during breaks. Hearing tests employing a distortion product otoacoustic emission (DPOAE) method, have been performed twice -- before and after the exposure to noise. The noise dose analysis based on average time spent by a pupil at school is also presented. The obtained results reveal that an unfavorable noise climate occurred in surveyed schools. This was also confirmed by the results of the subjective examinations. The conducted hearing tests did not reveal essential changes in the cochlea activity of examined pupils. This means that the noise during breaks and physical exercises did not constitute a risk to their hearing system. However, it may be considered as an essential source of annoyance.

Key words: noise measurement, hearing, noise at school

1. Introduction

Numerous research studies indicate that noise at schools is a serious threat both for pupils and teachers [2],[3]. The pilot study presented in this article also took into consideration the acoustic conditions in classrooms [2],[3],[4]. The presented noise and hearing measurements are the continuation of the earlier screening hearing tests carried out by means of the “I can hear” system in numerous schools around the whole country [5]. The obtained results revealed frequent occurrence of various hearing problems among pupils. For this reason, it was decided that information about the acoustic climate in some selected schools should be gathered. Noise measurements were performed by means of an automatic noise measurement station designed in the Multimedia System Department, Gdansk University of Technology [6]. At the same time, hearing examinations for selected persons were performed employing the otoacoustic emission method.

2. Materials and Methods

The results of noise measurement, obtained by means of the Multimedia Noise Monitoring System (MNMS), are presented below. The measurements were performed in selected schools. The outdoor noise were neglected (schools were located in quiet places). The data gathered were utilized to perform the noise dose analysis. This was done to determine the noise exposure in considered places. In designated cases (i.e. in schools) the noise dose analysis was expanded by the assessment of hearing. To achieve this, the distortion product otoacoustic emission (DPOAE) method was applied. The hearing was examined twice. First, directly before the exposure to a given type of noise, and then immediately after the exposure. The performed analysis combined the obtained noise and hearing measurement results.

The following noise parameters L_{AFmin}, L_{Aeq}, L_{AFmax} were measured independently over broadband and in one-third octave bands (L_{AFmin}, L_{AFmax}, - the lowest and highest A-weighted sound levels for fast time weighting, that occurred during the measurement, and L_{Aeq} - the A-weighted equivalent continuous noise level over a specified period of time that represents the same energy as the actual time varying noise signal [8]). A cumulative distribution for time history values of L_{AF} instantaneous levels was also calculated. A measuring microphone was located 1.9 m above the floor level for every measurement. For all measuring series, a place where pupils gather most often was selected in order to determine correctly a real noise dose to which they are exposed.

Hearing examinations employed the DPOAE method using GSI 60 DPOAE system. The following parameters of the stimuli were used during tests: L1 equals 65 dB, L2 equals 55 dB, f2/f1 = 1.2, DP frequency (geometric mean): 1062, 1312, 1562, 1812, 2187, 2625, 3062, 3687, 4375, 5187, 6187, 7375 Hz. A DP signal level and a noise floor for every stimuli were registered. The test result was accepted if the difference between evoked otoacoustic emission signals and the noise floor was not less than 10 dB. The reason of such selection of parameters was because the noise impact on the hearing system is the strongest for middle and high frequencies. The test was carried out in rooms specially adapted for this purpose. Some measurements performed in schools were interfered with sounds coming from adjoining rooms.

For the DPOAE hearing measurements a single measurement unit was used. For that reason only a few pupils could have been examined during a single measurement series which included the measurement of hearing before and after the exposure to noise. This means that the assessment was done only for one pupil directly before the exposure to noise. The hearing examination for all other pupils was delayed with respect to the beginning/end of the exposure to noise. This could possibly influence the change of activity in the cochlea for these people. Therefore, the effect evoked by the exposure could decrease directly before the examination. Taking these facts into consideration, we may assume that only a small group of pupils could have been successfully examined.

In addition, an objective noise measurement was extended by a subjective measurement by means of a dedicated survey. The survey consisted of three parts. The first part involved getting information such as age, sex, class, school. The second part included questions about noise in places of residence and exposure to noise related to musical preferences. The last part concentrated on noise climates in schools in typical circumstances (lessons, breaks, etc.).

3. Analysis of Noise Measurement Results

The results of measurements for particular schools are presented in Table 1. This table includes all measurements taken in every school in any one of the measurement series. In addition, for every measurement series the cumulative distribution and one/third spectrum were also calculated. The equivalent level was used to determine the noise dose that occurred during breaks [1] (Noise Dose 1). Moreover, the noise dose analysis was extended by a daily noise dose estimation. Time of the exposure to noise corresponds to duration of breaks of a typical learning day (Noise Dose 2).

Table 2. Measurement results obtained for particular schools. Noise levels were expressed in dBA (reference level: 2×10^{-5} [Pa])

No.	L_{AFmin}	L_{Aeq}	L_{AFmax}	*Time* [s]	Noise Dose 1 [%]	*Time* total [s]	Noise Dose 2 [%]
School No. 1							
1	64.5	86.9	102.1	600	**3.2**	3000	**16.2**
2	67.4	89	105.5	600	**5.2**	3000	**26.2**
3	74.5	86.1	111.3	1200	**5.4**	3000	**13.4**
School No. 2							
4	67.2	85.5	106.8	900	**3.5**	3600	**14.0**
5	69.5	84.3	103.1	900	**2.7**	3600	**10.6**
School No. 3							
6	56.5	79.1	93.4	600	**0.5**	3600	**3.2**
7	72	83.6	97.4	600	**1.5**	3600	**9.0**

Taking equivalent levels into consideration, it was affirmed that the highest noise level occurred at school No. 1 (the highest L_{Aeq} was equal to 89 dBA !). This was related to pupils' behavior. They behave extremely vigorously, and were the main source of noise during breaks. It is worth emphasizing that pupils from school No. 1 were the youngest from all of those examined. In school No. 2, children were between the ages of 13 and 15. Youth aged from 16 to 19 attended school No. 3. The age of pupils explains their behavior during breaks. In school No. 2, noise levels were slightly less obtrusive (L_{Aeq} about 85 dBA). In this school, additional source of noise was loud music played from the loudspeakers. Long corridors without any sound absorbing materials were also the factor that heightened noise level. The lowest noise levels were identified in school No. 3 (83.6 dBA). For school No. 1, a more dynamic range of noise levels was obtained, which was also reflected in the shape and width of the cumulative distribution curve (see Fig 1). Cumulative distributions obtained for schools No. 2 and 3 showed essential similarity. High and steep slopes indicated that noise levels concentrated near one constant sound level. This reflects the character of the main noise source. As mentioned before, this was the loudspeaker system in school No. 2, for example. In school No. 3, the noise was produced by loud conversations.

The noise spectra were fairly similar for all schools in general. The significant difference occurred for low and middle frequencies. For elementary schools No. 1 and 2, high levels were identified for frequencies lower than 100 Hz. This was certainly related to pupils' vigorous behavior characteristic for their age. As mentioned before, pupils from school No. 3 were adolescents thus they behaved more calmly. The greatest noise levels in the range between 630 and 2500 Hz were observed in school No. 1.

However, the analysis it was affirmed that the noise dose during a single break is insignificant from the statistical point of view. The noise dose amounts to 5% for school No. 1, approximately 3% for school No. 2 and 1% for school No. 3. These values are obviously greater for a total daily exposure -- the biggest for school No. 1 (26%) and to some extent lower for schools No. 2 and 3.

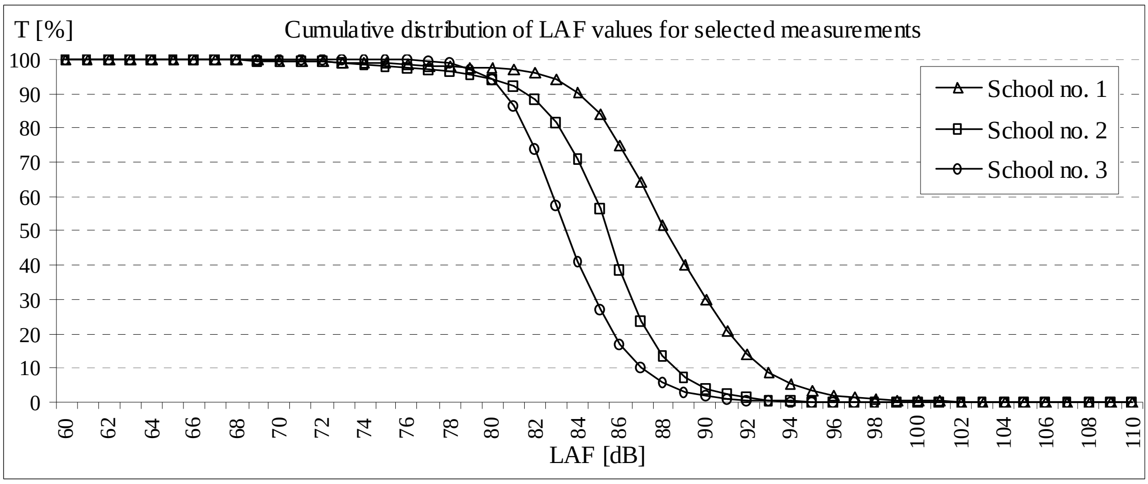

Figure 1. Cumulative distributions for particular schools (for selected measurements)

3.1 Hearing Measurement Results

Twenty persons overall took part in hearing tests. Ten of them were examined in school No. 1, five in school No. 2, and the remaining in school No 3. Two different aspects were taken into consideration while analyzing the results. First, the number of "passed" and "failed" tests for the second examination were determined. The result of the first examination served as reference. The symbol "+Pass" indicates that a pupil failed the first examination and passed the second one. The symbol "–Pass" signifies a reverse situation (a test passed in the first examination and failed after the exposure to noise). The results are presented in Table 2, in the "DPOAE test results" column. The second kind of analysis determined how the DP signal level changed under the influence of the exposure to noise. The results of this analysis are presented in Table 2, in "The average changes of DP signal level" column.

Table 2. Hearing testing results using DPOAE method (in %)

School	DPOAE test results			The average changes of DP signal level		
	+ Pass	- Not Pass	No change	Increase	Decrease	No change
No. 1	11.0	13.6	75.4	30.3	28.1	41.6
No. 2	10.0	19.2	70.8	27.5	30.0	42.5
No. 3	3.3	12.5	84.2	36.7	34.2	29.1

The cochlea activity characteristics that were obtained by means of the DPOAE method do not clearly confirm that the noise occurring during breaks has negative impact on the hearing of examined persons. The average changes of the DP signal level for examined persons substantiated in this situation. Differences between the increase and decrease of the DP level induced by the exposure to noise measured for every group of pupils, were insignificant regardless of the type of school. Differences for the DP levels characteristics were within the range of measurement error which may be produced by a different location of the measurement sensor in the ear canal. It is important to emphasize that to obtain reliable results with the DPOAE method, a very silent room is required. From all considered cases, the best measurement conditions were in school No. 1. The measurements in school No. 2 and 3 were done in the headmasters' offices. In these circumstances some measurements were disturbed by sounds from adjoining rooms. However, the measurements were repeated in such situations.

3.2 Survey Result Analysis

Evaluation of noise at place of residence

On the basis of the answers related to noise at place of residence, it was found that the questioned people's environment is loud during the day and quiet in the night. The most often indicated source of noise in the place of residence was a roadway noise (38.6%). The neighborhood noise (34.1%) was a second one.

Evaluation of noise at school

The noise measurement results are consistent with the survey results for the noise during breaks. More than 62% persons questioned in all types of schools estimated the noise at breaks as very loud while almost 30% of the remaining pupils said that it was loud. The presence of loudspeaker systems is typical in schools for older children and youths. Such an installation can constitute an essential source of noise. Noise during lessons was in most cases assessed as low or moderate.

Noise during lessons was the largest problem with the youngest pupils from school No. 1. This concerned speech intelligibility. More than half of the questioned pupils from this school judged the noise as loud or very loud. Pupils also noticed that noise should be reduced especially during breaks (more than 60% answers). Some of them mentioned the noise problem in classrooms during lessons (about 20%). Merely 14% of the questioned persons pointed out that noise should be diminished during physical activities. Approximately 10% of the pupils did not notice the noise problem at school at all.

Evaluation of noise concerning music and entertainment preferences

The analysis of preferences on music and entertainment of pupils from different groups of age provided very interesting information. As many as 60% questioned pupils from school No. 3 (youth) listen to music at loud or very loud levels, 30% preferred moderate levels. However, 60% of questioned pupils from school No. 2 and 1 pointed out that they listen to music at moderate or low levels. This data show, how the preferences change with age. This could be an essential factor in the loss of hearing inducted by noise amongst adolescents. Using headphones for a long time is the next cause of the hearing impairment risk. Older pupils more often used such kind of equipment. Nowadays, there is a very large offer of portable music players on the market and teenagers willingly use them. Pupils were also asked how much time they spend listening to music. Their answers confirm the intuitive presumption that duration of a single session of listening to music grows with age. Listening to music is very popular among the youths. The next type of noise hazard is participation in loud parties and musical events. Also in this case, the percentage of youngsters who prefer such type of free time activities grows with age.

