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(54) **METHOD AND SYSTEM OF NOISE REDUCTION IN A HEARING AID**

VORRICHTUNG UND VERFAHREN ZUR RAUSCHMINDERUNG IN EINEM HÖRHILFEGERÄT

PROCEDE ET SYSTEME PERMETTANT DE REDUIRE LE BRUIT DANS UN DISPOSITIF D'AIDE AUDITIVE

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- **PREVES D A ET AL: "A FEEDBACK STABILIZING CIRCUIT FOR HEARING AIDS" HEARING INSTRUMENTS, HARCOURT BRACE JOVANOVICH PUBL. DULUTH, MINNESOTA, US, vol. 37, no. 4, April 1986 (1986-04), pages 34,36-41,51, XP000796174 ISSN: 0092-4466**

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Description

BACKGROUND OF THE INVENTION

1. Field of the invention

[0001] The present invention relates to the field of hearing aids and more specifically to hearing aids utilizing noise reduction techniques. The invention further relates to methods for adjusting the hearing aid gain for noise reduction. In addition the invention relates to a system of reducing noise in a hearing aid.

2. Description of the related art

[0002] Hearing aids are adapted for providing at the users eardrum a version of the acoustic environment that has been amplified according to the users prescription. This is normally achieved by providing a device with a microphone, an amplifier and a miniature loudspeaker situated in an earpiece placed in the users ear canal. It is well known that there may be acoustic leaks around the earpiece. There may e.g. be a non-sealed fit or there may be a vent deliberately arranged in the ear piece for considerations about user comfort, e.g. for relieving the sound pressure created by the users own voice. Such leaks may cause a loss in sound pressure and they may allow sound to bypass the hearing aid to reach the ear drum.

[0003] Unpublished PCT application PCT/EP2005/055305 titled "Method and system for fitting a hearing aid", provides a method for estimating otherwise unknown functions such as the vent effect and the direct transmission gain for an in-situ hearing aid. The vent effect estimate is used for correcting the in-situ audiogram and the hearing aid gain. US 2005/013456 refers to a hearing aid wherein the penetration of direct sound through a ventilation channel of the hearing aid device is prevented.

[0004] WO-A-2005/051039 provides a dynamic speech enhancement technique, where speech intelligibility in noise is improved by optimizing a speech intelligibility index, such as SII (see also Methods for Calculation of the Speech Intelligibility Index. ANSI S3.5-1997), AI (see also American National Standard Methods for the Calculation of the Articulation Index. ANSI S3.5-1996). Noise reduction techniques, where speech intelligibility in noise is improved by optimizing a speech intelligibility index, increase of decrease the gain in selected frequency bands, taking into account human auditory masking.

[0005] The sound input to the hearing aid user is a combination of the sound amplified according to the hearing aid gain as well as the direct transmitted sound. As long as the amplified sound dominates the direct transmitted sound in all frequency bands the noise reduction techniques will provide good results. Noise reduction according to the state of the art to enhance SII is based on an assumption that the earplug provides a tight fit between the earplug and the ear canal. However a ventilation canal or a leakage path allows for the sound to be directly transmitted into the ear. Thus, at a certain threshold the sound input to the hearing aid user may be dominated by the direct transmitted sound, so that a decrease of the hearing aid gain will not affect the sound input to the user. If the direct transmitted sound is not taken into account the speech intelligibility may suffer as a consequence.

[0006] Therefore, acoustic effects of the ventilation canal and possible leakage paths between the hearing aid and the ear canal are still challenges in today's hearing aid fitting strategies.

[0007] Thus, there is a need for improved hearing aids as well as improved techniques for utilizing noise reduction in hearing aids.

SUMMARY OF THE INVENTION

[0008] It is therefore an object of the present invention to provide hearing aids and methods of processing signals in a hearing aid taking in particular the mentioned requirements and drawbacks of the prior art into account.

[0009] It is in particular an object of the present invention to provide a hearing aid and a respective method providing a noise reduction technique that take the relative amount of directly transmitted sound through the vent into account.

[0010] It is a further object of the present invention to provide a hearing aid and a respective method providing a SII optimization where speech intelligibility in noise is improved.

[0011] According to a first aspect of the present invention, there is provided a hearing aid that comprises at least one microphone, a signal processing means and an output transducer, wherein the signal processing means is adapted to receive an input signal from the microphone, wherein the signal processing means is adapted to apply a hearing aid gain to the input signal to produce an output signal to be output by the output transducer, and wherein the signal processing means further comprises means for adjusting the hearing aid gain according to a direct transmission gain calculated for the hearing aid.

[0012] This hearing aid with means for adjusting the hearing aid gain according to a direct transmission gain gives a knowledge about the amount of directly transmitted sound and provides information about how much a certain frequency band may be attenuated before the direct sound becomes dominant over the amplified sound.

[0013] According to other aspects of the present invention, the hearing aid and the method are capable of incorporating knowledge of the amount of direct sound into the applied noise reduction algorithm, which thereby is optimized taking the knowledge of vent effect and leakage into account. This provides a more accurate and effective noise reduction than would be otherwise obtainable.

[0014] According to another aspect of the present invention, there is provided a hearing aid that is capable of avoiding phase disruption in the output signal by taking the direct transmitted sound into account when calculating the hearing aid gain to produce the output signal.

[0015] According to another aspect of the present invention, there is provided a method of compensating direct transmitted sound in a hearing aid which comprises the steps of estimating an effective vent parameter for the hearing aid, calculating a direct transmission gain based on the effective vent parameter, and applying a hearing aid gain to produce an output signal from an input signal wherein the direct transmission gain is used as a lower gain limit below which the hearing aid gain is not set.

[0016] According to still another aspect of the present invention, there is provided a method of determining direct transmitted sound in a hearing aid which comprises the steps of estimating an effective vent parameter for the hearing aid, and calculating a direct transmission gain based on the effective vent parameter.

[0017] The provided methods enable a calculation of the direct transmission gain once when fitting the hearing aid which may then be used according to further methods and systems according to the present invention for the dynamic correction of also other hearing aid parameters than gain.

[0018] It may be seen as a true advantage that the hearing aids, systems and methods according to the present invention provide the ability to dynamically adjust the applicable speech intelligibility index gain and the resulting noise reduced hearing aid gain for the direct transmission gain in real time and, thus, the amount of gain that the hearing aid or system may apply at any given instance.

[0019] According to an embodiment of the present invention the hearing aid is able to adjust the hearing aid gain in each frequency band based on the instantaneous gain level, the further SII input parameters and the direct transmission gain in order to improve the overall speech intelligibility. This offers a new approach according to which the direct transmission gain is taken into account in the noise reduction technique, giving the user a better speech intelligibility in noise.

[0020] The invention, according to further aspects, provides a system of reducing noise in a hearing aid, a computer program and a computer program product as recited in claims 26, 27 and 28.

[0021] Further specific variations of the invention are defined by the further claims. Other aspects and advantages of the present invention will become more apparent from the following detailed description taken in conjunction with the accompanying drawings which illustrate, by way of example, the principles of the invention.

BRIEF DESCRIPTION OF THE DRAWINGS

[0022] The invention will be readily understood by the following detailed description in conjunction with the accompanying drawings, wherein like reference numerals designate like structural elements, and in which:

Figs. 1a	depicts a schematic diagram regarding calculation of the direct transmitted sound;
Fig. 1b	depicts a block diagram of a hearing aid according to the present invention.
Fig. 2	depicts the level of signal versus frequency that results by adding contributions of two sound signals;
Fig. 3	depicts the phase disruption range as a function of the difference between the amplitude of the two signals;
Fig. 4	shows a graph of the directly transmitted sound versus frequency;
Fig. 5	shows diagrams illustrating the principle of optimizing the SII (Speech Intelligibility Index) taking into account the directly transmitted sound, according to the present invention; and
Fig. 6	depicts a block diagram of part of a hearing aid according to an embodiment of the present invention.

DESCRIPTION OF EMBODIMENTS OF THE INVENTION

[0023] Reference is first made to Fig. 1 a for an explanation regarding calculating the DTG. The calculation of the DTG is done by performing a feedback test (FBT) as schematically illustrated in Fig. 1 a. Then, the in-situ vent effect is estimated and the DTG is calculated from the vent effect. Document PCT/EP2005/055305 (mentioned above) describes this in detail.

