EVALUATION OF DIGITAL HEARING AID ALGORITHMS ON WEARABLE SIGNAL PROCESSOR SYSTEMS

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ABSTRACT

The benefit of hearing aid algorithms in everyday life can hardly be estimated from results obtained in the laboratory. Extensive field tests with many hearing impaired subjects are necessary to evaluate these processing schemes. A wearable digital hearing aid prototype is described which was developed specifically for that purpose. It is based on a fixed-point digital signal processor. This unit enables the testing of even highly sophisticated algorithms, with a changing interval of the accumulator pack of 10 hours. As application examples, a very flexible three channel dynamic compression algorithm and a binaural processing scheme for enhancing speech signals in noisy and reverberant environments are described. Application of 20 units in 3 European clinics has been started recently.

1   INTRODUCTION

Several signal processing algorithms for digital hearing aids have been published in the last few years which aim at better fittings for sensorineural hearing impaired persons [2, 4, 9]. To overcome the limitations of analogue and sampled-analogue hearing aids, these algorithms must necessarily be complex and sophisticated. Thus, a high computational performance is necessary, which up to now could be provided only by stationary multi-processor equipment. Therefore, only laboratory tests, in which only few subjects participated, were possible. However, it is not yet proved in which and how many cases hearing impaired persons can actually gain a significant alleviation of their everyday hearing problems from these algorithms. To clarify this, extensive field tests are necessary. The availability of digital signal processors (DSP) with high computational performance now enables evaluations of even highly sophisticated digital hearing aid algorithms. For wearable units, however, it is advisable to use DSPs with fixed-point

arithmetic to minimize the size and weight of the units (by keeping system complexity and battery volume low). Numerical restrictions of the fixed-point architecture do not pose a remarkable practical problem if a processor with a large word length, e.g. the 24 bit *Motorola* DSP56002, is used. These limitations, however, are a realistic anticipation of the problems which arise when DSPs for cosmetically attractive hearing aids are developed. The fact that a hearing aid DSP can only be supplied by a single button cell limits processing speed and complexity considerably [3].

This paper presents a wearable digital signal processor system. The following section gives a short overview of the hardware components and their features. As a first application, a binaural dynamic compression algorithm with three frequency channels was implemented. On principle, this processing scheme is like that used in the *Siemens* hearing aid TRITON3000 [7], but much more flexible. This algorithm is described in section 3. Section 4 sketches another processing scheme which enhances speech signals by suppressing lateral sounds and by removing reverberation components. The clinical evaluation is subject of the last section.

2   HARDWARE DESIGN

In order to limit size and weight of the device, the fixed-point DSP *Motorola* DSP56002 is used. This DSP employs a 24 bit arithmetic unit. The high precision of 24 bit computation seems to be sufficient for all audiological applications. The DSP56002 is embedded in a complete stand-alone system, the miniKit56002 from *Electronic Tools GmbH*, Ratingen, Germany. This credit card sized system includes the required data and program memory and a serial interface which is used here for connecting the hearing aid prototype to a PC. It offers high computational performance (20 MIPS at 40 MHz clock) at an acceptable power consumption level. Fig. 1 shows a block diagram of the wearable digital hearing aid.

The FLASH-EPROM on the miniKit module can hold up to four different algorithms and parameter sets, fitted for different acoustical situations, e.g. quiet surrounding or traffic noise. The hearing aid wearer can select that programme which ensures him best possible hearing comfort in the actual acoustical situation, by means of a switch on the signal processing unit.

Sound pick-up and reproduction takes place in two analogue hearing aids, either behind-the-ear (BTE) or in-the-ear (ITE) units. The analogue hearing aids are modified to connect them to the A/D converter and the D/A converter in the signal processing unit. A combined single-chip stereo A/D and D/A converter, the audio codec CS4215 (*Crystal Semiconductor Corp.*, TX, USA), acts as the interface to the analogue domain. This codec provides 16 bit stereo samples at sample rates up to 48 kHz. Measurements of the complete hearing aid unit showed a dynamic range of 82 dB for sinusoidal signals, equivalent to 14 bits of useable resolution of the digital signal.