4. Summary and Conclusions

The pilot study performed shows that the noise climate in the considered schools is adverse. The main reasons of the high level of noise in schools are pupils' behavior and the lack of sufficient absorption of walls in classes and corridors. In some cases loudspeaker systems constitute another essential factor of the increase in total noise level. Taking this into consideration, it is necessary to emphasize that noise in schools can be a key source of tiredness and stress, not only for pupils but also for teachers. On the other hand, for older pupils, listening to music and participation in loud sound events constitute a high risk of developing a hearing loss. Based on the criteria of risk of hearing loss induced by noise [7], it was affirmed that for elementary schools the noise level during breaks may contribute to real hearing damage risks. For other types of schools the risk is between mid and high. The data analysis of hearing measurements at schools does not confirm the negative influence of noise on hearing, as yet. This is because the time of the exposure to noise was too short to produce measurable changes in the activity of the inner ear.

The results from the survey showed that pupils are exposed to annoying noise not only in their place of residence but also in their schools. The type of school, behavior of pupils, installation of loudspeakers, noise absorbing materials, etc., all form acoustical climate. Also, it was observed that older pupils have greater tolerance for excessive noise in their environment. Younger pupils tend to avoid loud sounds. As they grow up, they put themselves at loud noise threats more voluntarily.

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Contactless Hearing Aid for Infants Employing Signal Processing Algorithms

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Abstract: The proposed contactless hearing aid is designated to be attached to the infant's crib for sound amplification in a free field. It consists of 4 electret microphone matrix, and a prototype DSP board. The compressed speech is transmitted and amplified via miniature loudspeakers. Algorithms that are worked out deal with parasitic feedback, which occurs due to the small distance between microphone and monitors and potentially high amplification required. The beamforming algorithm is based on an artificial neural network (ANN). The ANN is used as a nonlinear filter in the frequency domain. Principles of algorithms engineered and the prototype DSP unit design are presented in the paper. Also, results of experiments simulating the real-life conditions are analyzed and discussed.

1. Introduction

An early intervention in rehabilitating an infant having a hearing loss is of a great importance. However this poses a unique set of problems. Since it is not possible to evaluate infant's hearing loss by subjective methods, and then to check the validity of the prescribed gain, thus evoked potential (ABR – Auditory Brainstem Response) assessment is required to measure the hearing loss and establish hearing aid fitting targets [1]. An audiogram is to be predicted on the basis of the ABR assessment, and then the appropriate amplification of the hearing aid is set up. This is especially important in terms of the infant's speech skill development and understanding. It should however be remembered that hearing loss evaluation and fitting are only first elements of the rehabilitation chain. Prescribing a typical hearing aid for infants is not very practical. An infant tries to take the hearing aid out, plays with it, etc., which may cause changes to the hearing aid settings or may even damage the device. The size of the behind the ear (BTE) hearing aid is quite large in comparison to the infant's head/pinna, and it is uncomfortable in case when an infant wants to lay its head on a side. On the other hand, any in-the-ear (ITE) or insert (ITC – in the canal or CIC – completely in canal) hearing aids are not recommended to wear for an infant because of the growing of the ear canal and changing its anatomical shape in time. In addition, a hearing aid may cause some malformation of the bony ear canal.

Due to the rapid development and an increase of power of miniature signal processors along with DSP algorithms it is possible to think up a totally different approach to the hearing aid for infants. The proposed contactless hearing aid is designated to be attached to the infant's crib for sound amplification in an acoustical field. It transmits an amplified and compressed signal of an infant mother's speech (or any person taking care of an infant) via miniature loudspeakers. A design of the dedicated extension card for the TMS320VC5509A DSP development board has been made at the stage of research work preparation. The prototype was engineered at the Multimedia Systems Department, to implement and test algorithms enabling functioning of the contactless hearing aid [2].

Algorithms that are worked out deal with some obvious limitations of the free-field hearing aid such as parasitic feedback, which occurs due to the small distance between microphone and monitors and potentially high amplification needed. That is why one of the algorithms implemented is beamforming which controls a feedback between microphones and monitors. For this purpose a matrix of 2 electret microphones has been employed. The beamforming algorithm is based on an artificial neural network (ANN), thus the main problem concerns choosing appropriate feature vectors that are feeding the given algorithm inputs. The ANN is used as a nonlinear filter in the frequency domain. The main task of the spatial filter is to estimate the desired signal arriving from the front direction. It is neither desirable nor possible to completely attenuate signals from lateral and backward directions. Because spatial filter works in the frequency domain, it is assumed that each spectral component representing signals coming from unwanted directions is to be attenuated by at least 40 dB [3][4][5]. The spectral components representing signals coming from the forward direction should remain unaltered. The effectiveness of the algorithm engineered and the resultant speech intelligibility depends on the proper decision made by the neural network, thus the learning procedure of the ANN is very important. This decision is made basing on the values of the parameters of sound that are similar to those used by the human auditory system. These parameters represent both interaural intensity ratio and interaural time difference. In the testing phase various combinations of signals are introduced to the ANN inputs and the algorithm is checked as to its effectiveness.

Apart from beamforming, various techniques based on signal processing are commonly employed in digital hearing aids in order to prevent the occurrence of the acoustic feedback [6][7]. Two such techniques (adaptive notch filtering and adaptive feedback cancellation) were studied and the former one was chosen for experiments and implementation as an additional protection against feedbacks.

Another algorithm implemented in the hearing device is the voice activity detector (VAD). Its goal is to prevent the amplification of loud unexpected sounds as well as any other acoustic signals that should not be presented to the infant.

This paper is structured as follows. In Section 2, a brief description of the contactless hearing aid set-up is introduced. In Section 3, the beamforming algorithm based on ANN is described. Other methods for the acoustic feedback elimination are discussed and compared in Section 4. The Section also contains preliminary experimental results. Section 5 presents practical implementation remarks, especially concerning the design of an extension card for the digital signal processor. Finally, in Section 6 the authors summarize the results obtained and outline future research related to the contactless hearing aid (PCT patent pending).

2. Contactless Hearing Aid

In contrast to all standard digital hearing aid solutions, where a microphone, digital signal processor and receiver are enclosed in one shell, the contactless hearing aid set up comprises of the following three separated modules:

- microphone array (4 electret microphones) with preamplifiers mounted in front of the infant's bed,
- DSP unit responsible for signal processing,
- miniature loudspeakers mounted near the infant's head.

In Figure 1, the infant's crib equipped with a contactless hearing aid is presented.

Figure 1. Contactless hearing aid application

The distributed structure of the device allows not only for its convenient installation in the infant's crib but is also essential for reducing the possibility of the parasitic feedback occurrence, as the microphones and loudspeakers can be easily separated from each other. Moreover, the DSP unit may be mounted far from the infant's body or even be hidden under the bed in order to minimize its eventual negative influence on crib ergonomics.

Since the hearing aid is not a simple sound reinforcement system, advanced signal processing techniques have to be applied in order to meet specific requirements of the signal capture and hearing loss compensation. The functional diagram illustrating the processing modules of the hearing aid is shown in Figure 2.

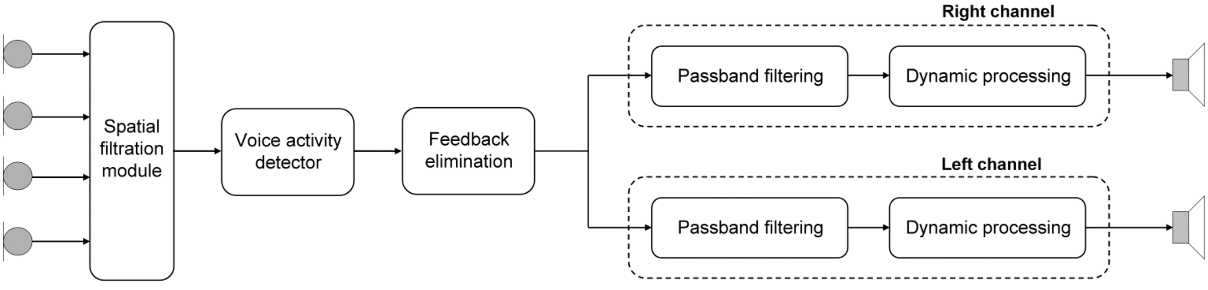

Figure 2. Block diagram of signal processing in the contactless hearing aid

The first module comprises of four omni directional microphones which capture sounds coming from all directions. In order to attenuate signals emitted through the loudspeakers as well as the signals emitted by the infant itself or any other undesired sources, the spatial filtration module is employed. Spatial filtration (beamforming) takes the advantage of the fact that the distance from the source to each microphone in the array is different, which means that the signals captured by the microphones will be phase-shifted replicas of each other. Knowing the amount of the phase-shift at each microphone in the array is sufficient to calculate the direction [8]. It is assumed that the desired signal (e.g. coming from a speaker) that should be presented to the infant's ears may come only from the sources located in front of the crib. Thus the adaptive spatial filtration algorithm tracks the speaker only in a limited area. As the basic aim of this technique is to attenuate all signals that are undesired for further processing, this can also be viewed as a particular method for the parasitic feedback cancellation.

In order to prevent the amplification of loud unexpected sounds as well as any other acoustic signals that should not be presented to the infant, the voice activity detector (VAD) is employed. This algorithm operates in the frequency domain and takes the properties of the captured signal spectrum into account to decide whether the signal should be further processed [9]. Furthermore, based on the detector state particular groups of electronic components (e.g. power amps, digital to analog converters) are either activated or disabled. It allows minimizing the power consumption of the device, which is especially important when the hearing aid is battery powered during outdoor operation.

Despite the beamforming technique, it can be expected that in some particular situations the acoustic feedback may occur. Thus the captured signal is continuously analyzed in order to eliminate this parasitic phenomenon. Although the signal is further processed by two separated modules for left and right channel respectively, and then emitted through the miniature loudspeakers, the acoustic feedback elimination procedure is applied to the mono signal just before the separated channel processing occur. This seems reasonable since the characteristic of the feedback is common for these two channels.

The last module in Figure 2 implements signal processing algorithms that are typically employed in digital hearing aids. In the first step the signal is divided into subbands. Further, either constant amplification or dynamic processing with compression characteristic is applied in each signal subband based on the characteristics of an infant's hearing impairment. One can notice from the Figure 2 that hearing loss is compensated for each ear independently. Finally, two processed signals are emitted through the miniature loudspeakers mounted near the infant's head.

3. Neural Beamformer

Automatic identification of sound sources direction is still an unsolved problem in many real-life applications, such as for example, hearing prostheses or contemporary teleconferencing systems. The main reason for this is background noise, high reverberation and/or with many concurrent speakers. One approach to reducing this noise is to provide directional field of hearing. Source identification (spatial filtration) system should allow for tracking a target speaker automatically without much delay in order to avoid picking up concurrent speakers by the same microphone channel. This may be done in various ways, however, generally two approaches can be found in literature. One of them is a conventional approach to this problem based on delay summation algorithms, superdirective arrays and adaptive algorithms, non-linear frequency domain microphone array beamformers, etc. [10][11][12][13][14][15][16]. The effectiveness of these algorithms decreases however while performing in reverberant environments. Examples of such algorithms were broadly reviewed in literature, thus they will not be recalled here. The second approach to this problem was proposed in the Multimedia Systems Department, GUT in collaboration with the Institute of Physiology and Pathology of Hearing, Warsaw in previous studies. Namely Artificial Neural Networks (ANNs) have been applied for the purpose of the automatic sound source localization [3][4][17][18][19][20][21][22][23]. Since the current study requires an effective spatial filtration algorithm, thus the approach based on beamforming employing ANNs was reviewed here within the context of contactless hearing aid. The ANN was used as a nonlinear filter in the frequency domain (also time domain neural beamformer is easy to implement).

The first step of experiments consisted in extracting feature vectors to be fed to the ANNs. During the feature extraction process the signal acquired was divided into frames of the length of 256, 512 or 1024 samples. The feature set was based on previously defined parameters under the assumption that a neural network provides an effective non-linear filtering algorithm of an acoustic signal transformed into the frequency-domain [3][4][5][22]. It was assumed that the number of microphone channels should be limited to two. Signal arriving at both microphones can be written as:

$$l(t) = s(t) + n_l(t);\ r(t) = s(t) + n_r(t) \quad (1)$$

where:

- $l(t)$, $r(t)$ - signals received by the left and right microphones,
- $s(t)$ - desired signal arriving from the front direction,
- $n_l(t)$, $n_r(t)$ – signals coming from the lateral or backward directions arriving to the left microphone and to the right microphone. These signals are treated as noise.