[0024] Reference is now made to Fig. 1 b, which shows a hearing aid 200 according to the first embodiment of the present invention.

[0025] The hearing aid comprises an input transducer or microphone 210 transforming an acoustic input signal into an electrical input signal 215, and an A/D-converter (not shown) for sampling and digitizing the analogue electrical signal. The processed electrical input signal is then fed into signal processing means 220, which includes an amplifier with a compressor for generating an electrical output signal 225 by applying a compressor gain in order to produce an output signal suitable for compensating a hearing loss according to the users requirements. The compressor gain characteristic is, according to an embodiment, non-linear to provide more gain at low input signal levels and less gain at high signal levels. The signal path further comprises an output transducer 230, i.e. a loudspeaker or receiver, for transforming the electrical output signal into an acoustic output signal.

[0026] The compressor operates to compress the dynamic range of the input signals. It is useful for treatment of presbyscusis (loss of dynamic range due to haircell-loss). Actually, compressing hearing aids often apply expansion for low level signals, in order to suppress microphone noise while amplifying input signals just above that level. The compressor may also include a soft-limiter in order to limit maximum output level at safe or comfortable levels. The compressor has a non-linear gain characteristic and, thus, is capable of providing less gain at higher input levels and more gain at lower input levels. Hearing aids embodying a compressor in the signal processor are often referred to as non-linear-gain or compressing hearing aid.

[0027] The signal processing means further comprises memory 240 and adjusting means 250 for adjusting the hearing aid gain further over what the processor basically decides based on the users hearing deficit and the prevailing sound environment. This further adjustment is intended to take into account certain effects of sounds bypassing the hearing aid, e.g. by bypassing the earpiece or by propagating through the vent, as will be explained below.

[0028] For the sake of computations, the sound bypassing the hearing aid is expressed in terms of direct transmission gain (DTG). The direct transmission gain (DTG) is defined as the sound pressure at the ear drum that is generated by an acoustic source outside the ear relative to a sound pressure at the exterior vent opening generated by the same source. As the direct transmission gain is typically less than one, i.e. the log value expressed in dB, will normally be a negative number. However, as there is a natural Helmholz resonance by an earpiece placed in an ear canal there will be frequencies where the DTG is above one, i.e. the log value is a positive number. Information about the direct transmitted sound in the single frequency bands can be estimated by e.g. the methods described in the document PCT/EP2005/055305 to calculate a direct transmission gain for the hearing aid gain used by a certain user.

[0029] The DTG 245 calculated for the hearing aid as a set of frequency dependent gain values is stored in memory 240 of the hearing aid. The DTG is then used by the adjusting means 250 to adjust the hearing aid gain in order to reduce noise, avoid phase disruption or provide any other useful optimization or improvement of the signal quality in the combined acoustic signal on the ear drum resulting from the amplified output signal and the direct transmitted sound.

[0030] Reference is now made to Fig. 2, which depicts the level of signal versus frequency that results by adding contributions of two sound signals, and more specifically shows two frequency dependent signals with a relative phase which are added here, to clarify the principle of adding two sound signals at the eardrum. The black dotted lines are the magnitude of the two signals. The gray dash-dotted line represents the sum of these signals, when the two signals are in phase for all frequencies (upper curve), and when they are out of phase for all frequencies (lower curve), respectively. The full line shows what happens, if the phase difference varies linearly with frequency.

[0031] The sound level at the eardrum of the user is a superposition of the unaided direct sound and the sound amplified by the hearing aid. The interference of the two sound sources may lead to phase disruptions, i.e. fluctuations in the sound input at frequencies where the unaided direct sound and the amplified sound from the hearing aid has about the same magnitude but has opposite phase. This general phenomenon is illustrated in Fig. 2, which illustrates the addition of two signals with differing magnitude and phase.

[0032] At a certain frequency, the sum of two harmonic signals can be written as

$$A_1 \cos(2\pi f t + \varphi_1) + A_2 \cos(2\pi f t + \varphi_2) \qquad (1)$$

[0033] In our example, A1 = 1, $\varphi_1 = 0$ and A2 $\propto$ f. φ_2 is either 0, π or $\propto$ f. With simple calculations, both constructive and destructive interference can be made clear, whereas the sum of two signals with frequency dependent phase and amplitude is more complex to describe analytically. In this case, the resulting phase disruption will depend on the amplitudes and phases of the signals. However, since constructive and destructive interference constitutes the upper and lower limit of the phase disruption, respectively, we know, that a phase disrupted signal lies somewhere in between these lines, as shown in Fig. 2 for the case $\varphi_2 \propto$ f. It is to be noted that the ratio of the absolute amplitude corresponds to the difference of the amplitudes in dB, since dB is calculated as 20log10(A). An amplitude of 0 thus corresponds to -∞ dB.

[0034] The lower dash-dotted gray line shows that in case the two signals are out of phase by π with the exact same amplitude, the total signal cancels out and becomes infinitely small. This is called destructive interference *or phase cancellation.* On the other hand, if the two signals are in phase at all frequencies, the amplitudes simply add up in a constructive interference, and gives 6 dB more sound pressure at the frequency where the two signals have the same amplitude, which can be seen in the upper dash-dotted gray line at 5 kHz. These two cases, however, are rarely met with respect to the hearing aid sound and the direct sound, since both have a varying frequency dependent phase. The black line therefore exemplifies how the total sound pressure might look like, if the relative phase depends linearly on frequency. Note, that at some frequencies, constructive interference increases the magnitude of the total signal, whereas for other frequencies, destructive interference diminishes the total signal. Since the signals do not cancel out as such at frequencies where the relative phase is almost π and the relative amplitude is not quite 1, this phenomenon is called *phase disruption.*

[0035] The above example is general, and can be extrapolated to the situation in a users ear, where the amplified sound and the direct sound superpose. This in turn means that the amplified sound has to exceed a certain level before the total sound pressure at the eardrum remains unperturbed by the direct sound with respect to phase disruption. Maintaining the hearing aid gain at a similar magnitude to the direct sound would result in an increased risk of phase disruption, which is avoided with the current invention.

[0036] As is observed in Fig. 2, the difference in amplitude between the amplified sound and the unaided direct sound must be higher than a certain amount (a safety margin) to minimize phase disruption. Thus there is a lower threshold for the gain setting, equal to the directly transmitted gain + *k,* as suggested by the scale in Fig. 4 to the right. The safety margin is the factor *k*, which in principle could be set to anything. If *k* is negative and numerically large, the interaction between direct and amplified sound is neglected and nothing extraordinary is ever done to take the interaction into account. If *k* is large and positive, measures are taken all the time, which is also not optimal. Choosing the factor k is therefore a trade-off between minimizing the risk of phase disruption and limiting the SII-optimization.

[0037] Fig. 3 shows the phase disruption range versus signal amplitude ratio. Fig. 3 more specifically shows the difference in dB between the amplitude of the in-phase summed signal and the out-of-phase summed signal as a function of the difference between the amplitudes of the two signals shown in Fig. 2. The curve thus shows the uncertainty or possible spread of the total sound pressure due to phase disruption. The signal amplitude ratio in dB is the difference between the hearing aid sound (expressed in terms of gain) and the directly transmitted sound (expressed in terms of gain) in each band, i.e. HA - DTG (Direct Transmitted Gain) in dB, i.e. A_1 is DTG and A_2 is HA. Note, that the DTG is fixed once the earplug is made, whereas the hearing aid gain may change with the sound input. The hearing aid sound is thus the only variable, once the vent has been chosen.

[0038] For example it may be read from the curve that if one signal is 10 dB larger than the other, the phase disruption may in a worst case scenario cause the amplitude of the summed signal to vary up to -5 dB from the in-phase summed signal. Values from 1 and upward are applicable, preferably between 5 and 15 dB. Of course, a value of about 1 dB would incur a high risk of phase disruption. A value of *k* = 7 or *k* = 8 gives a phase disruption range of about +-3 dB, which may be considered acceptable.

[0039] If the hearing aid was turned off, the sound from the hearing aid would be $-\infty$ (completely silent), obviously meaning that the DTG would dominate totally. This would correspond to $-\infty$ on the x-axis in Fig. 3, which gives no phase disruption problems, as we would expect. On the contrary, if the hearing aid gain is e.g. 60 dB and the direct transmitted sound -10 dB, the direct sound is negligible in comparison, and no phase disruption is risked. It is only when the sound level of the direct sound and the hearing aid sound are comparable ($A_2 \approx A_1$), that the strength of the summed signal may vary significantly as indicated in Fig. 3.