The digital hearing aid is supplied from a 6 V accumulator pack with a capacity of 3.4 Ah. This enables an operation time of approximately 10 hours, depending on the sampling frequency and the computational complexity of the algorithm. In case of the dynamic compression algorithm described below, the supply current is 320 mA at a sampling frequency of 22.05 kHz.
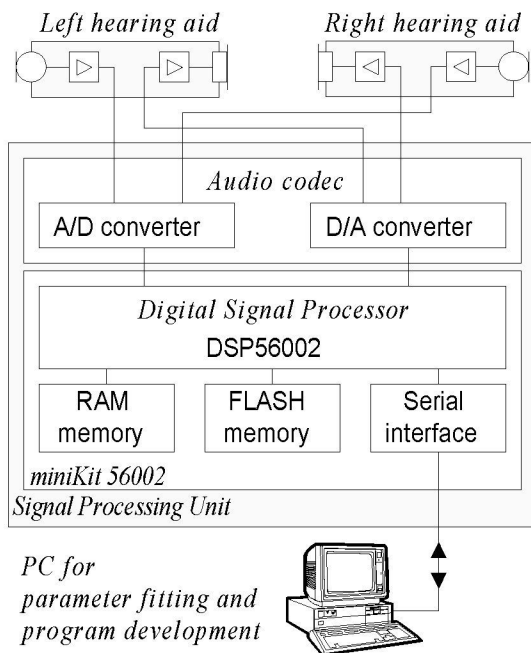

Fig. 1: Block diagram of the experimental digital hearing aid

3 DYNAMIC COMPRESSION ALGORITHM

For the first clinical tests, a very flexible algorithm for binaurally compressing the dynamic range of speech signals in three frequency channels was implemented. The principle of the three channel compression amplification is well known and realized in hearing aids of several manufacturers, which all use the same signal processing chip [5, 7]. It has been proven successful for the compensation of complex sensorineural hearing losses. These losses show the recruitment phenomenon, i.e. the loss is not only dependent on signal frequency but also on signal level. Especially, much gain is needed for small levels, but no gain or even attenuation for very loud sounds. In contrast to the above mentioned sampled-analogue chip [7], our full digital implementation on the DSP56002 is more flexible and provides more parameters for the fitting to the individual hearing loss.
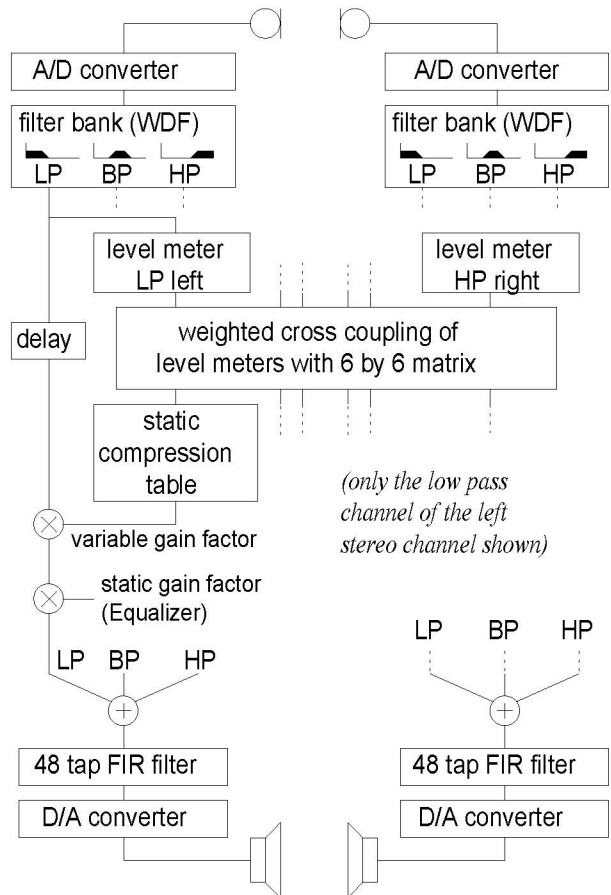