The main task of the spatial filter is to estimate the desired signal s(t) arriving from the forward direction. Because spatial filter works in the frequency domain, it is assumed that each spectral component, which represents signals coming from unwanted directions should be attenuated by at least 40 dB (see Figure 3). In Figure 3 a prototype spatial characteristics is shown. The spectral components that represent signals coming from the forward direction should remain unchanged. This can be described by the following expressions:

$$\tilde{L}\left(e^{j\omega}\right)=\sum_{i=1}^{N} g(i) L^{i}\left(e^{j\omega}\right); \qquad \tilde{R}\left(e^{j\omega}\right)=\sum_{i=1}^{N} g(i) R^{i}\left(e^{j\omega}\right) \tag{2}$$

where:

- i – spectral component index,
- $\tilde{L}\left(e^{j\omega}\right)$, $\tilde{R}\left(e^{j\omega}\right)$ - estimates of a signal component L^i in the left, and R^i in the right channel,
- $g(i)$ – attenuation coefficient of noisy components described by the following formula:

$$g(i)=\begin{cases} 1; i \in \textit{Signal} \text{ components} \\ 0.01; i \in \textit{Noise} \text{ components} \end{cases} \tag{3}$$

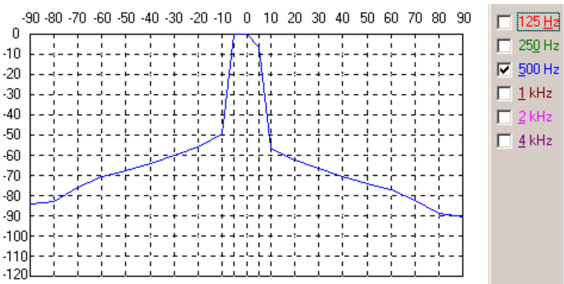

Figure 3. Desired directional characteristic (the same for all frequencies). *x* - axis represents angle, *y* - axis represents attenuation in [dB]

The effectiveness of this algorithm and the resultant speech intelligibility will depend on the proper decision made by the neural network, so the learning procedure is very important. This decision is made basing on the values of some parameters of sound that are similar to those used by the human auditory system. These parameters represent both interaural intensity ratio and interaural time difference. The first parameter, which expresses the interaural spectral magnitude ratio, is described by the following expression:

$$M^{i}=\frac{\min\left(\left|L^{i}\right|,\left|R^{i}\right|\right)}{\max\left(\left|L^{i}\right|,\left|R^{i}\right|\right)} \tag{4}$$

where:

- i – spectral component index,
- L^i, R^i – left and right signals for the ith spectral component,
- M^i –magnitude ratio for the ith spectral component

The second parameter, which expresses the interaural phase difference is described by the following expression:

$$A^{i}=\left|\angle L^{i}-\angle R^{i}\right| \tag{5}$$

where:

- $\angle$ – denotes the signal phase,
- A^i – phase difference of the ith frequency component of left and right channels

The third parameter used in learning phase is defined as:

$$D^{i}=\frac{\left|L^{i}-R^{i}\right|}{\left|L^{i}\right|+\left|R^{i}\right|} \tag{6}$$

where: D^i – relative ratio of the ith spectral component for the left and for the right channel.

It can be shown that parameters described by Eqs. (4) and (6) are in a simple functional relationship and therefore one of them is superfluous. In such a case, parameters representing a single spectral bin can consist of parameters given by Eqs. (4) and (5).

During the learning phase the *Mean Square Error* (MSE) was observed. As seen from Eq. 7 MSE represents the squared error between the current value at the output of the network o and the desired response of the network d.

$$MSE=\frac{1}{PK}\sum_{l=1}^{P}\sum_{k=1}^{K}\left(o_{lk}-d_{lk}\right)^{2} \tag{7}$$

where P is the number of training patterns, and K denotes the number of outputs.

Feed-forward ANNs

The proposed neural network structure and its properties were such as follows: one hidden layer consisted of 9 neurons, the standard error backpropagation algorithm with momentum was used in the learning phase. The BP learning algorithm parameters were as follows: η = 0.5 (learning rate); α = 0.01(momentum ratio). Spectral components were obtained with 512 point FFT procedure using Blackmann window with an overlap of 256 samples. The training file consists of logatoms of every 15^0 elevation. Each direction was represented by 10 sound examples (5 female and 5 male voices). In addition sounds from ±5° were used in this phase. These directions were treated similarly to 0° direction, thus the gain factor was equal to 1, whereas for other directions a value of 0.01 was used.

In the testing phase various combinations of signals were introduced to the ANN inputs. Namely such signals as: tones, tone plus noise, a phoneme (logatom) plus tone, a phoneme plus noise, phonemes and phrases were employed as testing material. Always one of the signals was coming from the front direction (0^0), and the other was the unwanted one and was localized at the angle between 15^0 to 90^0 (horizontal plane). An example of spatial characteristics obtained after the learning phase are presented in Figure 4. As expected sharper minima and maxima were obtained for higher frequency spatial characteristics for the whole angle range. The slope of low frequency characteristics for 15^0-90^0 azimuth is very smooth.

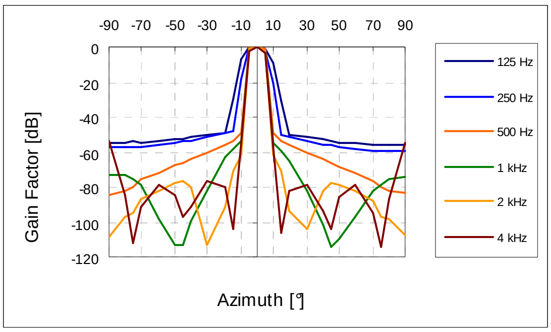

Figure 4. Spatial characteristics of the ANN based filtration algorithm obtained with a multi-tone signal

In Figure 5 an example of a signal spectral representation (sonograms) before and after processing are shown. In Figure 5a combination of signals that was processed by the neural beamformer is shown. In this case the target signal was a logatom and the disturbing one was a 250 Hz harmonic tone. As seen from Figure 5 the disturbing signal is eliminated, but the proposed algorithm causes some distortions that are noticeable in the spectral domain. As seen from the sonogram analysis the target signal has got a formant around the same frequency as such of the concurrent signal. That is why the algorithm after processing cuts off this frequency along with the formant. However the signal-to-noise ratio equals to -60dB, so the distortions do not influence substantially the overall audio quality.

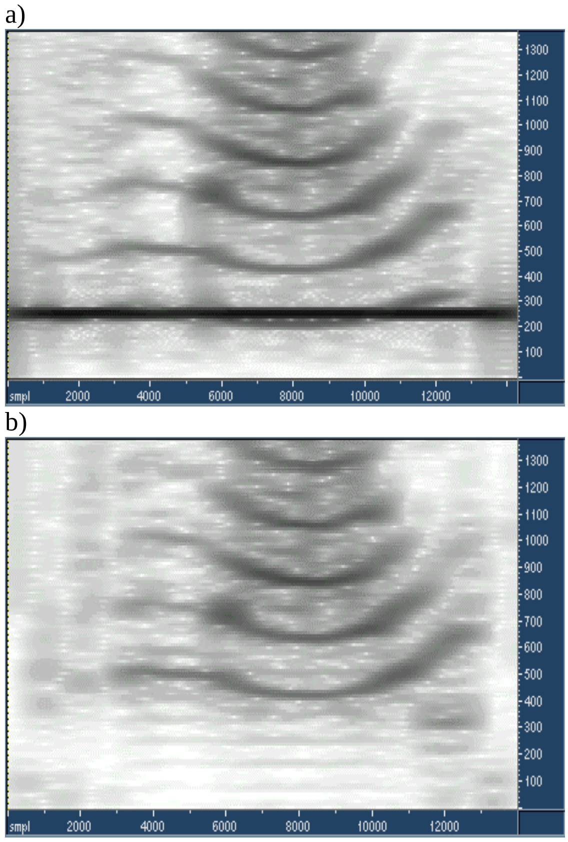

Figure 5. Spectral representation of signals (phoneme, 0°)+(signal f_0=250Hz, azimuth 45°), before processing (a), after processing (b)

After processing various combinations of signals and azimuths it was observed that worse filtration effects were observed when a concurrent signal was close to the target signal (15° azimuth). In this case the dependence of the filtration effects on the character of the signal was also noticed. It can be also observed that definitions of parameters (Eq. 4) and (Eq. 5) cause that signals of the same spectrum composition coming from concurrent directions may not be effectively filtered out by such a beamformer algorithm. This is the most important drawback of the proposed method of spatial filtering, however in such a case a conventional beamformer does not perform well, either.

4. Acoustic Feedback Elimination Methods

This section describes two methods for acoustic feedback elimination (adaptive notch filtering and adaptive feedback cancellation) based on direct signal processing. Such a method forms an additional protection against feedbacks in the designed contactless hearing aid. Adaptive feedback cancellation turns out to be more suitable for DSP implementation, thus preliminary experiments regarding this method are also presented in this Section.

4.1 Algorithms for feedback elimination

Standard methods of dealing with feedbacks based on direct signal processing (e.g. passband equalizing) are static and unable to adapt to changes occurring in the system itself (e.g. microphones and loudspeakers movements) or in the acoustic environment. Two dynamic methods are often used to limit feedbacks: adaptive feedback cancellation and adaptive notch filtering.

The adaptive feedback cancellation method is very similar to algorithms used in acoustic echo cancellation for teleconferencing systems. The idea is to accurately model the loudspeaker to microphone transfer function *F* and then use this model to remove all of the audio sent out the loudspeaker from the microphone signal. An illustration of the method is presented in Figure 6.

There are many methods available for estimating the coefficients of an adaptive filter *F'*, for example NLMS (Normalized Least Mean Squares), RLS (Recursive Least Squares) [24]. However, the resulting estimators are biased because the source signal *v* and the loudspeaker signal *u* are correlated. The bias can be eliminated by reducing the correlation. This can be achieved directly in the signal loop (by delaying or non-linearly distorting the loudspeaker signal *u*) or in an additional identification loop (by means of prefiltering the input signal *y* and the output signal *u* which assures that both the source and the loudspeaker signals *v* and *u* are whitened) [7]. The latter variant is computationally more complex.

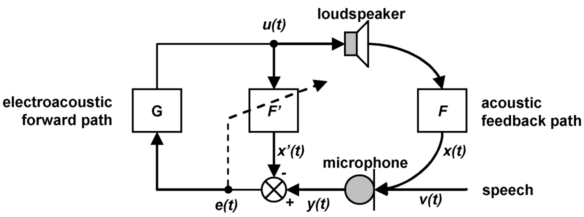

Figure 6. Diagram of the adaptive feedback cancellation method

Adaptive feedback cancellation requires a significantly more powerful digital signal processor than adaptive notch filtering. It is capable to eliminate any audible signs of a feedback at the cost of some minor sound distortions.

The goal of notch filters deployed in the electroacoustic forward path between the microphone and the speaker is to eliminate frequency components resulting from the acoustic feedback. In this method, local maxima of the signal amplitude spectrum are detected and classified whether they represent feedback components. If a maximum representing a feedback component is identified, a notch is deployed with a centre frequency equal to the frequency of the local maximum [25].

There are two main steps in automatic notching algorithms: feedback discrimination and notch deployment. Feedback discrimination is based on a few properties of feedbacks that are very useful in separating them from natural sound features. The amplitude of a feedback component rises monotically and exponentially, while its frequency remains constant, which is illustrated in Figure 7. There are usually no harmonics of a feedback component, however non-linearity of electroacoustic devices working with high-level signals can be responsible for creating them. After a notch is placed on a potential feedback, its amplitude not only decreases by some value but continues to decrease at an exponential rate. This helps to verify the correctness of the feedback discrimination [26].

Notch deployment algorithm determines the parameters of new notch filters and rules for their deployment. A notch filter cannot be too narrow because of gradual changes in the acoustic environment and because of a limited precision of frequency identification. The width of the filter should be equal to 5 or 10 Hz, which guarantees the high effectiveness of the feedback suppression for a longer period of time. The depth of the filter equal to approx. 6 dB is sufficient to bring a feedback frequency back into stability; deeper filters would only decrease the sound quality.

The amount of notch filters is usually limited to the range from 12 to 20, since this number is sufficient to suppress all feedbacks that occur simultaneously. Because of the gradually changing acoustic environment, frequencies of feedback components are never static. Thus already-allocated notch filters is redeployed if required. When a new feedback component is detected, it is checked whether a filter has already been deployed at such frequency. If yes, the existing filter is deepened. If there is a filter with the frequency similar to the frequency of the feedback, the filter is widened to cover both frequencies. If all filters are allocated then the oldest filter is reset and redeployed at the new frequency.

The computational complexity of the adaptive notch filtering method is rather low while its effectiveness in the feedback elimination is high. Furthermore, sound distortions introduced by this method are insignificant. The only disadvantage is the fact that in order to be detected and eliminated the feedback components must first appear in an audio signal and may be audible for a short time.