[0040] Thus, in the current invention, the factor k, which is indicated as an example in Fig. 3, constitutes a lower limit, below which the hearing aid gain should not be set during the optimization process, without risking a large amount of phase disruption.

[0041] Information about the direct transmitted sound in the single frequency bands can be estimated by e.g. the methods described in the document PCT/EP2005/055305 to calculate a direct transmission gain for the hearing aid gain used by a certain user. This knowledge will then be used to optimize SII. If the direct sound e.g. dominates the lowest band, it is possible to find a new optimum for SII by changing the gain in some of the bands where the amplified sound dominates.

[0042] According to an embodiment, the adjusting means is a means for optimizing a speech intelligibility index (SII) by applying a respective noise reduction technique taking the DTG into account to give the user a better speech intelligibility in noise, as will now be described in detail.

[0043] The Figs. 4 and 5 show the principle in the combination of SII (Speech Intelligibility Index) - based noise reduction technique and the directly transmitted sound through the vent.

[0044] The Fig. 4 shows the directly transmitted sound in dB. This gain function, called the direct transmission gain, represents the sound pressure at the eardrum relative to the sound pressure at the entrance of the vent by a sound source external to the ear. The direct transmission gain may be determined during the feedback test, as in the above-mentioned PCT/EP2005/055305.

[0045] The values in this example are calculated for 15 frequency bands between 100 Hz and 10 kHz. The figure has two y-scales, where the left represents the direct transmission gain, and the right represents a minimal amplification, which the hearing aid gain must exceed in order to dominate the total sound at the eardrum. The minimum amplification is determined as the hearing aid gain necessary to avoid the risk of phase disruption problems caused by adding two sound pressures of same magnitude but opposite phase. Such phase disruption results in bad sound quality, which may be described as metallic or raspy at the frequencies in which phase disruption occurs.

[0046] The letter k in these figures refers to a limit in dB where the amplified sound is large enough to dominate the total sound pressure at the eardrum relative to the direct sound. k is a limit that divides the action of the algorithm into two states: one, where actions need to be taken to avoid phase disruption, and one where no action is needed. If the amplified sound - *k* is less than the direct sound, there is a risk of phase disruption, and something must be done. See Fig. 3 for clarification on the *k*-factor. In the Fig. 4 the direct transmission gain and the minimum amplification is emphasized for frequency band 4 and frequency band 5 for an estimated vent diameter of 1 mm (dark color) respectively 3 mm (light color).

[0047] In the diagrams of Fig. 5, the minimum amplification for k = 8 dB for the two frequency bands are marked on the graphs, containing the hearing aid gain adjustment necessary to find the optimum gain setting with regards to speech intelligibility. These graphs show how the direct transmission gain interacts and interferes with the hearing aid gain in the search for the optimum gain setting with regards to the SII.

[0048] The graphs illustrate how the SII varies as a function of the hearing aid gain for two frequency bands, with a given vent diameter and hearing loss. The SII is illustrated as contour curves. The SII varies between 0 and 1. It is approximately monotonous though it may have some local minima or maxima. By varying the gain in one or more frequency bands an optimum setting of the gain in each frequency band is determined leading to an optimum SII for the hearing aid.

[0049] The diagrams in Fig. 5 illustrate the gain for a frequency band 4, having a center frequency of 500 Hz, and for a frequency band 5, having a center frequency of 634 Hz. The contour curves show how the SII is a function of the setting of the gain in each frequency band.

[0050] The SII optimization according to the prior art does not presently take the direct sound arriving through e.g. the vent into account. However, the direct sound adds to the hearing aid amplified sound and thus in practice it will not be possible to obtain a gain lower than the gain originating from the direct sound. The presence of a large vent in the ear mould in combination with a relatively mild hearing loss may thus imply that only the direct sound is heard, since it might overwhelm the amplified sound.

[0051] A further explanation on how SII is used for noise reduction in a hearing aid is found in WO-A-20051051039.

[0052] The diagrams in Fig. 5 also illustrate and exemplify the actual interval of the gain when k has been chosen to 8 for each of the frequency bands 4 and 5, for two vent diameters (1 mmø and 3 mmø) in combination with two hearing losses (flat 40 dB HL and flat 80 dB HL).

[0053] The optimization of the SII in the hearing aid is performed in all bands, i.e. 15 dimensions in this example. However, illustrating an optimization procedure in 15 dimensions rather impedes than facilitates an easily understandable visualization of the principle. The diagrams in Fig. 5 are therefore limited to illustrate a way of optimizing the SII in two selected bands (bands 4 and 5). An example of a linear optimization method where the gain for frequency band 4 is kept constant and where the gain of frequency band 5 is varied in steps until an optimum SII for that setting has been detected, then the gain of frequency band 4 is varied and the previously detected optimum setting of frequency band 5 is kept constant until an optimum setting of frequency band 4 has been detected.

[0054] The diagrams in Fig. 5 illustrate an optimization procedure where the optimization is continued until it is not possible to obtain a better SII. Naturally other optimization methods can be implemented, as long as the method takes the direct sound into account. The contour plot shows the SI-index as a function of the absolute gain in each band. The theoretical optimum i.e., when it is assumed that the sound at the eardrum is provided only by the hearing aid, is easily detected as an 'island' in the plot. However, the direct sound (plus *k*), which is illustrated on the axes by use of the same symbols as in the top plot, influences not only whether that optimum is attainable or not, but also the path leading to the optimum. The gray area illustrates the region, which would be counterproductive to enter. The iterative optimization process, which could be performed in many ways, is here illustrated as a sequential adjustment of each band. A star indicates the result of the optimization method.

[0055] In the graph (upper right pane) for a severe hearing loss (HTL 80dB) combined with a small vent (1 mm), no changes occur to the optimum parameter setting resulting in the optimum SII when the minimum amplification is taken into consideration compared to the conventional optimum parameter setting where the gain can be varied in the entire area. In contrary, a large vent (3 mm) and a mild hearing loss (HTL=40 dB) may allow enough direct sound to enter through the vent to influence or even dominate the total sound pressure at the eardrum (lower left pane), such that the optimum gain setting of the frequency bands is quite different when the minimum amplification is used to limit the gain settings of the frequency bands, than if the frequency bands are varied without taken this into account. In such cases this would lead to a much better parameter setting of the gain in the various frequency bands.

[0056] Therefore the iterative optimization path may be different from what would otherwise be carried out and the optimum parameter setting may also be different from what would else be determined as optimum according to other embodiments.

[0057] A main advantage for the present invention is therefore that the SII is optimized under consideration of the actual in-situ acoustic surroundings.

[0058] It is evident for the person skilled in the art that the shown iterative path may vary greatly from a real iterative path, both due to the optimization method and to the fact that optimization occurs in all bands.

[0059] Reference is now made to Fig. 6, which shows a part of a hearing aid 300 according to another embodiment of the present invention.

[0060] SII optimization block 610 as means for optimizing a speech intelligibility index produces the SII gain 615, which is fed to the combiner or summation block 620, where the signal 615 is subtracted from the amplified sound signal 605 produced by the signal processor or compressor by applying the hearing aid gain. The output of the combiner may be considered as the noise reduced output signal 625 fed to the output transducer and also fed to the comparator 630. The comparator 630 compares the noise reduced output signal 625 plus the safety margin k in block 640 with the direct transmitted sound according to the DTG in block 245, both also supplied to the comparator. If the level of the noise reduced output signal plus the safety margin k is at or below the DTG, the comparator produces an error signal 635 which is fed to the SII optimizer 610 as a further input parameter which is taken into account during optimization of the SII so that the noise reduced output signal will not be attenuated below the threshold anymore in order to avoid phase disruption.

[0061] In a modified embodiment the hearing aid comprises a band-split filter for converting the input signal into band-split input signals of a plurality of frequency bands and the hearing aid is adapted to process the band-split input signals in each of the frequency bands independently.

[0062] According to embodiments of the present invention, systems and hearing aids described herein may be implemented on signal processing devices suitable for the same, such as, e.g., digital signal processors, analogue/digital signal processing systems including field programmable gate arrays (FPGA), standard processors, or application specific signal processors (ASSP or ASIC). Obviously, it is preferred that the whole system is implemented in a single digital component even though some parts could be implemented in other ways - all known to the skilled person.