Fig. 2: Block diagram of the three channel dynamic compression algorithm

A filter bank, based on lattice wave digital filters (WDF) [1], subdivides each signal (left or right, resp.) into three frequency channels. The dynamic compression is implemented as a feed-forward control algo-
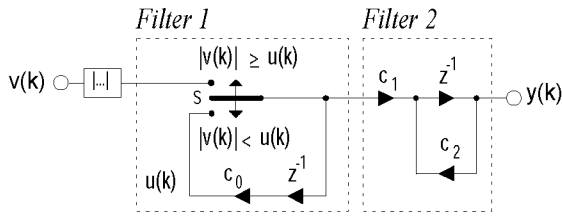
Fig. 3: Level meter filter arrangement

rithm [8] (Fig. 2). The input signal level in each channel is measured by the digital simulation of a RC integration filter (Fig. 3). This slope-sensitive filter enables the realization of different attack and release times. The signal level is computed as a sliding absolute mean value to avoid audibility of non-linear distortions in the output signals (caused by the level dependent gain variations). When the input signal amplitude is increasing, the switch S disables filter 1. Therefore, only the coefficients c_1 and c_2 of filter 2 are responsible for the attack time of the compression algorithm. The attack time can be varied in a wide range. When the signal slope is negative both filters are cascaded. In this case, mainly the coefficient c_0 determines the release time (since $c_2 \ll c_0 \approx 1$). With this filter arrangement various dynamic compression characteristics can be implemented, e.g. syllabic compression with short attack times [9] or the slow acting automatic volume control (AVC). The feed forward structure implemented here has the advantage over analogue compression circuits that a properly chosen delay in the main signal path can reduce or even avoid signal overshoots after sudden level increases.

The six level estimates are cross coupled that each compression table input value can also be influenced by the level estimates of selected other channels. Hence, several perceptional properties of the human auditory system (e.g. upward spread of masking [10] or binaural loudness summation [6]) can be taken into account.

The compression characteristic in each channel is implemented as a look-up table of gain factors, the delayed channel signal is multiplied with. Thus, nearly arbitrary static compression characteristics can be programmed independently for each channel. After the three frequency channel signals are added again (selective in the left and right stereo channel, resp.), a 48 tap FIR filter compensates linear distortions of the analogue hearing aid and the ear canal acoustics.

4   SUPRESSION OF LATERAL SOUNDS AND DEREVERBERATION ALGORITHM

A great advantage of normal hearing persons is their binaural hearing ability, which is very important for speech perception in noisy environments. By evaluating level differences and delays between the two ears, the normal hearing system can localize spatially distributed sound sources, suppress interfering noises and remove room reverberations from the speech signal. If these abilities are impaired, they can be aided by binaural digital signal processing. Suitable DSP algorithms were proposed by Kollmeier et al. [4].
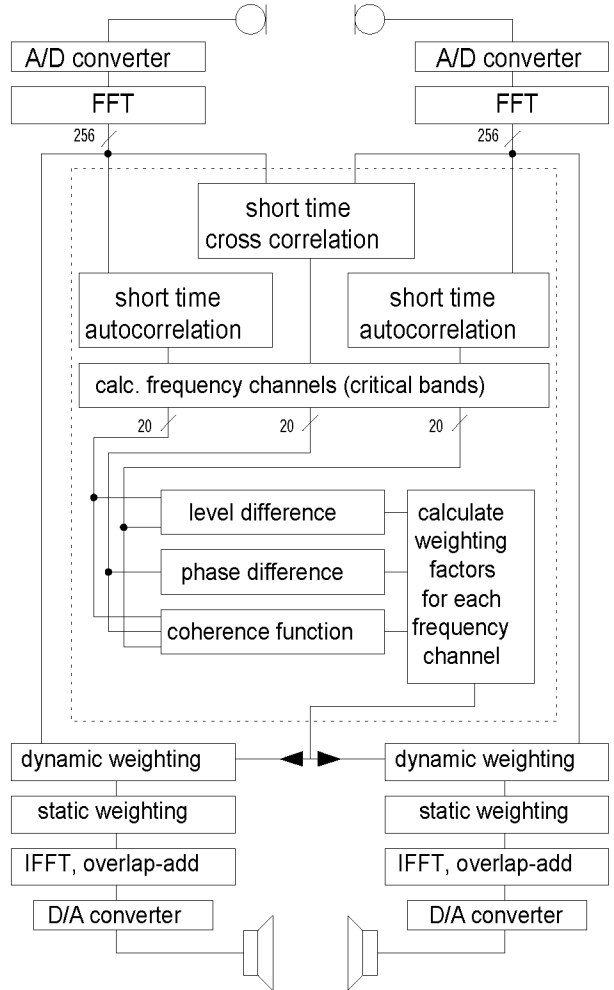