Because of very strict power consumption constraints placed upon the hardware used in the embedded hearing aid device and because of its computational capabilities, the adaptive notch filtering method is chosen for experiments and implementation.

4.2 Experiments

The experiments regarding acoustic feedback elimination were focused on the implementation of the feedback component discrimination algorithm on the PC platform. During the experiments, pieces containing both speech and music parts were played through a computer speaker and recorded with a microphone. The placement of the speaker and microphone and their output and input gains were altered in order to produce a large variety of feedbacks. The recorded sequences were then processed to identify all local maxima in the signal spectrum and to recognize any feedback components.

For the purpose of the acoustic feedback cancellation feedbacks can be divided into three groups, as illustrated in Figure 7. The first group contains feedback components with amplitudes rising slower than approx. 6 dB per one time frame of the signal (the frame length of 46 ms was used). Such components are considered to be parasitic if their amplitudes continue to increase for 6 time frames in a row. The second group is formed by the potential feedbacks with amplitudes rising faster than approx. 6 dB per time frame. These components are suppressed if their amplitudes rise monotically for 3 frames in a row. The last group consists of feedbacks which are impossible to track directly because their amplitudes increase from the background level to the maximum level allowed in the system almost instantly (during the length of one or two time frames). There is only one way of dealing witch such feedbacks: all components with amplitudes higher than a given threshold are unconditionally considered as feedbacks. All components classified as resulting from feedback are eliminated by a notch filter. The algorithm guarantees that the louder a component and the faster it rises, the shorter time is required to detect and eliminate it.

The experiments carried out show that the algorithm engineered is able to detect and identify feedback components with a good accuracy. In a test sequence lasting 1.5 min that was infected with feedbacks, the algorithm detected 37 frequencies on which feedbacks occurred. In order to determine the effectiveness of classification, the algorithm was used to detect feedbacks in the original sequence (without feedbacks). As a result, only 3 false detections were identified. The amount of false positives can be further reduced through the observation of a new component for a few frames after notch filter deployment. If its amplitude decreases at an exponential rate then the classification is correct.

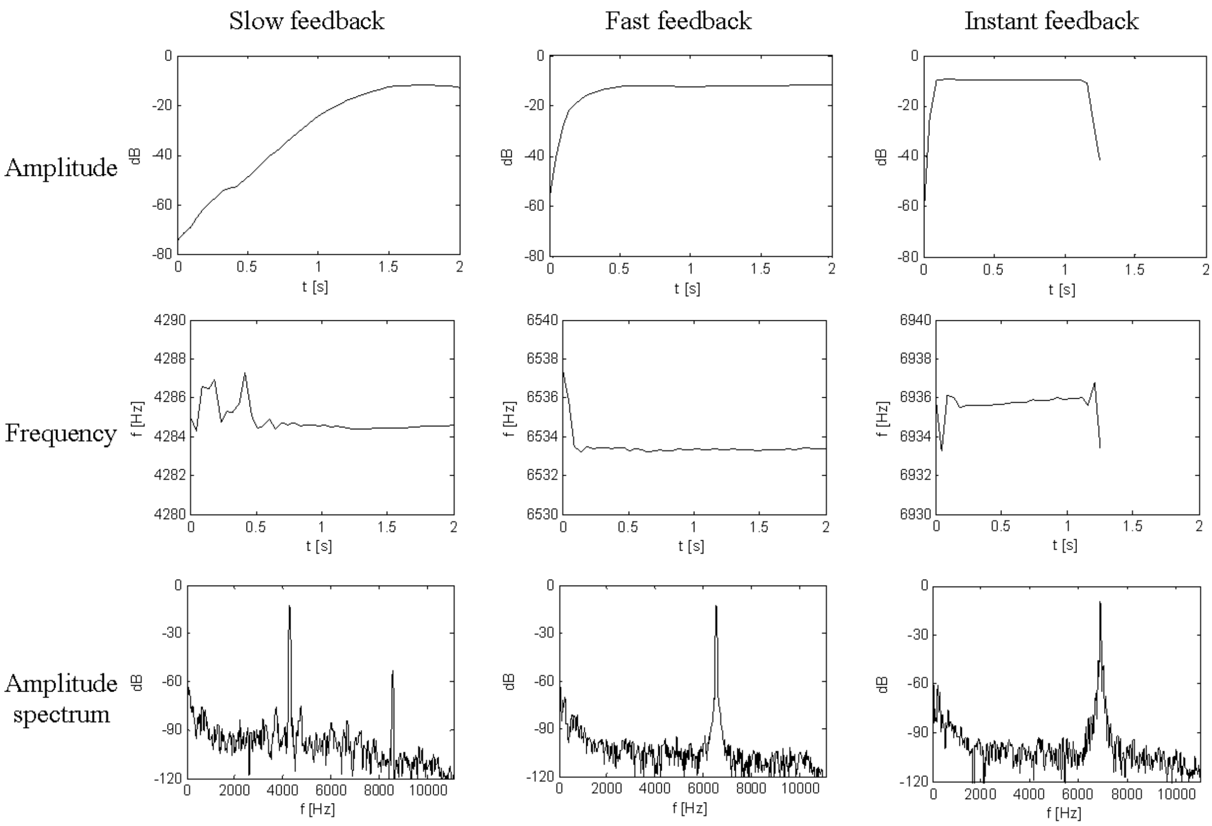

Figure 7. Examples of slow, fast and instant feedbacks

5. Implementation

This Section presents algorithm implementation details and describes the dedicated extension card for the Texas Instruments DSP development kit.

5.1 Hardware requirements

Since the contactless hearing aid is an autonomous device all algorithms responsible for signal processing must be implemented in the digital signal processor (DSP). In order to select the appropriate processor architecture it is necessary to declare the dynamics of the analog to digital and digital to analog conversion. Concerning the dynamic requirements for speech signal processing it was decided that the 16-bit precision is sufficient. In addition it was assumed that the device should be operable even if the power line is unavailable for some time (e.g. outdoor). In this case the power consumption of the DSP is the next issue that must be taken into account since it directly influences the battery life. Another important requirement is connected with processor peripherals. As stated in Section 2, the contactless hearing aid employs four microphones (and two miniature loudspeakers) in order to provide phase shifted signals into the spatial filtration module. Therefore, DSP must provide appropriate peripherals allowing for combining at least four A/D and D/A converters.

It is worth mentioning that the engineered algorithms described in the paper were evaluated in the Matlab simulator employing floating-point numbers representation. Thus, it seems at first that it would be reasonable to select the floating-point DSP architecture as well, since it might reduce the implementation effort. Unfortunately, the power consumption of the floating-point DSPs is significantly higher than the fixed-point ones and thus floating-point DSPs are not recommended for battery powered applications. Therefore, the 16-bit fixed-point architecture of DSP was found suitable for implementing contactless hearing aid algorithms. It has to be pointed out however, that in the case of the fixed-point DSP implementation the special care must be taken in order to maintain the robustness of the algorithms originally designed for floating-point processing.

During the selection of an appropriate DSP for the contactless hearing aid application various families of processors manufactured by Analog Devices (ADSP218x, ADSP219x), Texas Instruments (TMS320C54xx, TMS320C55xx) and Freescale Semiconductors (DSP56300) were taken into account [27][28][29]. Based on the requirements described above the Texas Instruments TMS320C55xx DSP family was chosen as the most suitable for the contacless hearing aid application. It has to be mentioned that the precise computational complexity of the algorithms incorporated into the contactless hearing aid is difficult to estimate before the implementation happens. That is why the possibility of altering the clock rate (108/144/200 MHz) determining the performance of TMS320C55xx is one of its advantages. Furthermore these processors provide also an advanced power management allowing extending the battery life. It was finally decided that all algorithms are going to be implemented using TMS320VC5509A processor.

5.2 Extension card design

In order to allow implementing and evaluating the algorithms the TMS320VC5509A development board manufactured by Spectrum Digital was employed [30]. It has to be pointed out that this system provides only one stereo audio codec, and thus it is impossible to evaluate the ANN spatial filtration algorithm. Therefore a dedicated extension card that extends the basic functionality of the development board was designed. The architecture of the development system consisting of Code Composer Studio environment, DSP development board and designed extension card is shown in Figure 8.

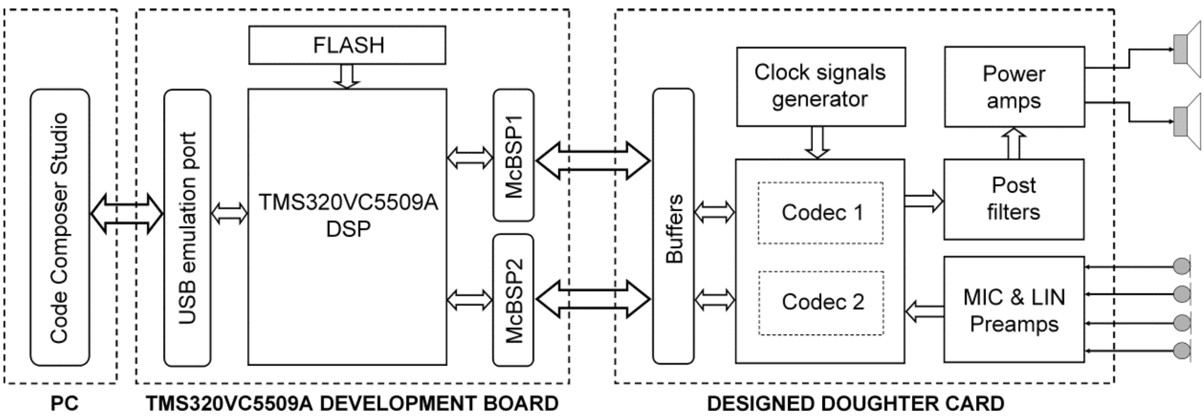

Figure 8: Block diagram of the contactless hearing aid development system

The main purpose of the extension card is enabling the A/D and D/A conversion of four autonomous signals and providing them to the digital signal processor. One of the extension card modules consists of the set of electret microphones and line levels preamplifiers. Although microphones can be plugged directly through the mini-jack ports (e.g. popular computer microphones), the dedicated port for microphone array module is also available. In addition, it is possible to capture the line level signals for evaluation purposes in every channel. The A/D conversion is obtained using two stereo audio codecs (PCM3008) that transmit and receive 16-bit audio samples over the I^2S serial interface [31]. Moreover, the sampling rate may be set to the one of the following values: 8, 16, 24, 32, and 48 kHz. Because the codecs can operate only in the slave mode, all necessary clock signals are generated locally. It is seen in Figure 8 that transmitted signals are buffered in order to prevent signal degradation. The physical connection between the extension card and the development board is accomplished using dedicated 80-pins peripheral slot [30].

Although TMS320VC5509A processor has three McBSP (Multichannel Buffered Serial Ports), only two of them are involved during the operation. The DMA (Direct Memory Access) controller of the processor is responsible for handling samples and feeding them to the appropriate buffers [32]. Then samples can be further processed according to the algorithms implemented with the Code Composer Studio environment. The similar scenario is utilized for transmitting processed samples from the DSP to the codecs incorporated in the extension card.

After D/A conversion signals are filtered employing low-pass, fourth-order, Butterworth filters in order to attenuate conversion artefacts. The extension card filters were designed with Texas Instruments Filter Pro application [33]. Relatively high order filters are required because codecs introduce significant distortion when the low sampling rate is chosen (e.g. 8 or 16 kHz) [34]. After the filtration analog signals are amplified using the D-class power amplifiers and are presented to miniature loudspeakers through chinch connectors.

The DSP development board along with a dedicated extension card is a complete prototype of the contactless hearing aid. Furthermore, the emulation link between the Code Composer Studio environment and development board allows for efficient implementation of the algorithms because of the availability of all debug functions. Finally, algorithms incorporated in the contactless hearing aid may be evaluated in real-life conditions which is essential for their proper tuning. The final prototype of the contactless hearing aid can be easily designed in future by supplementing the extension card with the DSP processor itself and wit a booting memory.

6. Conclusions

In this paper a novel contactless hearing aid dedicated to infants is presented and its structure is thoroughly described. The reduction of the acoustic feedback is a major issue in this study, thus two different approaches are implemented. The first one utilizes the beamforming method based on an artificial neural network. The second one employs direct signal processing algorithm and forms additional protection against feedbacks. Two such techniques (adaptive notch filtering and adaptive feedback cancellation) were studied and the former one was chosen for experiments because it is more suitable for digital signal processors implementation.

In the experiments the nature of feedbacks has been examined and the algorithm for feedback component discrimination has been implemented. The results of the experiments prove that the algorithm is able to detect feedbacks with a good accuracy and it may be implemented on a digital signal processor.

The paper also presents algorithm implementation details and describes the dedicated extension card that together with a Texas Instruments DSP development kit will form a complex environment for the four channel audio processing.

Future work will be focused on evaluating the hardware prototype within the context of contactless hearing aid device effectiveness.