[0063] Hearing aids, methods, systems and other devices according to embodiments of the present invention may be implemented in any suitable digital signal processing system. The hearing aids, methods and devices may also be used by, e.g., the audiologist in a fitting session. Methods according to the present invention may also be implemented in a computer program containing executable program code executing methods according to embodiments described herein. If a client-server-environment is used, an embodiment of the present invention comprises a remote server computer, which embodies a system according to the present invention and hosts the computer program executing methods according to the present invention. According to another embodiment, a computer program product like a computer readable storage medium, for example, a floppy disk, a memory stick, a CD-ROM, a DVD, a flash memory, or another suitable storage medium, is provided for storing the computer program according to the present invention.

[0064] According to a further embodiment, the program code may be stored in a memory of a digital hearing device or a computer memory and executed by the hearing aid device itself or a processing unit like a CPU thereof or by any other suitable processor or a computer executing a method according to the described embodiments.

[0065] Having described and illustrated the principles of the present invention in embodiments thereof, it should be apparent to those skilled in the art that the present invention may be modified in arrangement and detail without departing from such principles.

Claims

1. A hearing aid (200) comprising at least one microphone (210), a signal processing means (220) and an output transducer (230), wherein said signal processing means is adapted to receive an input signal from the microphone, wherein said signal processing means is adapted to apply a hearing aid gain to said input signal to produce an output signal to be output by said output transducer,
characterised in that
said signal processing means further comprises means for adjusting said hearing aid gain according to direct transmission gain calculated for the hearing aid;
wherein said means for adjusting said hearing aid gain is adapted to adjust said hearing aid gain to a value not below said direct transmission gain.

2. The hearing aid according to claim 1, wherein said means for adjusting said hearing aid gain comprises means for applying dynamic noise reduction techniques.

3. The hearing aid according to claim 2, wherein said means for adjusting said hearing aid gain are adapted to improve speech intelligibility in noise of said output signal.

4. The hearing aid according to claim 3, wherein said means for adjusting said hearing aid gain further comprises means for optimizing a speech intelligibility index.

5. The hearing aid according to claim 4, wherein said means for adjusting said hearing aid gain is adapted to optimize said speech intelligibility index to produce a set of frequency dependent speech intelligibility index gain values for each time sample of said input signal.

6. The hearing aid according to one of the preceding claims, wherein said means for adjusting said hearing aid gain provides a safety margin k and is adapted to adjust said hearing aid gain to a value not below said direct transmission gain plus said safety margin.

7. The hearing aid according to one of claims 4 to 5, wherein said means for optimizing a speech intelligibility index is adapted to calculate a speech intelligibility index gain as a function of a plurality of input parameters.

8. The hearing aid according to claim 7, wherein said input parameters comprises at least one of a frequency dependent hearing threshold level, an estimated noise level, and an estimated speech level.

9. The hearing aid according to claim 5 or 7 to 8, wherein said means for adjusting said hearing aid gain is adapted to calculate a noise reducing hearing aid gain from an initial hearing aid gain and said optimized speech intelligibility index gain, and to adjust said noise reducing hearing aid gain to a value not below a threshold level.

10. The hearing aid according to claim 9, wherein said threshold level is the level of said direct transmission gain.

11. The hearing aid according to claim 9, wherein said threshold level is the level of said direct transmission gain plus a safety margin.

12. The hearing aid according to one of claims 9 to 11, wherein said means for adjusting said hearing aid gain is adapted to detect the level of said noise reducing hearing aid gain before adjustment and, if said noise reducing hearing aid gain would be below said threshold level, to input said noise reducing hearing aid gain before adjustment as a further input parameter to said means for calculating a speech intelligibility index.

13. The hearing aid according to claim 6, wherein said safety margin is a gain value in the range of 0 to 15 dB, preferably in the range of 5 to 15 dB.

14. The hearing aid according to claim 6, wherein said safety margin is a gain value of 5 to 8 dB, preferably 7 to 8 dB.

15. The hearing aid according to one of the preceding claims, further comprising a band-split filter for converting said input signal into band-split input signals of a plurality of frequency bands and wherein said hearing aid is further adapted to process said band-split input signals in each of said frequency bands independently.

16. A method of reducing noise in a hearing aid (200) comprising at least one microphone (210) producing an input signal, a signal processing means (220) producing an output signal from said input signal, and an output transducer (230) outputting said output signal, wherein said method comprises:

- storing a direct transmission gain calculated for said hearing aid and its user in a memory of said hearing aid; and
- applying a hearing aid gain to said input signal to produce said output signal,

characterised in that
said hearing aid gain is adjusted by said direct transmission gain so that said hearing aid gain is not set to a value below said direct transmission gain.

17. The method according to claim 16, wherein said step of adjusting said hearing aid gain comprises the step of applying dynamic noise reduction techniques.

18. The method according to claim 17, wherein said step of adjusting said hearing aid gain comprises improving speech intelligibility in noise of said output signal.

19. The method according to claim 18, wherein said step of adjusting said hearing aid gain further comprises a step of optimizing a speech intelligibility index.

20. The method according to one of claims 16 to 19, wherein said step of adjusting said hearing aid gain comprises calculating a speech intelligibility index gain reducing the noise in said output signal and adjusting said hearing aid gain by said speech intelligibility index gain.

21. The method according to claim 19, wherein said step of adjusting said hearing aid gain comprises optimizing said speech intelligibility index to produce a set of frequency dependent speech intelligibility index gain values for each time sample of said input signal.

22. The method according to claim 20, wherein said speech intelligibility index gain is calculated with said direct transmission gain as a constraint to ensure that said hearing aid gain is not set to a value below said direct transmission gain.

23. The method according to one of claims 16 to 22, wherein said hearing aid gain is not set to a value below said direct transmission gain plus a safety margin k.

24. The method according to one of claims 20 to 23, further comprising the step of converting said input signal into band-split input signals of a plurality of frequency bands and wherein said method is further carried out for each of said frequency bands.

25. The method according to one of claim 20 or 22, wherein said speech intelligibility index gain comprises a set of frequency dependent gain values which are calculated simultaneously for an actual time sample of said frequency dependent input signal.

26. A system of reducing noise in a hearing aid comprising means adapted to carry out a method according to one of claims 16 to 25.

27. A computer program comprising executable program code which, when executed on a computer, executes a method according to one of claims 16 to 25.

28. A computer program product containing a computer readable medium with executable program code which, when executed on a computer, executes a method according to one of claims 16 to 25.

Patentansprüche

1. Hörgerät (200) umfassend zumindest ein Mikrofon (210), eine Signalverarbeitungseinrichtung (220) und einen Ausgangsmesswandler (230), wobei die Signalverarbeitungseinrichtung geeignet ist zum Empfangen eines Eingangssignals von dem Mikrofon, wobei die Signalverarbeitungseinrichtung geeignet ist, zum Anwenden einer Hörgeräteverstärkung auf das Eingangssignal, um ein Ausgangssignal zu erzeugen, das von dem Ausgangsmesswandler ausgegeben werden soll,

dadurch gekennzeichnet, dass
die Signalverarbeitungseinrichtung weiterhin Mittel zum Justieren der Hörgeräteverstärkung entsprechend einer für das Hörgerät berechneten direkten Übertragungsverstärkung umfasst;
wobei das Mittel zum Justieren der Hörgeräteverstärkung geeignet ist zum Justieren der Hörgeräteverstärkung auf einen Wert, der nicht unterhalb der direkten Übertragungsverstärkung liegt.

2. Hörgerät nach Anspruch 1, wobei das Mittel zum Justieren der Hörgeräteverstärkung Mittel zum Anwenden von dynamischen Rauschverminderungstechniken umfasst.

3. Hörgerät nach Anspruch 2, wobei die Mittel zum Justieren der Hörgeräteverstärkung geeignet sind zum Verbessern der Sprachverständlichkeit im Rauschen des Ausgangssignals.

4. Hörgerät nach Anspruch 3, wobei das Mittel zum Justieren der Hörgeräteverstärkung weiterhin Mittel zum Optimieren eines Sprachverständlichkeitsindexes umfasst.