Fig. 4: Binaural speech enhancement algorithm

Level and phase differences between left and right ear signal must be evaluated frequency-specific (as is the case in the normally functioning hearing system). Therefore, the left and right input signals are transformed with a 256-point FFT to the frequency domain (Fig. 4). This can be interpreted as a bank of bandpass filters linearly spaced in frequency. It has proven to be sufficient to calculate the level and phase differences in 20 frequency channels, which resemble a subset (restricted to the hearing aid frequency range) of the critical bands of human hearing [10]. The nearly logarithmic frequency spacing of these channels is approximated by combining adjacent channels of the

short time autocorrelation and cross correlation computed from the FFT output.

From level and phase differences between corresponding left and right frequency channels, the weighting factors

$$w_1(f) = \ln\left\{\frac{S_{xx}(f)}{S_{yy}(f)}\right\} \qquad (2)$$

$$w_2(f) = \arg\left\{S_{xy}(f)\right\} \qquad (3)$$

are computed with f representing the number of the frequency channel. S_{xx} and S_{yy} are the short time auto-correlation of the left and right channel, resp. The short time cross correlation is denoted S_{xy}. A third weighting factor results from the coherence function of the left and the right channel.

$$w_3(f) = \frac{\left|S_{xy}(f)\right|^2}{S_{xx}(f)\cdot S_{yy}(f)} \qquad (4)$$

This factor is a measure of the reverberation contained in a frequency channel. The stereo channels are considered not to be correlated in presence of strong reflected sound components in that respective frequency channel. In this case the weighing factor w_3 will be close to zero. Is the main excitation resulting from direct sound the coherence function yields a value near 1.

The three sets of weighting factors are combined to a single set (one factor for each frequency channel). The FFT channels belonging to each frequency channel are multiplied with the corresponding weighting factor. The result is an attenuation of components of the short time spectrum which contain mostly reverberant or lateral parts of the sound (directional filtering). Inverse FFT yields filtered left and right output time series, resp., which represent an enhanced speech signal. To avoid processing artifacts ("musical noise") which might be caused by concatenation of short-time-FFT processed signal blocks, proper windowing and overlap-add of blocks is used. This algorithm can now for the first time be tested outdoors under real-life conditions.

5   CLINICAL APPLICATION AND FURTHER WORK

The realization of this hearing aid prototype is part of a compound research project dealing with development and testing of algorithms and fitting strategies for digital hearing aids. The Medical Physics department of the University of Oldenburg (Germany) first develops the algorithms theoretically and then performs initial tests on powerful stationary signal processing systems. Evaluation of these

algorithms in extensive field studies is done at the ENT clinics of Amsterdam and Rotterdam, the Netherlands, and Giessen, Germany, by means of several units of the wearable digital hearing aid described here.

To reduce the size and weight of the prototype units and, thus, increasing the wearing comfort, the signal processing unit will soon be implemented using 3.3 V low power components. The necessary components, especially a low voltage 16 bit analogue to digital converter, became available recently. We are working now on a new version of the hearing aid prototype system.

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