Acknowledgements

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Employing Fuzzy Logic to Processing of Loudness Scaling Test Results

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Abstract: While fitting process of modern hearing aids hearing dynamics characteristics is required. The hearing dynamics characteristics is calculated on the basis of loudness scaling test results. The problem is that the loudness scaling test results are represented on a loudness category scale, but a hearing aid requires numerical parameters. A fuzzy logic method is the one of the simplest artificial intelligence methods which is useful for processing parameters expressed in a human natural language. In this article the fuzzy system for loudness scaling result processing is presented. Additionally, the method for shortening loudness scaling test is explained.

1. Introduction

The problem of hearing aid fitting includes many issues, but the most important one is the gain control problem. Generally, a hearing aid fitting process can be described as scaling of a wide-ranged dynamics of speech to a narrow-ranged dynamics of impaired hearing. To solve this problem, most of hearing aids use dynamics processors such as compressor and expander (fig. 1.1)[1][2][3].

To obtain a hearing dynamics (HD) characteristics, loudness scaling test (LST) results are needed. To asses loudness level natural language is used, but a hearing equipment requires parameters on a numerical scale. For this reason human natural language processing methods are required. The processing methods usually base on artificial intelligence. Fuzzy logic is the one of the simplest artificial intelligence methods, which is especially useful in converting parameters expressed in a natural language to proper parameters on a numerical scale [4][5][6].

The article is dedicated to the use of fuzzy logic to *LST* results processing. When developing the fuzzy logic system (FS) the way to shortening of *LST* was elaborated. The details are presented in the following paragraphs.

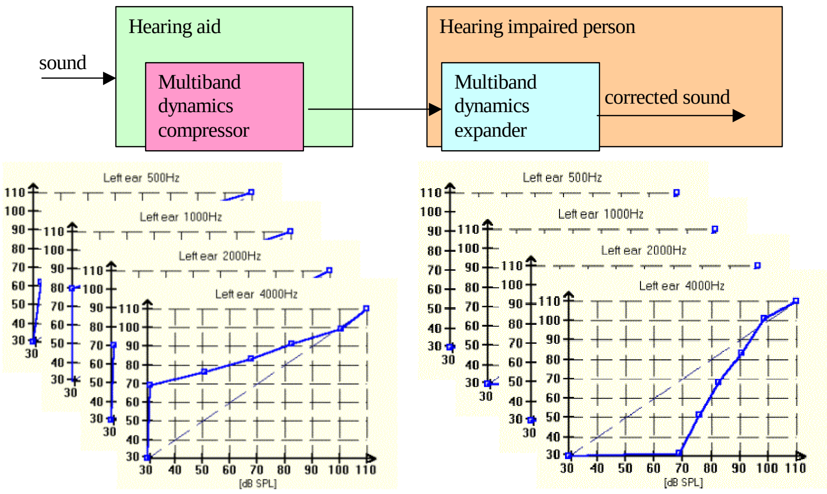

Fig. 1.1. Hearing compensation model

2. The loudness scaling test rules

In clinical practice several types of *LST* are used. For the developed method was chosen a well-known LGOB method (*Loudness Growth in ½ Octave Bands*). The LGOB rules are simply and easy to implement. Moreover the LGOB principles are similar to other commonly used *LST* like e.g the WHF (Wurzburger Horfeld) method, but from the author's point of view the LGOB test is easier to understand [7][8][9][10].

During the LGOB test a patient is listening to test signals which loudness level is changing randomly. The patient has to asses loudness sensation for each test signal using the seven point loudness category scale:

0. Nothing (the patient can't hear),

1.Very soft (the patient only detects a sound, but a loudness level is too low),

2.Soft (the patient can hear a sound, but she/he would like to listen to a louder sound),

3.Comfort (MCL) (a loudness level is comfort for the patient),

4.Loud (the patient can hear a sound, but she/he would like to listen to a lower sound level),

5.Very loud (the patient would like to definitely listen to a much lower sound level),

6. Too loud (a sound level makes pain).

Test signals are in the white noise signal form filtered in four half octave bands with the middle frequencies: 500 Hz, 1000 Hz, 2000 Hz, 4000 Hz (fig. 2.1). Each test signal duration is 3 seconds and the signal's form is presented in fig. 2.2.

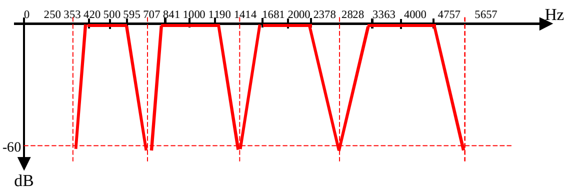

Fig. 2.1. Half octave band filters characteristics

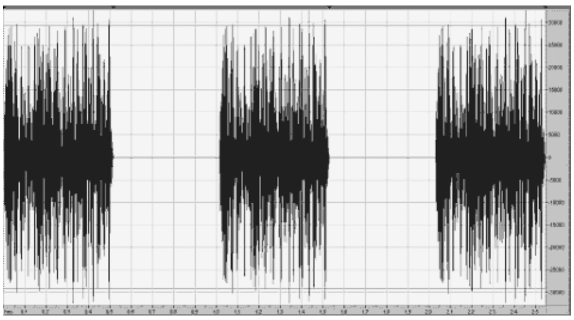

Fig. 2.2. The form of the test signal

The *LST* contains two phases. The first phase is a training phase. During this phase the hearing threshold and the uncomfortable level is measured. During the second phase test signals are presented in a random order. Each test signal is presented at least three times. Patient's responses are collected for each test signal. When the test is finished, a loudness scaling characteristics is determined. First, the median value of collected responses values for each test signal is collected. Next, for given frequency band the obtained median values are approximated by a curve. The obtained curves (for each examined frequency band and for each examined ear) describe loudness scaling characteristics (fig. 2.3).

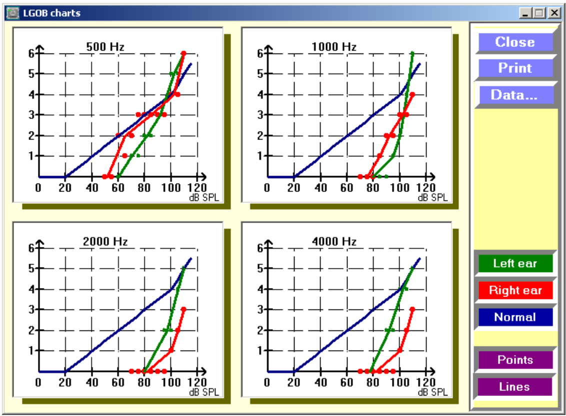

Fig. 2.3. Example of loudness scaling characteristics

Based on the LGOB principles a computer *LST* was implemented. The application works on PC computer and requires a 16 bit sound card. For examination, an earphones (e.g. ER 3A) and sound level calibration unit is also required. In this case the examination dynamic range is 30 – 110 dB SPL. The implemented *LST* was verified on normally hearing (NH) and hearing impaired (HI) persons and the obtained results were compared with the results obtained for the same persons using a dedicated ReSound P3 device. Although the ReSound device has wider dynamics range (20 – 120 dB), the obtained results were compatible [11][12].

3. The fuzzy logic system for loudness scaling processing

The *LST* results are presented as a loudness category scale vs. the input loudness level, but a hearing aid amplifier requires a characteristics in the form of the output loudness level as a function of the input one (fig. 3.1). It means that converting *LST* results from a loudness category scale to a numerical decibel scale, which output loudness level represents, is needed. In order to do this, the *FS* was designed.

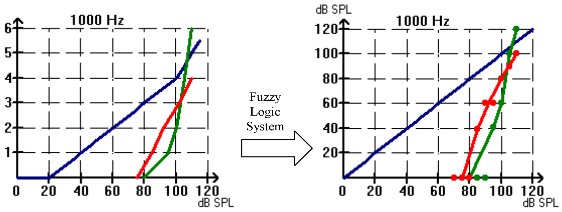

Fig. 3.1. The idea of obtaining a hearing dynamics characteristics.

The most suitable method for a converting category scale to numerical values is a fuzzy processing [5][6][12]. The developed *FS* requires data as follows:

· loudness scaling results for *NH* people – presented as input membership functions,

· loudness scaling results for a given patient – each loudness scaling result represents a vector of three parameters (test signal loudness level, frequency band, selected loudness category),

· knowledge concerning interpretation of difference between results obtained for the examinee and *NH* people (a fuzzy logic rules base).

· output membership functions, which describe all possible differences between the examinee and *NH* people loudness scaling results.

One of the most important information for the developed *FS* is the knowledge of *LST* results for *NH* people. This knowledge is represented in the *FS* by input membership functions. Because seven loudness categories are used in the *LST*, seven fuzzy sets per each frequency band were defined. Each fuzzy set is described by one membership function. In order to obtain these functions, about 51 *NH* students were examined. The details of obtained membership functions are presented in paragraph 4.

Comparing the *LST* results on a seven point scale, 13 kinds of differences can be defined:

· six positive differences (when the current *LST* result is greater than the proper result for a *NH* person),

· six negative differences (when the current *LST* result is smaller than the proper result for a *NH* person),

· no difference (when the current *LST* result is equal to the proper result for a *NH* person).

Both the negative and positive difference means hearing impairment, but in addition - positive differences usually denote a loudness recruitment problem. On the basis of these differences thirteen output membership functions were created (fig. 3.2) [13][14].

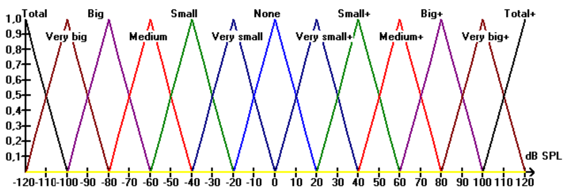

Fig. 3.2 Output membership functions

An appropriate fuzzy processing based on rules was defined in the fuzzy rules base (FR). A typical form of the *FR* base is as follows:

IF <input$_1$> AND <input$_2$> AND … AND <input$_n$> THEN decision

In this case two input variables were defined:

· *Norm* - represents the *LST results* for *NH* person,

· *Exam* - represents the *LST results* for the currently being examined person.

Because each variable uses the same loudness category scale, the variable which represents an examinee, uses labels written in capital letters. In the developed *FS* the simply *FR* base was defined (fig. 3.3). If for the same test signal *LST* results for the examinee and *NH* people are the same, the *FS* activates a rule with the output label "*no difference*". If such a difference is equal to one category, the *FS* activates a rule with the output labelled as „*very small*". If the difference equals two categories, the *FS* activates a rule with the output labelled as „*small*", etc. Experts from The Institute of Hearing Physiology and Pathology in Warsaw checked the rules.

The examiniee

	I DON'T HEAR	VERY SOFT	SOFT	MCL	LOUD	VERY LOUD	TOO LOUD
I don't hear	None	V.small+	Small+	Medium+	Big+	V.big+	Total+
very soft	V.small	None	V.small+	Small+	Medium+	Big+	V.big+
soft	Small	V.small	None	V.small+	Small+	Medium+	Big+
mcl	Medium	Small	V.small	None	V.small+	Small+	Medium+
loud	Big	Medium	Small	V.small	None	V.small+	Small+
very loud	V.big	Big	Medium	Small	V.small	None	V.small+
too loud	Total	V.big	Big	Medium	Small	V.small	None

Normal Hearning

Fig. 3.3. The rules base

As *LST* results are collected, the *FS* is ready for calculation of *HD* characteristics. As was mentioned, each *LST* result is represented by a three parameter vector which contains: test signal level, frequency band and loudness category, all given for this signal. In the ***FuzzyLGOB*** block (fig. 3.4), each *LST* result is processed as follows:

· First, the *FS* executes fuzzyfication process. It means that numerical input data are converted to a category domain. In this case, the test signal level and frequency band parameters are fuzzyfied according to the obtained input membership functions which describe the normally hearing loudness scaling results. The third parameter – the loudness category has already been in the fuzzy domain, but from the formal point of view it has to be fuzzyfied, too. For this case, the system uses fuzzyfication based on so called singletons [4][5][15].

· After the fuzzyfication process, the system starts to process rules. For each activated rule, the system calculates the rule strength.

· Next, these strength values for the rules activate adequate output membership functions and in result an output function is created (fig. 3.5)

· Finally, the system starts the defuzzyfication process, in which the centre of gravity for the created output function is calculated. The obtained value represents the difference on the dB scale between the hearing scaling results for the *NH* person and the examinee. The output level value is computed as the sum of the input test signal level and the obtained difference.