5. Hörgerät nach Anspruch 4, wobei das Mittel zum Justieren der Hörgeräteverstärkung geeignet ist zum Optimieren des Sprachverständlichkeitsindexes, um einen Satz von frequenzabhängigen Verstärkungswerten des Sprachverständlichkeitsindexes für jeden Zeitabtastwert des Eingangssignals zu erzeugen.

6. Hörgerät nach einem der vorhergehenden Ansprüche, wobei das Mittel zum Justieren der Hörgeräteverstärkung einen Sicherheitsspielraum k vorsieht und geeignet ist zum Justieren der Hörgeräteverstärkung auf einen Wert, der nicht unterhalb der direkten Übertragungsverstärkung plus dem Sicherheitsspielraum liegt.

7. Hörgerät nach einem der Ansprüche 4 bis 5, wobei das Mittel zum Optimieren des Sprachverständlichkeitsindexes geeignet ist zum Berechnen einer Sprachverständlichkeitsindexverstärkung als eine Funktion einer Mehrzahl von Eingangsparametern.

8. Hörgerät nach Anspruch 7, wobei die Eingangsparameter zumindest einen von einem frequenzabhängigen Hörschwellenpegel, einem geschätzten Rauschpegel und einem geschätzten Sprachpegel umfassen.

9. Hörgerät nach einem der Ansprüche 5 oder 7 bis 8, wobei das Mittel zum Justieren der Hörgeräteverstärkung geeignet ist zum Berechnen einer rauschvermindernden Hörgeräteverstärkung aus einer anfänglichen Hörgeräteverstärkung und der optimierten Sprachverständlichkeitsindexverstärkung und zum Justieren der rauschvermindernden Hörgeräteverstärkung auf einen Wert, der nicht unterhalb eines Schwellenpegels liegt.

10. Hörgerät nach Anspruch 9, wobei der Schwellenpegel der Pegel der direkten Übertragungsverstärkung ist.

11. Hörgerät nach Anspruch 9, wobei der Schwellenpegel der Pegel der direkten Übertragungsverstärkung plus einem Sicherheitsspielraum ist.

12. Hörgerät nach einem der Ansprüche 9 bis 11, wobei das Mittel zum Justieren der Hörgeräteverstärkung geeignet ist zum Erkennen des Pegels der rauschvermindernden Hörgeräteverstärkung vor dem Justieren und, wenn die rauschvermindernden Hörgeräteverstärkung unter dem Schwellenpegel sein sollte, zum Eingeben der rauschvermindernden Hörgeräteverstärkung vor dem Justieren als einen weiteren Parameter für das Mittel zum Berechnen eines Sprachverständlichkeitsindexes.

13. Hörgerät nach Anspruch 6, wobei der Sicherheitsspielraum ein Verstärkungswert in dem Bereich von 0 bis 15 dB, vorzugsweise in dem Bereich 5 bis 15 dB ist.

14. Hörgerät nach Anspruch 6, wobei der Sicherheitsspielraum ein Verstärkungswert in dem Bereich von 5 bis 8 dB, vorzugsweise in dem Bereich 7 bis 8 dB ist.

15. Hörgerät nach einem der vorhergehenden Ansprüche, weiterhin umfassend einen Bandteilungsfilter zum Umwandeln der Eingangssignale in bandgeteilte Eingangssignale einer Mehrzahl von Frequenzbändern und wobei das Hörgerät weiterhin geeignet ist zum unabhängigen Verarbeiten der bandgeteilten Eingangssignale in jedem der Frequenzbänder.

16. Verfahren zur Rauschverminderung in einem Hörgerät (200) umfassend zumindest ein Mikrofon (210), das ein Eingangssignal erzeugt, eine Signalverarbeitungseinrichtung (220), die aus dem Eingangssignal ein Ausgangssignal erzeugt, und einen Ausgangsmesswandler (230), der das Ausgangssignal ausgibt, wobei das Verfahren umfasst:

- Speichern einer direkten Übertragungsverstärkung, berechnet für das Hörgerät und seinen Benutzer, in einem Speicher des Hörgerätes; und
- Anwenden einer Hörgeräteverstärkung auf das Eingangssignal, um das Ausgangssignal zu erzeugen,

dadurch gekennzeichnet, dass
das Hörgerät mittels der direkten Übertragungsverstärkung justiert wird, sodass die Hörgeräteverstärkung nicht auf einen Wert gesetzt wird, der unter der direkten Übertragungsverstärkung liegt.

17. Verfahren nach Anspruch 16, wobei der Schritt des Justierens der Hörgeräteverstärkung den Schritt des Anwendens von dynamischen Rauschverminderungstechniken umfasst.

18. Verfahren nach Anspruch 17, wobei der Schritt des Justierens der Hörgeräteverstärkung umfasst Verbessern der Sprachverständlichkeit im Rauschen des Ausgangssignals.

19. Verfahren nach Anspruch 18, wobei der Schritt des Justierens der Hörgeräteverstärkung weiterhin einen Schritt des Optimierens eines Sprachverständlichkeitsindexes umfasst.

20. Verfahren nach einem der Ansprüche 16 bis 19, wobei der Schritt des Justierens der Hörgeräteverstärkung umfasst Berechnen einer Sprachverständlichkeitsindexverstärkung, die das Rauschen in dem Ausgangssignal vermindert, und Justieren der Hörgeräteverstärkung mittels der Sprachverständlichkeitsindexverstärkung.

21. Verfahren nach Anspruch 19, wobei der Schritt des Justierens der Hörgeräteverstärkung umfasst Optimieren des Sprachverständlichkeitsindexes zum Erzeugen eines Satzes von frequenzabhängigen Verstärkungswerten des Sprachverständlichkeitsindexes für jeden Zeitabtastwert des Eingangssignals.

22. Verfahren nach Anspruch 20, wobei die Sprachverständlichkeitsindexverstärkung berechnet wird, mit der direkten Übertragungsverstärkung als eine Bedingung, um sicher zu machen, dass die Hörgeräteverstärkung nicht auf einen Wert gesetzt wird, der unter der direkten Übertragungsverstärkung liegt.

23. Verfahren nach einem der Ansprüche16 bis 22, wobei die Hörgeräteverstärkung nicht auf einen Wert gesetzt wird, der unter der direkten Übertragungsverstärkung plus einem einen Sicherheitsspielraum k liegt.

24. Verfahren nach einem der Ansprüche 20 bis 23, weiterhin umfassend den Schritt des Umwandelns der Eingangssignale in bandgeteilte Eingangssignale einer Mehrzahl von Frequenzbändern und wobei das Verfahren weiterhin für jedes der Frequenzbänder ausgeführt wird.

25. Verfahren nach einem der Ansprüche 20 oder 22, wobei die Sprachverständlichkeitsindexverstärkung einen Satz von frequenzabhängigen Verstärkungswerten umfasst, die gleichzeitig für einen tatsächlichen Zeitabtastwert der frequenzabhängigen Eingangssignale berechnet werden.

26. System zur Rauschverminderung in einem Hörgerät umfassend Mittel geeignet zum Ausführen eines Verfahrens nach einem der Ansprüche 16 bis 25.

27. Computerprogramm umfassend ausführbaren Programmcode, welcher, wenn auf einem Computer ausgeführt, ein Verfahren nach einem der Ansprüche 16 bis 25 ausführt.

28. Computerprogramm-Produkt umfassend ein computerlesbares Medium mit ausführbarem Programmcode, welcher, wenn auf einem Computer ausgeführt, ein Verfahren nach einem der Ansprüche 16 bis 25 ausführt.

Revendications

1. Prothèse auditive (200) comprenant au moins un microphone (210), un moyen de traitement de signal (220) et un transducteur de sortie (230), dans laquelle ledit moyen de traitement de signal est à même de recevoir un signal d'entrée du microphone, dans laquelle ledit moyen de traitement de signal est à même d'appliquer un gain de prothèse auditive audit signal d'entrée pour produire un signal de sortie à délivrer par ledit transducteur de sortie, **caractérisée en ce que** :

   ledit moyen de traitement de signal comprend en outre un moyen de réglage dudit gain de prothèse auditive selon un gain de transmission directe calculé pour la prothèse auditive ;
   dans laquelle ledit moyen de réglage dudit gain de prothèse auditive est à même d'ajuster ledit gain de prothèse auditive à une valeur qui n'est pas inférieure audit gain de transmission directe.