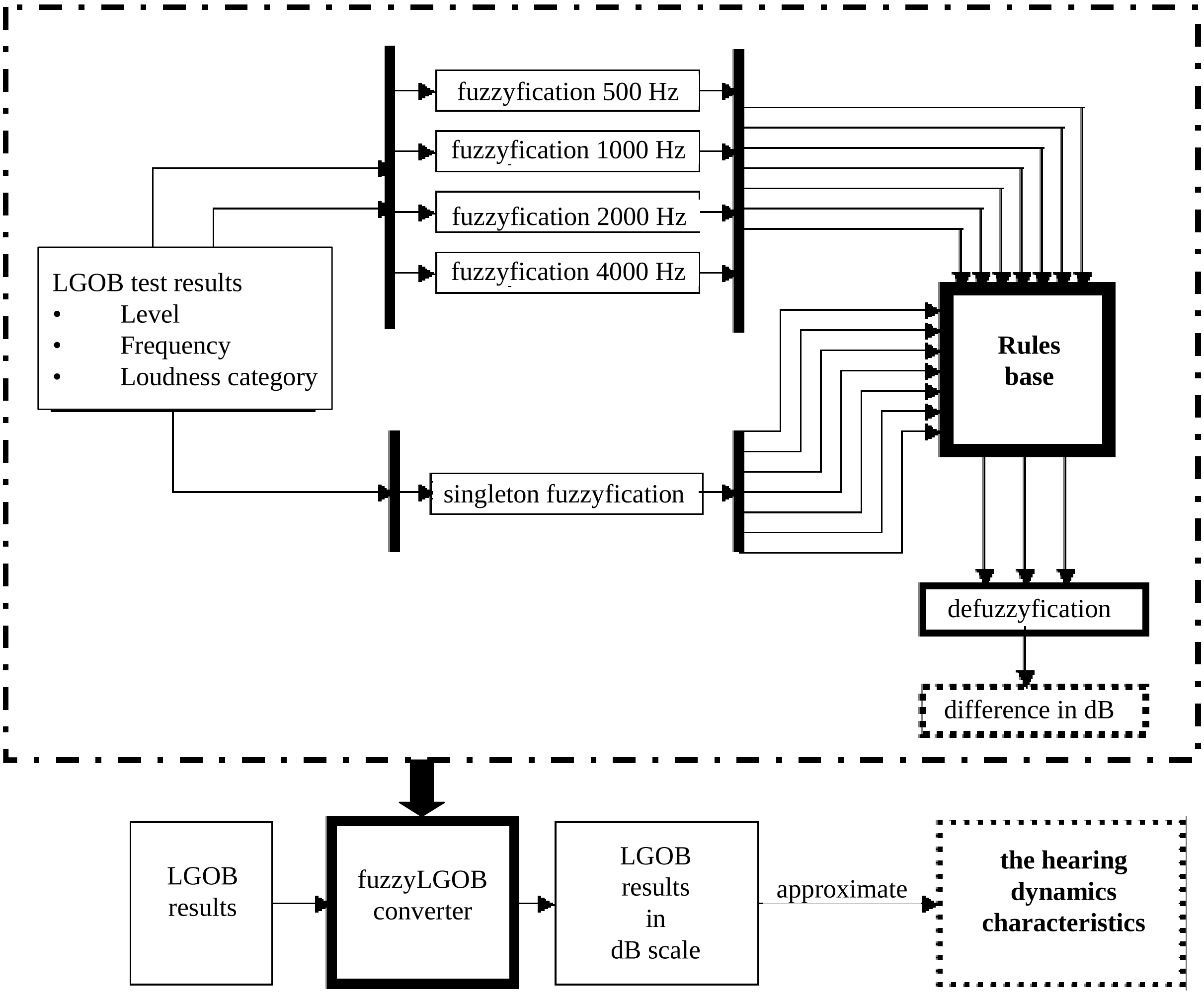

Fig. 3.4. The block diagram of the designed fuzzy system

Next, the system collects the processed *LST* results and creates an expected characteristics of the *HD* by approximating these results.

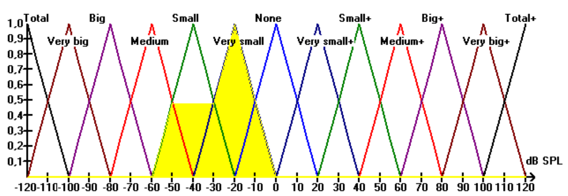

Fig. 3.5. An example of obtained output function

4. The estimation process of membership functions

In the fuzzy logic, membership functions usually assume a triangular or trapezoidal shape, rarely a sigmoidal shape or Gauss curves. A *FS* with triangular or trapezoidal functions is easier to analyse and implement [4][5].

Typically, membership functions are created on the basis of expert's knowledge. The knowledge can be explored directly from experts or indirectly on the basis of statistical analysis results. In the case of statistical analysis a statistical distribution has to be assumed first. If the distribution is unknown the normal one is assumed. Next, the minimal number of tests has to be estimate. To order this the Pearson's statistics (4.1) was calculated. As the results of Pearson's statistics it was estimated that at least 40 persons are required for examination. The Pearson's test can be useful in astatistical hypothesis verification process only if for each j Npj>=10 or nj>=5 [16].

$$\chi^2_{k-1} = \sum_{i=1}^{k} \frac{(n_i - N \cdot p_i)^2}{N \cdot p_i} \qquad (1)$$

where:

ξ - random variable;

F – distribution of a random variable

$F(a) = 0,\ F(b) = 1,\ a = a_0 < a_1 < a_2 < \ldots < a_k = b$ and $p_i = \mathcal{P}\{a_{i-1} < \xi <= a_i\},\ i=1,2,\ldots$

n_i = number of elemnts ξ, which fulfill the condition $a_{i-1} < \xi <= a_i$;

$p_i = \int_{r_i}^{r_{i+1}} p(x)dx$ - impact probability of random variable X in the interval i;

r_i – the minimum value in the interval i;

N – total number of observations.

During the *FS* developing 51 *NH* students were examined. The examined group consisted of persons in the ages twenty – twenty five, who had been experienced in subjective tests. The results obtained with the examined group are shown in fig. 4.1.

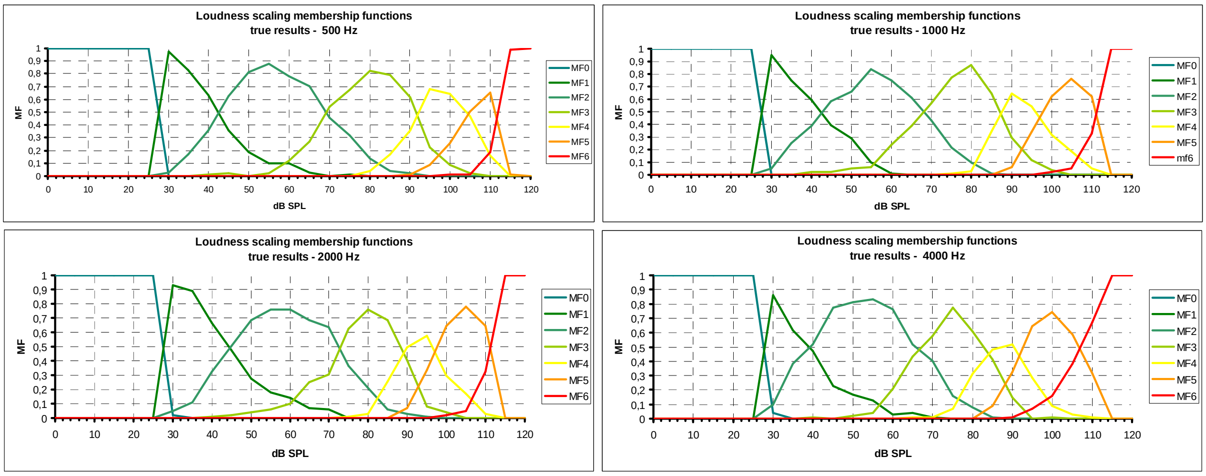

Fig. 4.1. The obtained membership functions which describe a loudness scaling for normally hearing persons

The next step in the course of the *FS* development was to find the shape of membership functions, which in the best way approximates *LST* results. Very important assumption was that membership function keeps features of the probability density function (4.2). During the calculation process two kind of functions were taken into consideration: the Gauss function (normal distribution function)(4.3) and trapezoid functions (4.4) [5][15].

$$\int_{-\infty}^{+\infty} p(x) = \int_{a}^{d} p(x) = 1 \qquad (2)$$

where $p(x)$ – probability density function

As was mentioned, the normal distribution function is assumed if the proper one is unknown, however the Gauss function never reaches the zero value. In the *FS* case, it means that each element belongs to all fuzzy set (all fuzzy sets overlap each other). In consequence, a *FS* works more hard, because each time every rule is activated and thus a defuzzyfication procedure is more complicated. The example of obtained membership functions in the Gauss function form is shown in fig. 4.2.

$$p(x) = \frac{1}{\sqrt{2\pi \cdot \sigma}} \cdot \exp\left(-\frac{(x-m)^2}{2 \cdot \sigma^2}\right) \qquad (3)$$

where:

σ - dispersion

m – average value;

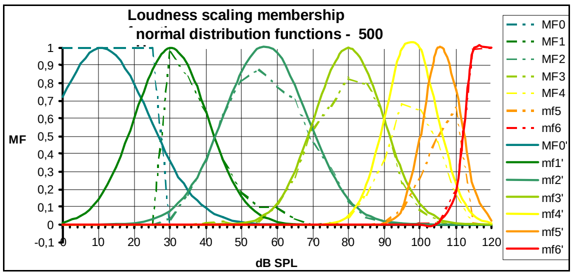

Fig. 4.2. Example of membership functions approximation with normal distribution functions

The most expected membership function shape is trapezoid. This kind of function requires only four parameters (trapezoid vertexes). In a particular case trapezoid can be reduced into a triangle. Examples of obtained membership functions in the trapezoid form are shown in fig. 4.3.

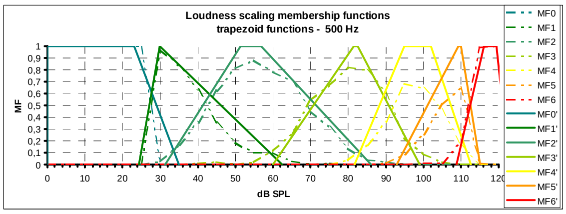

Fig. 4.3. Example of membership functions approximation with trapezoid functions

When comparing the Pearson test results for the Gauss shape and trapezoid shape of membership functions it can be concluded that trapezoid membership functions are better than Gauss functions. Table 4.1 contains some exemplary of the Pearsons test results for trapezoid and Gauss functions. A smaller value of the Pearson test result denotes a better approximation result.

Table 4.1 The Pearson test results for Gauss and trapezoid membership functions – 1000 Hz

loudness category	„0”	„1”	„2”	„3”	„4”	„5”	„6”
Trapezoid function	0,124	3,66	1,807	2,169	0,546	0,147	0,001
Gauss function	29,818	179,513	11,532	10,393	11,051	22,568	0,241

5. Results of the loudness scaling test based on fuzzy logic processing

The first step in the verification process of the designed method for *LST* results processing was to check correctness of the obtained *HD* characteristics. In this case, the obtained characteristics were compared with the characteristics calculated with standard method based on an average *LST* results for *NH* people. The comparison has shown that the *HD* characteristics obtained in the basis of the designed ***fuzzyLGOB*** converter are in accordance with expected results, which is shown by an example in figure 5.1 and table 5.1.

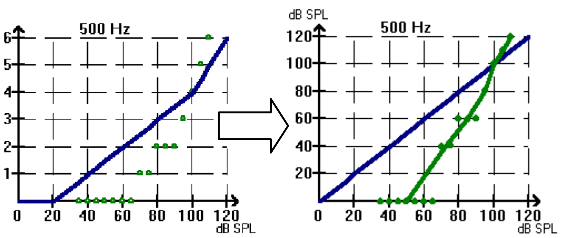

Fig. 5.1. An example of loudness scaling test results and it's corresponding hearing aid dynamics characteristics

Table 5.1. The comparison of loudness scaling test results processing in the standard method and in the fuzzyLGOB converter

Test signal level [dB]	The standard method [dB]	The fuzzyLGOB converter [dB]
35	20	20,79
40	20	20
45	20	19,21
50	20	20
55	20	20,79
60	20	20
65	20	19,21
70	40	40
75	40	40,79
80	60	60
85	60	59,21
90	60	60
95	80	80,79
100	100	100
105	110	110
110	120	120

The main inconvenience of the *LST* is its long duration. On the assumption that:

- each test signal duration approximates 3 seconds,
- number of frequency bands is 4,
- number of trials is 3,
- test signal amplitude changes from 30 dB SPL to 110 dB SPL with 5 dB step,

The average test duration can be even 10 minutes per ear (eq. 5.1). Too long test duration makes a patient tired and has bad influence on results. This was the reason to find out the way to shorten the test duration and to save results reliability.

$$T = f \cdot p \cdot t \cdot \left(\frac{L_{MAX} - L_{MIN}}{s} + 1 \right) \qquad (4)$$

where:

T – total time of *LST*
f – number of bands,

p – number of trials,
t – test signal duration,
L_{MAX} – maximum level [dB],
L_{MIN} – minimum level [dB],
s – amplitude changing step;

The most intuitive idea is to cut down the number of test signals. In this case, could be selected only the one characteristic test signal per each loudness category. In this way during the test could be used only seven different test signals per each frequency band. In this case the test duration could be shorten to about 4 minutes per ear (more than twice shorter) what is presented by expression 5.2.

$$T^{'} = f \cdot p \cdot t \cdot K \qquad (6)$$

where:
T' – total time of the shortened *LST*,
f – number of bands,
p – number of trials,
t – test signal duration,
K – loudness categories number

The test signal level which corresponds to the membership function maximum value is a good candidate for a selected test signal (fig. 5.2). It means that loudness level of these test signals were distinguish in the best way by *NH* people.

The designed shortened loudness scaling test (sLST) was verified. In the verification process took part group of twenty normal hearing students and three *HI* persons. First the normal hearing students have been examined with the standard *LST* and a few minutes later each of them done the shortened *LST*. During the tests time was measured. Comparing test results it can be observed that obtained loudness scaling characteristics in shortened mode are very similar to results gathered in standard mode (fig. 5.3). The table 5.2. shows examples of measured time during *LST*. The times achieved in shortened mode of *LST* are more or less twice shorter than in standard mode.

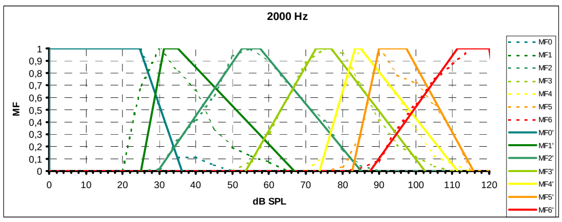

Fig. 5.2. An example of selected test signals for shortened loudness scaling test.

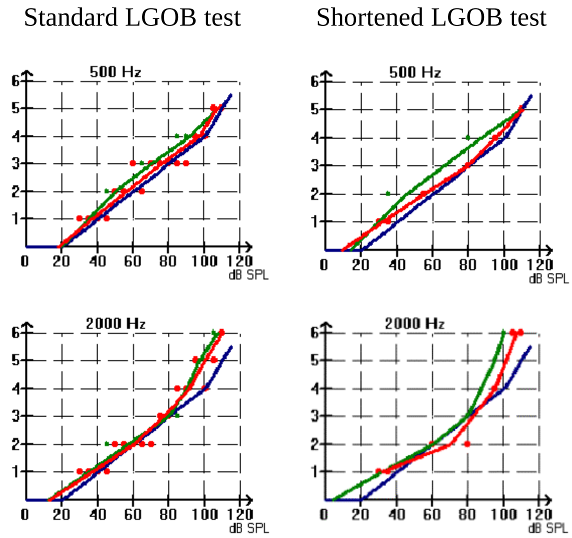

Fig. 5.3. Examples of loudness scaling results obtained in standard and shortened mode for normal hearing students

Table 5.2. Examples of loudness scaling tests duration

No	Standard LGOB test [min]	Shortened LGOB test [min]
1	9,5	6,5
2	8,5	4,5
3	7	4
…	…	…
20	9	5,5

In the *HI* person case, each type of *LST* was repeated several times. Aim of these examinations was to check how is changing the loudness scaling characteristics for the patient in period time (on average about two weeks). Figure 5.4 presents examples of *LST* results collected within two weeks. The solid lines represent results obtained in the standard mode and the dashed lines represents results obtained in shortened mode. It can be observed that the loudness scaling characteristics obtained in standard mode is changed in the similar way as the one obtained in shortened mode.

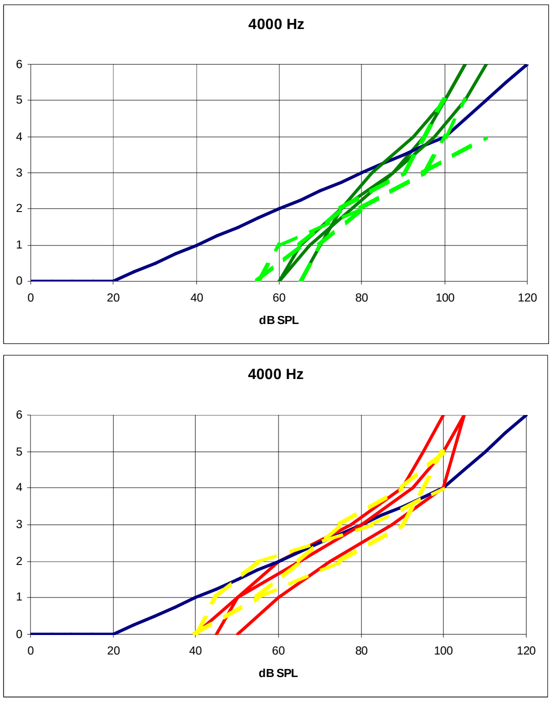

Fig. 5.4. Examples of loudness scaling results obtained in two loudness scaling test modes for the same hearing impaired person, the solid line – standard test LGOB, the dashed line – shortened LGOB test.

In this case test time was also measured (tab. 5.3). The proportion between duration of the standard mode and the shortened mode is the same as in the normal hearing people case. It can be observed that *LST* in *HI* person case consumes less time, because of narrower examination dynamics range.

Table 5.3. Examples of loudness scaling tests duration in the hearing impaired person case

No.	Standard LGOB test [min]	Shortened LGOB test [min]
1	5,5	3
2	7,5	3,5
3	6	3

6. Conclusions

Utilizing fuzzy logic for processing of *LST* results it has been approached two purposes:

- flexible and effective method for calculating *HD* characteristics was designed,
- the shortened LGOB test was created.

The designed fuzzyLGOB converter allows calculating precisely a *HD* characteristic which is useful for both making hearing impairment approximate simulation and for hearing aid fitting. From formal point of view utilizing the artificial intelligence method for processing of loudness scaling test results, expressed in loudness categories, is more correct than approximate calculation of *HD* characteristics in the basis of average results for *NH* people. The designed method is easy to implement and can be update to other frequency bands or different loudness category scale.

Additionally in the basis of statistical analysis of loudness scaling results for *NH* people it has been became possible to shorten the *LST* duration more or less twice. It means that in much shorter time it is possible to obtain the reliable *HD* characteristics.

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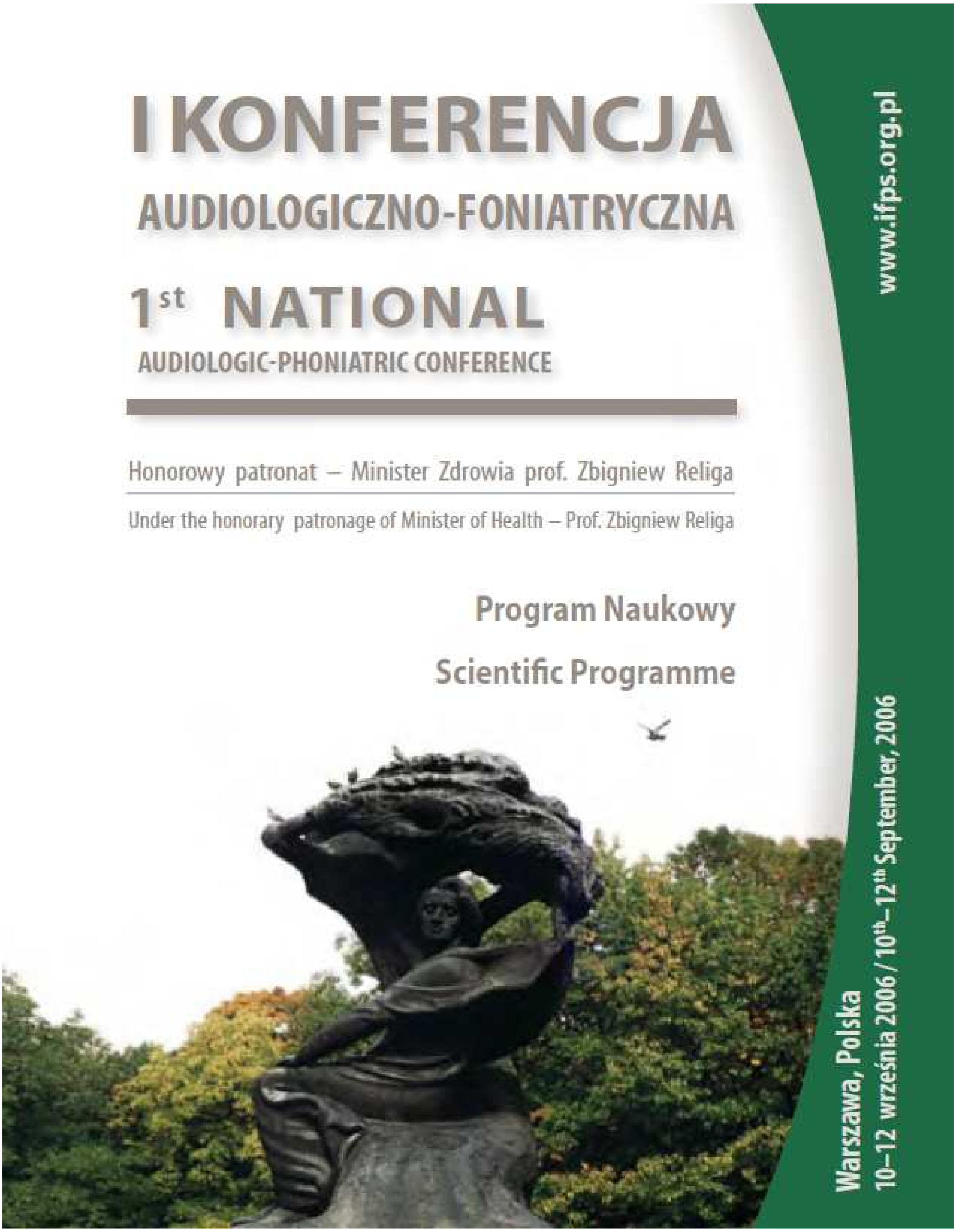
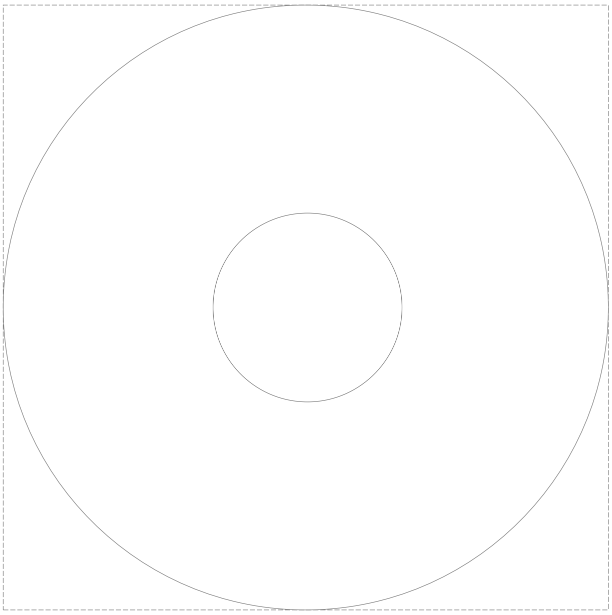


5th Plenary Session Video Report

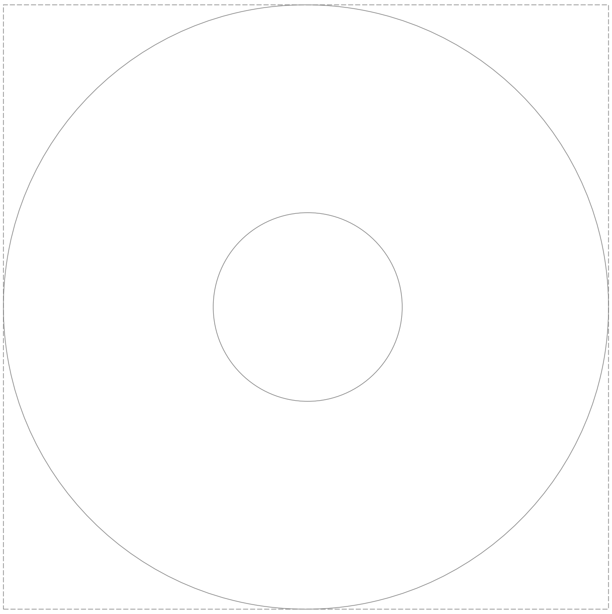
MONDAY September 11th 2006
PLENARY SESSION V / Rooms B, C
15:45 – 17:45
Chairpersons: Czyżewski A., Durrant J., Cohn E., Kayser K., Kostek B.

Czyżewski A., Kotus J. (Poland), *Intro to the Session* plus *Demo on "Distance monitoring of environmental noise impact on hearing"*.