2. Prothèse auditive selon la revendication 1, dans laquelle ledit moyen de réglage dudit gain de prothèse auditive comprend un moyen pour appliquer des techniques de réduction du bruit dynamiques.

3. Prothèse auditive selon la revendication 2, dans laquelle ledit moyen de réglage dudit gain de prothèse auditive est à même d'améliorer l'intelligibilité vocale dans le bruit dudit signal de sortie.

4. Prothèse auditive selon la revendication 3, dans laquelle ledit moyen de réglage dudit gain de prothèse auditive comprend en outre un moyen d'optimisation d'un indice d'intelligibilité vocale.

5. Prothèse auditive selon la revendication 4, dans laquelle ledit moyen de réglage dudit gain de prothèse auditive est à même d'optimiser ledit indice d'intelligibilité vocale pour produire un jeu de valeurs de gain d'indice d'intelligibilité vocale fonction de la fréquence pour chaque échantillon de temps dudit signal d'entrée.

6. Prothèse auditive selon l'une quelconque des revendications précédentes, dans laquelle ledit moyen de réglage dudit gain de prothèse auditive fournit une marge de sécurité k et est à même de régler ledit gain de prothèse auditive à une valeur qui n'est pas inférieure audit gain de transmission directe plus ladite marge de sécurité.

7. Prothèse auditive selon l'une quelconque des revendications 4 à 5, dans laquelle ledit moyen d'optimisation d'un indice d'intelligibilité vocale est à même de calculer un gain d'indice d'intelligibilité en fonction d'une pluralité de paramètres d'entrée.

8. Prothèse auditive selon la revendication 7, dans laquelle lesdits paramètres d'entrée comprennent au moins l'un ou l'autre d'un niveau de seuil auditif fonction de la fréquence, d'un niveau de bruit estimé et d'un niveau vocal estimé.

9. Prothèse auditive selon les revendications 5 ou 7 à 8, dans laquelle ledit moyen de réglage dudit gain de prothèse auditive est à même de calculer un gain réducteur de bruit de prothèse auditive à partir d'un gain initial de prothèse auditive et dudit gain d'index d'intelligibilité vocale optimisé et de régler ledit gain de prothèse auditive réducteur de bruit à une valeur non inférieure à un niveau de seuil.

10. Prothèse auditive selon la revendication 9, dans laquelle ledit niveau de seuil est le niveau dudit gain de transmission directe.

11. Prothèse auditive selon la revendication 9, dans laquelle ledit niveau de seuil est le niveau dudit gain de transmission directe plus une marge de sécurité.

12. Prothèse auditive selon l'une quelconque des revendications 9 à 11, dans laquelle ledit moyen de réglage dudit gain de prothèse auditive est adapté pour détecter le niveau dudit gain de prothèse auditive réducteur de bruit avant réglage et, si ledit gain de prothèse auditive réducteur de bruit se trouvait en dessous dudit niveau de seuil, saisir ledit gain de prothèse auditive réducteur de bruit avant réglage comme autre paramètre d'entrée dans ledit moyen de calcul d'un indice d'intelligibilité vocale.

13. Prothèse auditive selon la revendication 6, dans laquelle ladite marge de sécurité est une valeur de gain dans la plage de 0 à 15 dB, de préférence dans la plage de 5 à 15 dB.

14. Prothèse auditive selon la revendication 6, dans laquelle ladite marge de sécurité est une valeur de gain de 5 à 8 dB, de préférence de 7 à 8 dB.

15. Prothèse auditive selon l'une quelconque des revendications précédentes, comprenant en outre un filtre à bande scindée pour convertir ledit signal d'entrée en signaux d'entrée à bande scindée d'une pluralité de bandes de fréquences et dans laquelle ladite prothèse auditive est en outre à même de traiter lesdits signaux d'entrée à bande scindée dans chacune desdites bandes de fréquences indépendamment.

16. Procédé de réduction de bruit dans une prothèse auditive (200) comprenant au moins un microphone (210) produisant un signal d'entrée, un moyen de traitement de signal (220) produisant un signal de sortie à partir dudit signal d'entrée et un transducteur de sortie (230) délivrant ledit signal de sortie, dans lequel ledit procédé comprend les étapes consistant à :

- stocker un gain de transmission directe calculé pour ladite prothèse auditive et son utilisateur dans une mémoire de ladite prothèse auditive ; et
- appliquer un gain de prothèse auditive audit signal d'entrée pour produire ledit signal de sortie,

caractérisé en ce que :

ledit gain de prothèse auditive est réglé par ledit gain de transmission directe de sorte que ledit gain de prothèse auditive ne soit pas réglé à une valeur inférieure audit gain de transmission directe.

17. Procédé selon la revendication 16, dans lequel ladite étape de réglage dudit gain de prothèse auditive comprend

l'étape d'application de techniques de réduction de bruit dynamiques.

18. Procédé selon la revendication 17, dans lequel ladite étape de réglage dudit gain de prothèse auditive comprend l'amélioration de l'intelligibilité vocale dans le bruit dudit signal de sortie.

19. Procédé selon la revendication 18, dans lequel ladite étape de réglage dudit gain de prothèse auditive comprend en outre une étape d'optimisation d'un indice d'intelligibilité vocale.

20. Procédé selon l'une quelconque des revendications 16 à 19, dans lequel ladite étape de réglage dudit gain de prothèse auditive comprend le calcul d'un gain d'indice d'intelligibilité vocale, réduisant le bruit dans ledit signal de sortie et réglant ledit gain de prothèse auditive par ledit gain d'indice d'intelligibilité vocale.

21. Procédé selon la revendication 19, dans lequel ladite étape de réglage dudit gain de prothèse auditive comprend l'optimisation dudit indice d'intelligibilité vocale pour produire un jeu de valeurs de gain d'indice d'intelligibilité vocale fonction de la fréquence pour chaque échantillon de temps dudit signal d'entrée.

22. Procédé selon la revendication 20, dans lequel ledit gain d'indice d'intelligibilité vocale est calculé avec ledit gain de transmission directe comme une contrainte pour s'assurer que ledit gain de prothèse auditive ne soit pas réglé à une valeur inférieure audit gain de transmission directe.

23. Procédé selon l'une quelconque des revendications 16 à 22, dans lequel ledit gain de prothèse auditive n'est pas réglé à une valeur située en dessous dudit gain de transmission directe plus une marge de sécurité k.

24. Procédé selon l'une quelconque des revendications 20 à 23, comprenant en outre l'étape de conversion dudit signal d'entrée en signaux d'entrée à bande scindée d'une pluralité de bandes de fréquences et dans lequel ledit procédé est en outre réalisé pour chacune desdites bandes de fréquences.

25. Procédé selon l'une quelconque des revendications 20 ou 22, dans lequel ledit gain d'indice d'intelligibilité vocale comprend un jeu de valeurs de gain fonction de la fréquence, qui sont calculées simultanément pour un échantillon de temps courant dudit signal d'entrée fonction de la fréquence.

26. Système de réduction de bruit dans une prothèse auditive comprenant des moyens qui sont à même de réaliser un procédé selon l'une quelconque des revendications 16 à 25.

27. Programme informatique comprenant un code de programmation exécutable qui, lorsqu'il est exécuté sur un ordinateur, exécute un procédé selon l'une quelconque des revendications 16 à 25.

28. Produit de programmation informatique contenant un support lisible sur ordinateur avec un code de programmation exécutable qui, lorsqu'il est exécuté sur un ordinateur, exécute un procédé selon l'une quelconque des revendications 16 à 25.