Fagelson M., Bartnik G. (USA, Poland), *"Tinnitus cases remote consulting" – teleconference Poland – USA.*

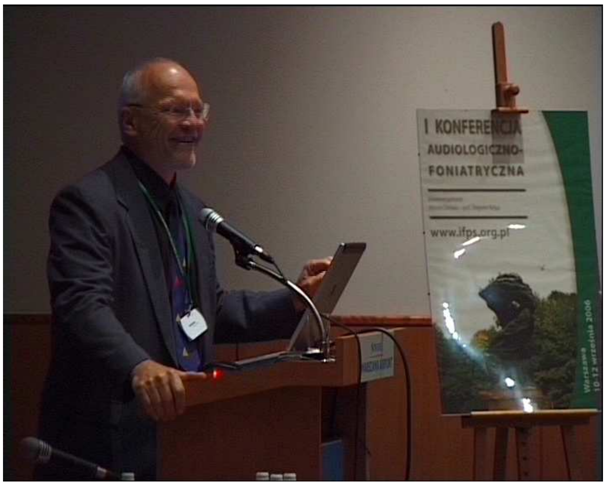


Czyżewski A., Kotus J. (Poland), *Intro to the Session* plus *Demo on "Distance monitoring of environmental noise impact on hearing"*.



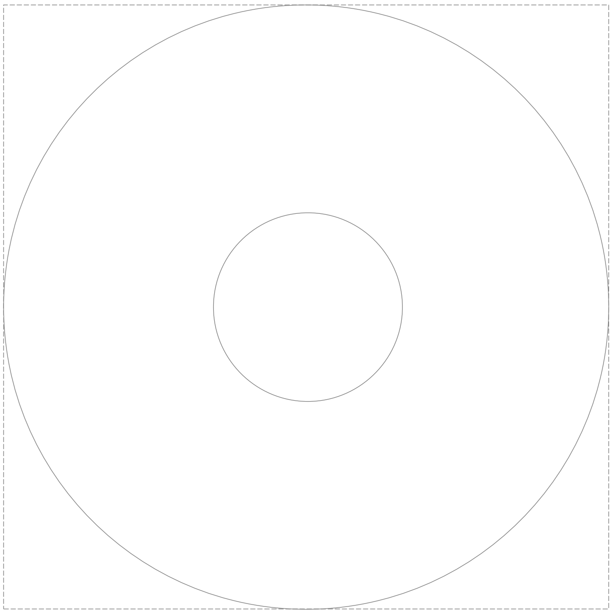
Fagelson M., Bartnik G. (USA, Poland), *"Tinnitus cases remote consulting" – teleconference Poland – USA.*

Kayser K. (Germany), *Applications of multimedia in computational diagnostic pathology*.



Moser L. (Germany), *Accuracy and precision of audiometric data, a statistical evaluation of more than 50000 audiograms by teleprocessing. A direct way to quality control in audiometry*.

Kayser K. (Germany), *Applications of multimedia in computational diagnostic pathology*.



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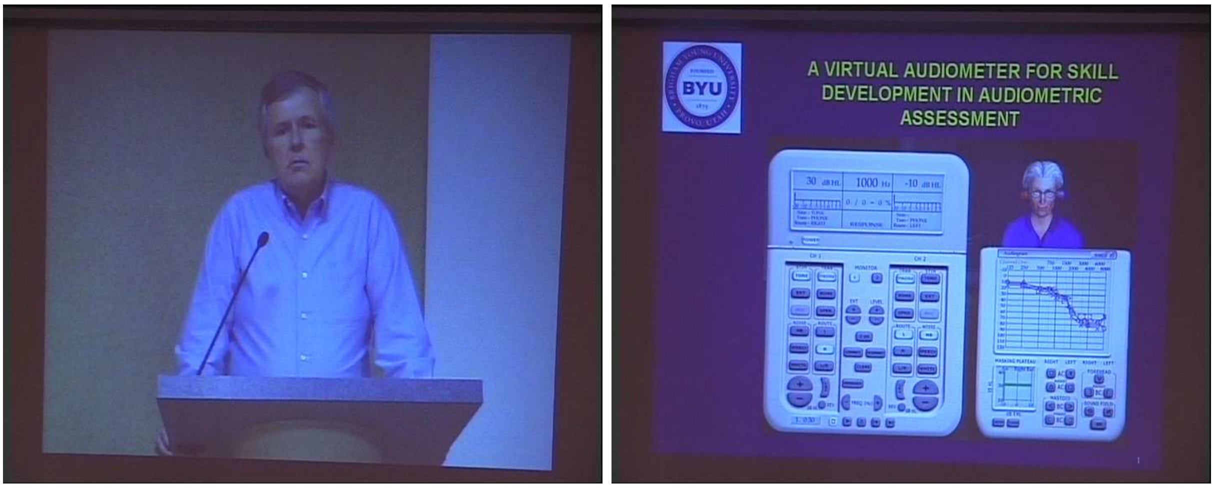
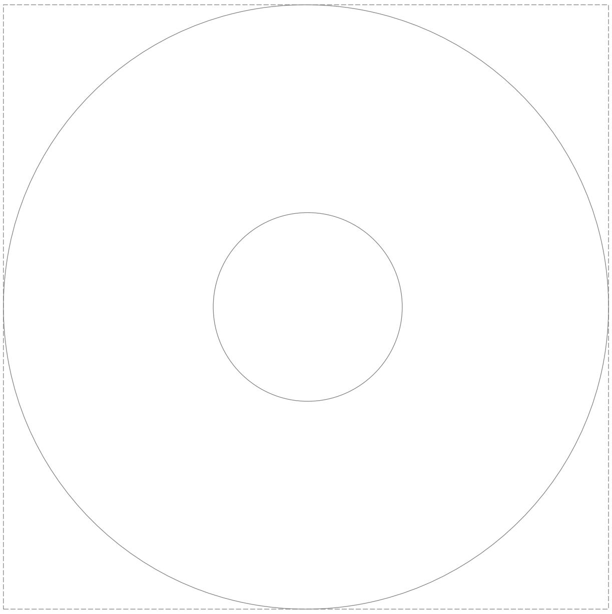


McPherson D., Harris R. (USA), *A virtual audiometer for education, training and skill development*.

Skarżyński H. (Poland), *Teletransmission of otosurgery*.



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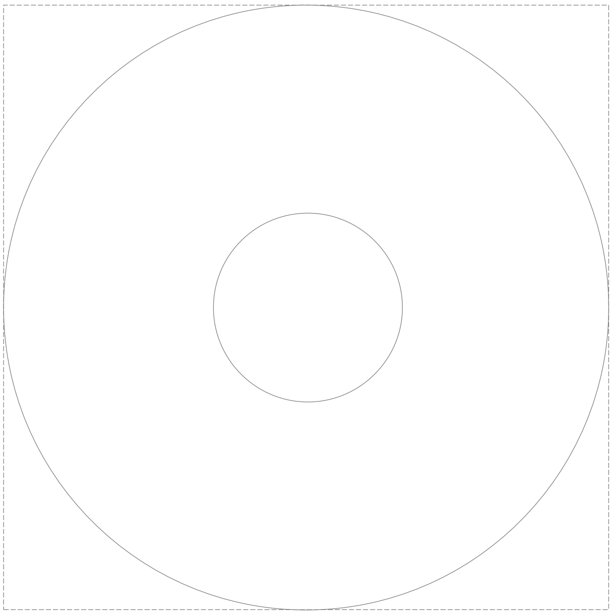

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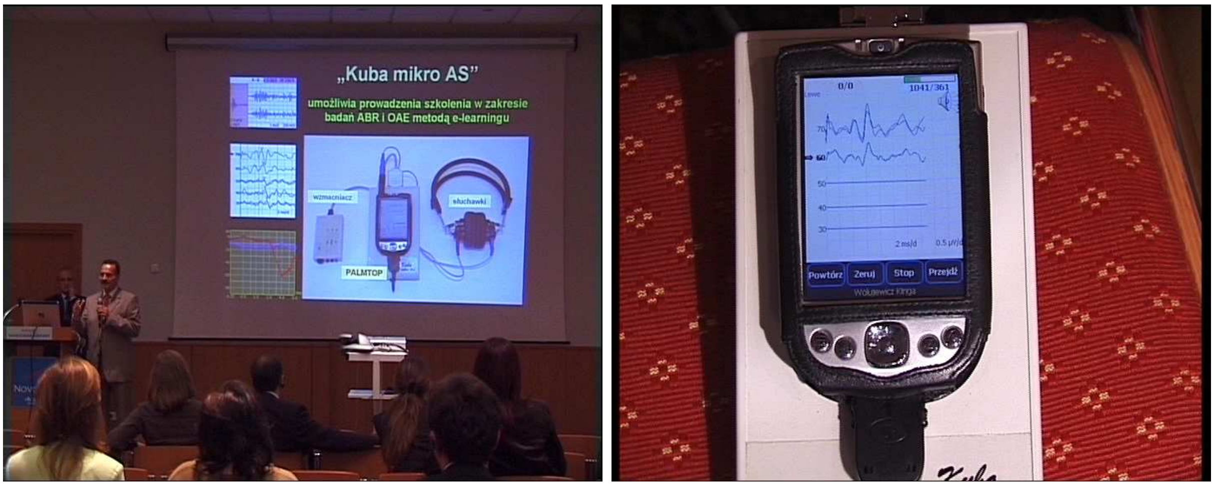
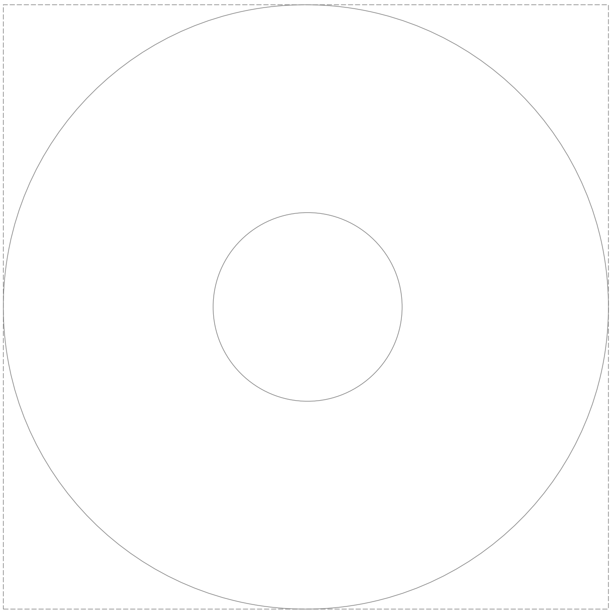


Kochanek K., Zając J., Pietraszek S., Piłka A., Skarżyński H. (Poland), *The possibilities of distant learning in ABR and OAE testing methodology by means of the Kuba Mikro AS device*.



Kochanek K., Zając J., Pietraszek S., Piłka A., Skarżyński H. (Poland), *The possibilities of distant learning in ABR and OAE testing methodology by means of the Kuba Mikro AS device*.

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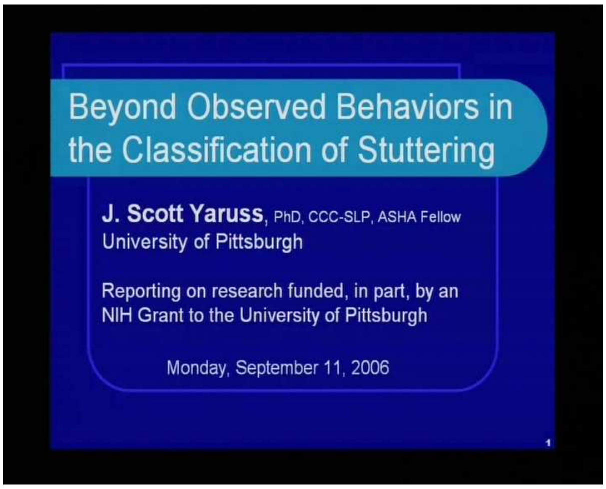

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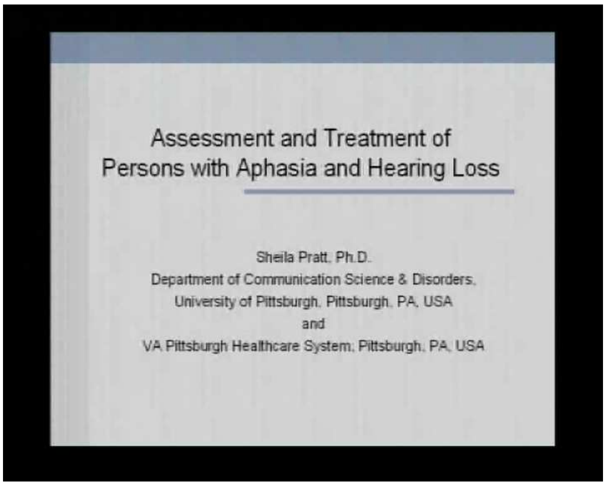

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Yaruss J. S. (USA), *Beyond observed behaviors in the classification of stuttering*.

Pratt S. (USA), *Assessment and treatment of aphasics with hearing impairment*.

McCue M. (USA), *Tele-rehabilitation project-up-date*.

McCue M. (USA), *Tele-rehabilitation project-up-date*.

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