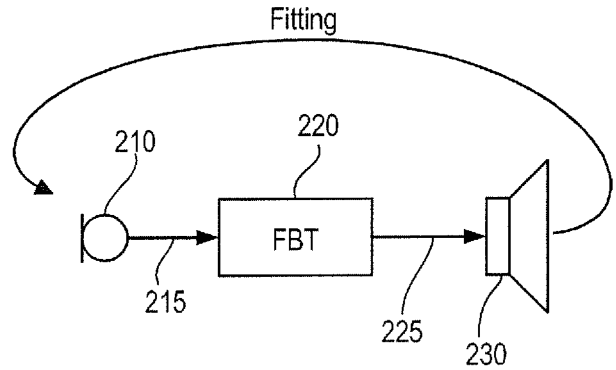

Fitting
210
220
FBT
215
225
230

Fig. 1a

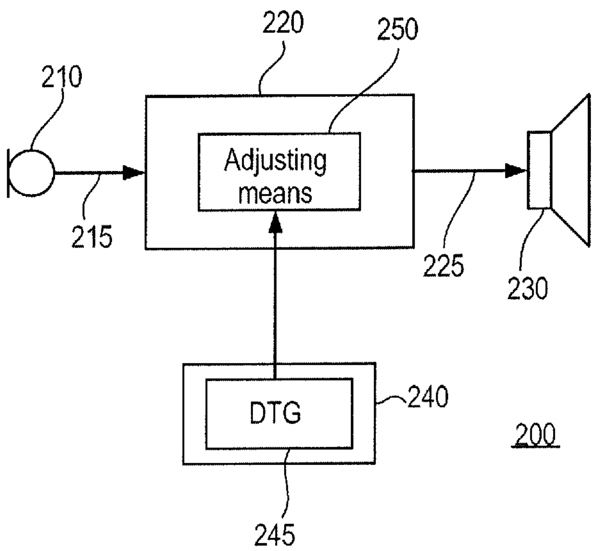
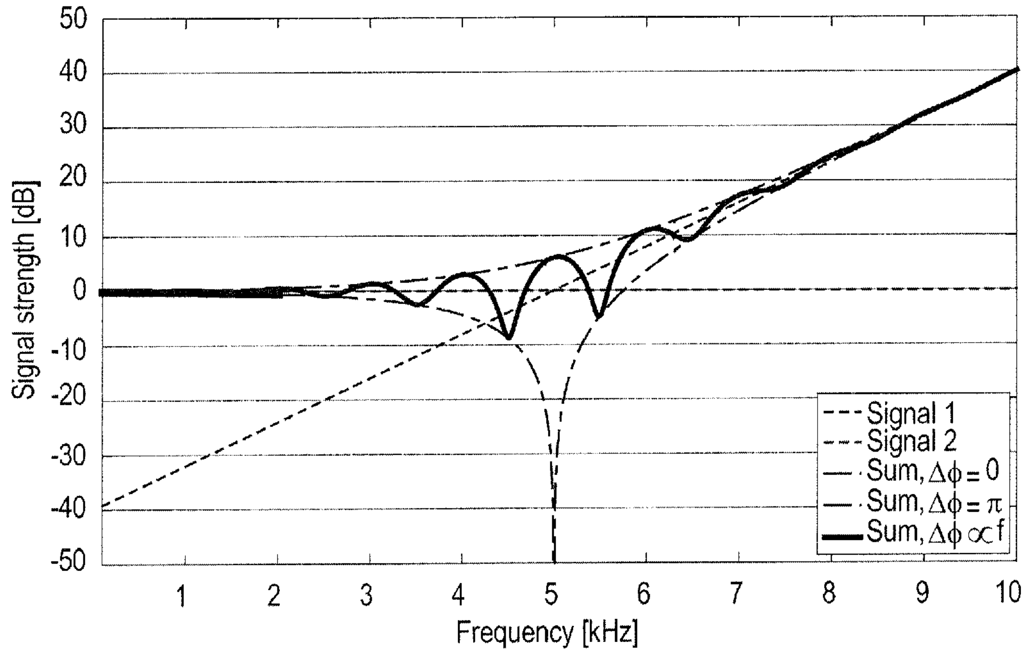
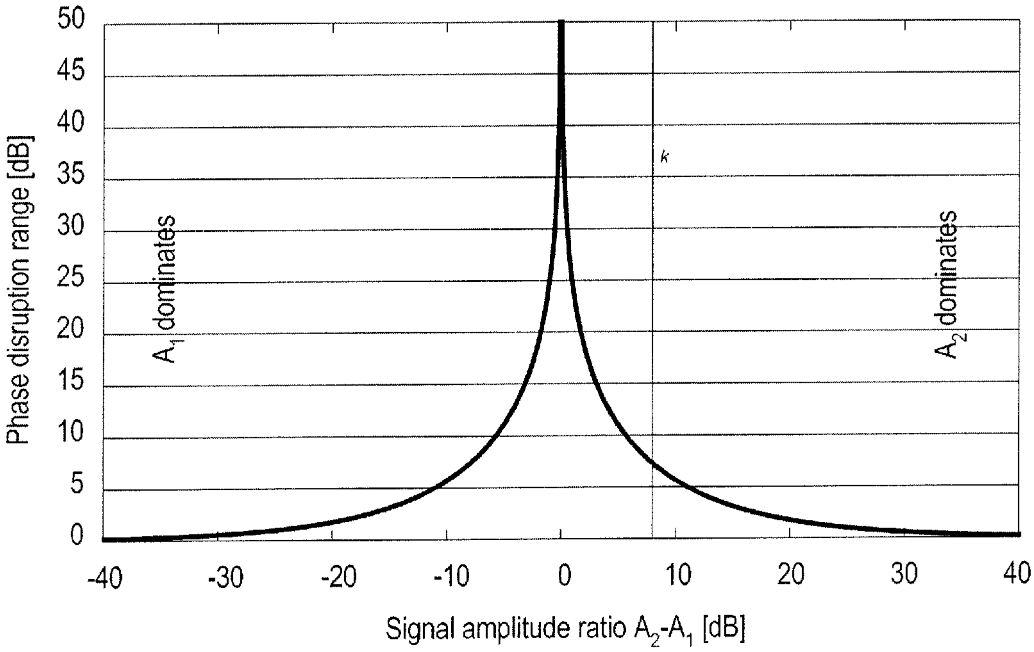

220
250
210
Adjusting
means
215
225
230
DTG
240
200
245

Fig. 1b



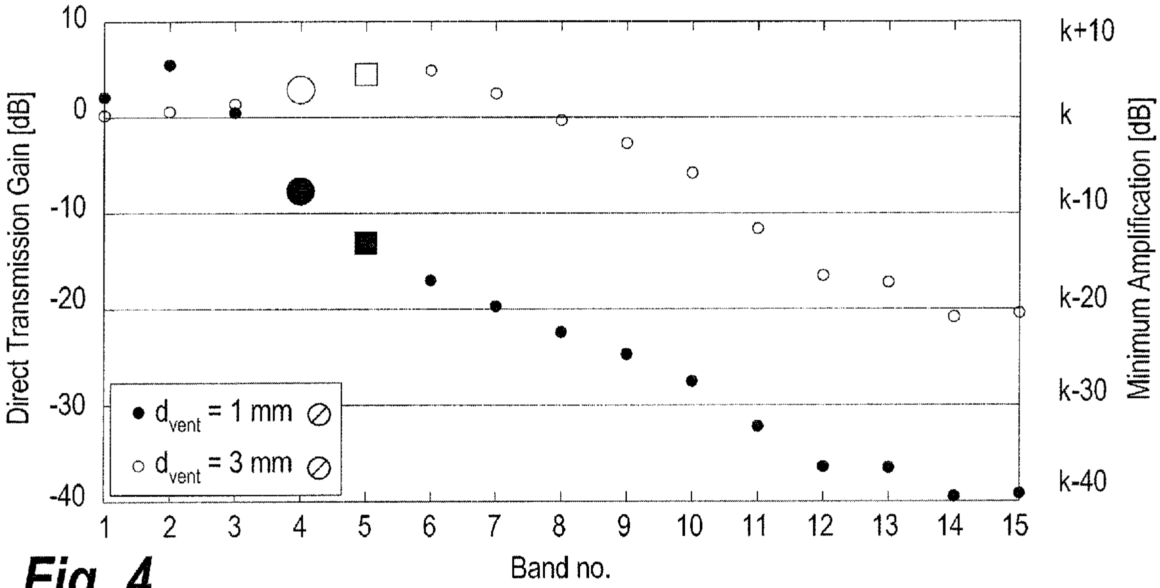
50
40
30
20
10
0
-10
-20
-30
-40
-50
Signal strength [dB]
1
2
3
4
5
6
7
8
9
10
Frequency [kHz]
Signal 1
Signal 2
Sum, $\Delta\phi = 0$
Sum, $\Delta\phi = \pi$
Sum, $\Delta\phi \propto f$

Fig. 2



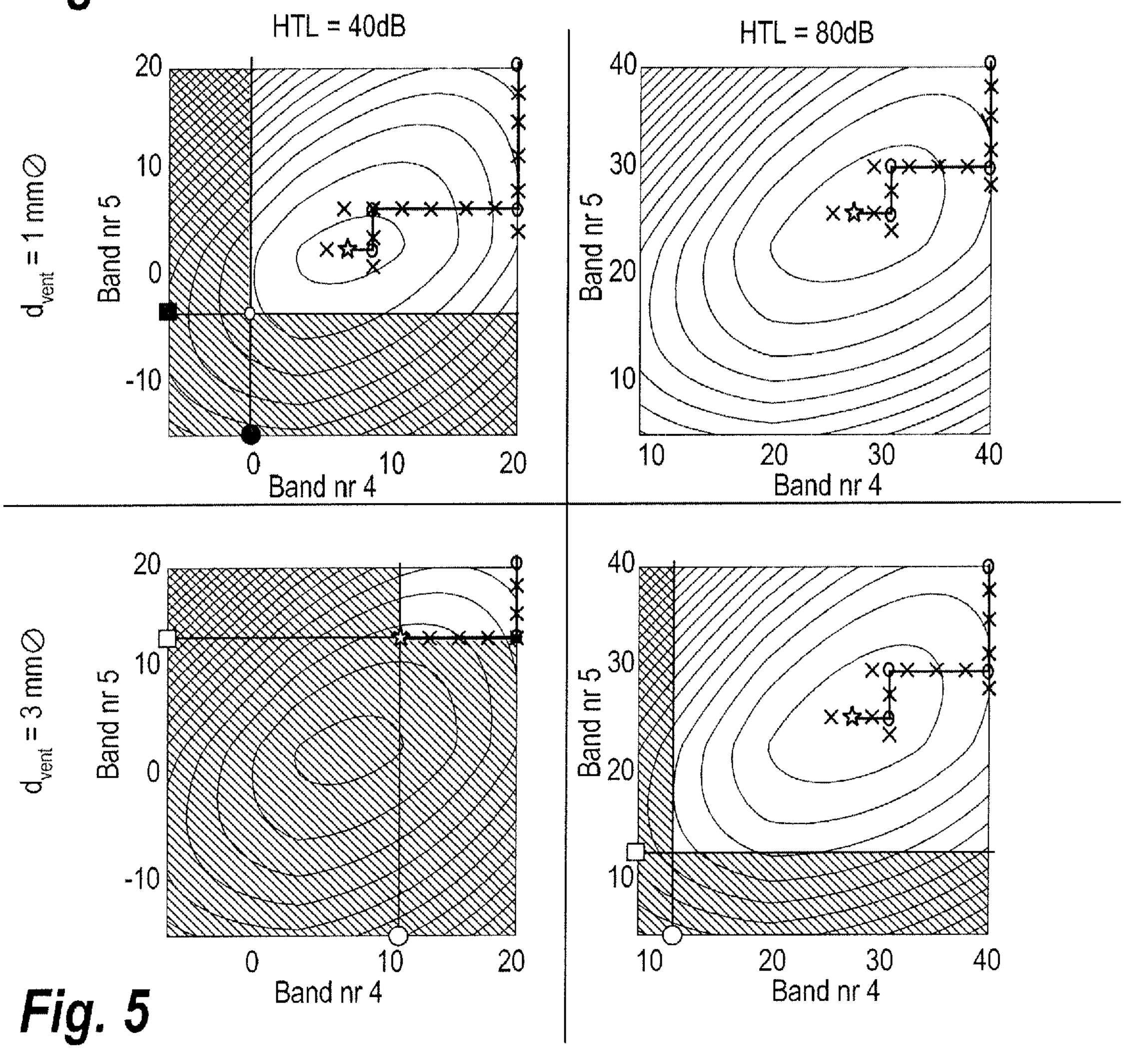
50
45
40
35
30
25
20
15
10
5
0
Phase disruption range [dB]
A_1 dominates
A_2 dominates
k
-40
-30
-20
-10
0
10
20
30
40
Signal amplitude ratio A_2-A_1 [dB]

Fig. 3



Direct Transmission Gain [dB]
Minimum Amplification [dB]
10
0
-10
-20
-30
-40
k+10
k
k-10
k-20
k-30
k-40
d_{vent} = 1 mm ⌀
d_{vent} = 3 mm ⌀
1 2 3 4 5 6 7 8 9 10 11 12 13 14 15
Band no.

Fig. 4



HTL = 40dB
HTL = 80dB
d_{vent} = 1 mm⌀
d_{vent} = 3 mm⌀
Band nr 5
Band nr 4

Fig. 5

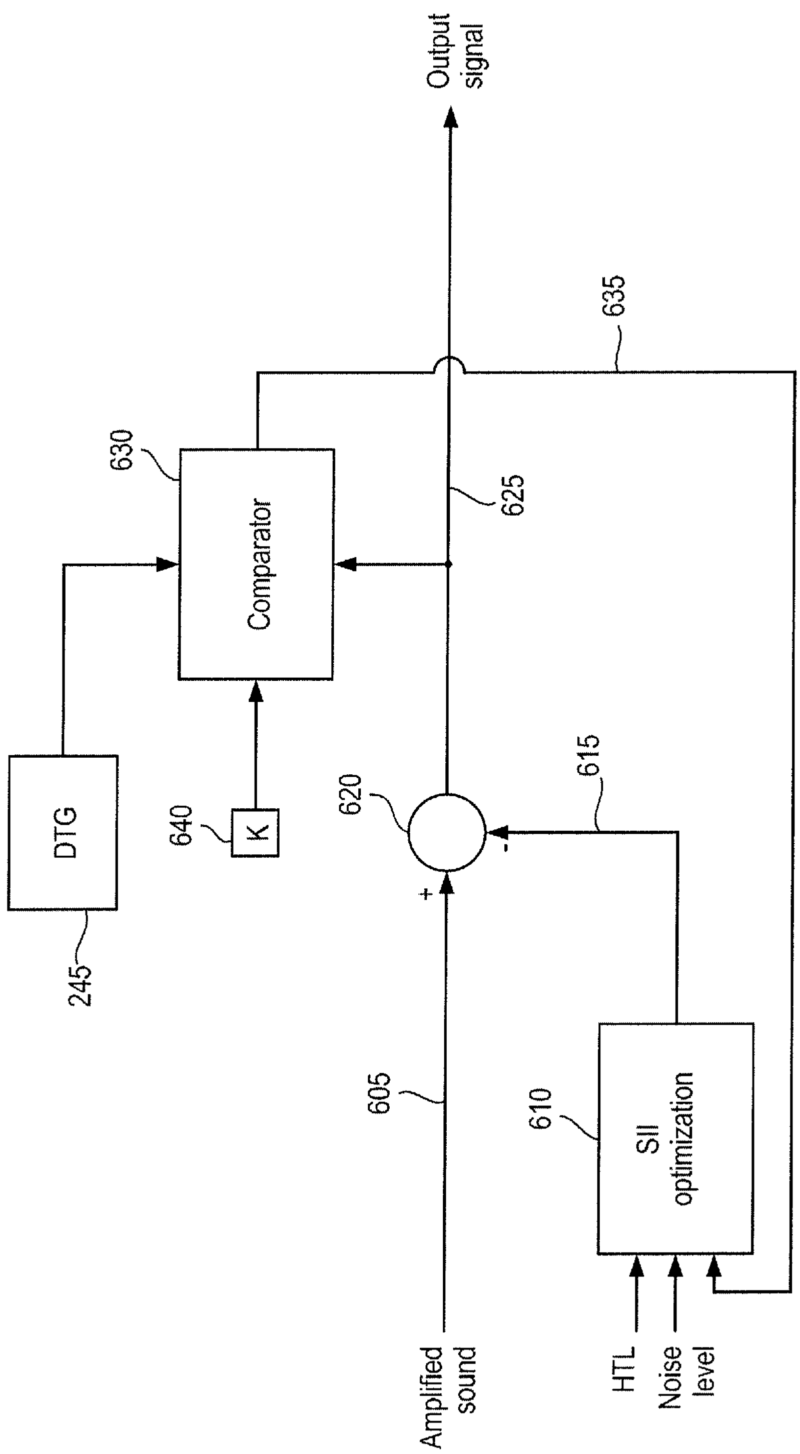

Output signal
635
630
Comparator
625
DTG
640
K
620
615
+
-
245
605
610
SII optimization
Amplified sound
HTL
Noise level

Fig. 6

REFERENCES CITED IN THE DESCRIPTION

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