(12) INTERNATIONAL APPLICATION PUBLISHED UNDER THE PATENT COOPERATION TREATY (PCT)

(19) World Intellectual Property Organization
International Bureau

(43) International Publication Date
20 February 2014 (20.02.2014)

WIPO | PCT

(10) International Publication Number
WO 2014/028658 A2

(51) International Patent Classification:
H04R 25/00 (2006.01)

(21) International Application Number:
PCT/US2013/055004

(22) International Filing Date:
14 August 2013 (14.08.2013)

(25) Filing Language: English

(26) Publication Language: English

(30) Priority Data:
61/683,668 15 August 2012 (15.08.2012) US

(71) Applicant: **MEYER SOUND LABORATORIES, INCORPORATED** [US/US]; 2832 San Pablo Avenue, Berkeley, CA 94702 (US).

(72) Inventors: **MEYER, John, D.**; 928 Arlington Ave., Berkeley, CA 94707 (US). **SZUTS, Toban, A.**; 7249 Blake Street, El Cerrito, CA 94530 (US).

(74) Agent: **BEESON, Donald, L.**; Beeson Skinner Beverly, LLP, One Kaiser Plaza, Suite 750, Oakland, CA 94612 (US).

(81) Designated States *(unless otherwise indicated, for every kind of national protection available)*: AE, AG, AL, AM, AO, AT, AU, AZ, BA, BB, BG, BH, BN, BR, BW, BY, BZ, CA, CH, CL, CN, CO, CR, CU, CZ, DE, DK, DM, DO, DZ, EC, EE, EG, ES, FI, GB, GD, GE, GH, GM, GT, HN, HR, HU, ID, IL, IN, IS, JP, KE, KG, KN, KP, KR, KZ, LA, LC, LK, LR, LS, LT, LU, LY, MA, MD, ME, MG, MK, MN, MW, MX, MY, MZ, NA, NG, NI, NO, NZ, OM, PA, PE, PG, PH, PL, PT, QA, RO, RS, RU, RW, SA, SC, SD, SE, SG, SK, SL, SM, ST, SV, SY, TH, TJ, TM, TN, TR, TT, TZ, UA, UG, US, UZ, VC, VN, ZA, ZM, ZW.

(84) Designated States *(unless otherwise indicated, for every kind of regional protection available)*: ARIPO (BW, GH, GM, KE, LR, LS, MW, MZ, NA, RW, SD, SL, SZ, TZ, UG, ZM, ZW), Eurasian (AM, AZ, BY, KG, KZ, RU, TJ, TM), European (AL, AT, BE, BG, CH, CY, CZ, DE, DK, EE, ES, FI, FR, GB, GR, HR, HU, IE, IS, IT, LT, LU, LV, MC, MK, MT, NL, NO, PL, PT, RO, RS, SE, SI, SK, SM, TR), OAPI (BF, BJ, CF, CG, CI, CM, GA, GN, GQ, GW, KM, ML, MR, NE, SN, TD, TG).

Published:

— *without international search report and to be republished upon receipt of that report (Rule 48.2(g))*

(54) Title: HEARING AID HAVING LEVEL AND FREQUENCY-DEPENDENT GAIN

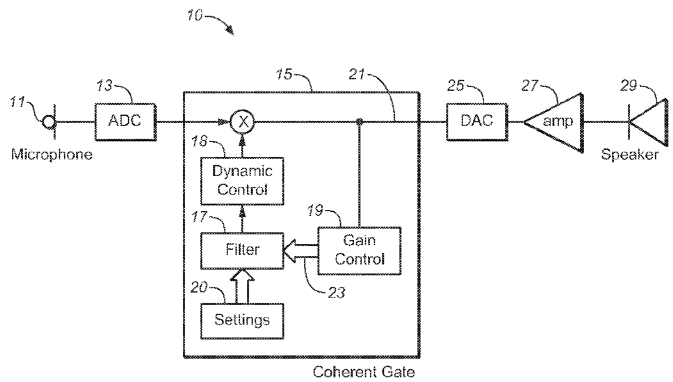

FIG. 1

(57) Abstract: An improved open-ear hearing aid to compensate for hearing loss includes a microphone 11 for picking up incident sound and converting it to an electrical audio signal. An ear insert 37 positionable within a human ear canal 35 is provided for producing an output sound amplified within one or more frequency bands in response to incident sound picked up by the microphone. The in-band gain of the amplified sound output of the ear insert's loudspeaker 29 is dependent on the user's hearing loss characteristics and the sound pressure levels of the incident sound. The form of the ear insert allows transmission of incident sound directly to the eardrum 39, where it is summed at the eardrum with the amplified sound output from the ear insert. Sound output is maximum at low incident sound pressure levels and minimum when the incident sound exceeds a set cut-off level.

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Hearing Aid Having
Level and Frequency-Dependent Gain

Technical Field

[0001] The present invention generally relates to hearing aids and more particularly relates to open-ear type devices that allow incident sound to reach the eardrum directly.

Background Art

[0002] Hearing aids typically consist of a microphone, a signal processor, and an output transducer (sometimes called a “receiver”). The output transducer is placed in the ear canal and can be part of a housing that either leaves the ear canal partially open (i.e., acoustically transparent) or seals the canal completely. Open-ear devices are generally preferred over closed-ear devices by users and are recommended whenever possible for persons with mild or moderate hearing loss. (Open hearing aids have inherent limitations in the amount of gain they can provide, and thus are not well suited for persons whose hearing loss is severe.)

[0003] One advantage of open-ear devices is comfort: the soft tip of open-ear designs is less irritating and easier to adapt to than hard-shell closed-ear inserts. There is also less risk of infection or impaction by cerumen (ear wax). No custom ear-mold is required, which substantially decreases the fitting time and allows such hearing aids to be used off the shelf with only minor modifications. Also avoided is the occlusion effect, where the closed ear canal forms a resonant chamber that boosts low frequency sounds generated by the user (such as speech or chewing), causing the user’s voice to sound unnatural and boomy. The occlusion effect is one of the primary reasons cited when users reject closed-style hearing aids.

[0004] Open-ear designs also allow better processing in complex acoustic environments, because they allow the incident sound to be heard at frequencies where the hearing aid provides no amplification. For example, a hearing aid fit to a high frequency hearing loss (above 1 kHz) doesn’t need to amplify low frequencies. The incident sound is worth preserving whenever possible because it carries perceptual cues required for localizing sound sources and rejecting background noise. Such perceptual cues include interaural timing differences, interaural loudness differences, and phase effects.

[0005] Despite their advantages, open ear hearing aids have significant drawbacks. One drawback comes from artifacts and distortion that can be produced at the eardrum by the combination of incident and amplified sound at frequencies amplified by the hearing aid. These artifacts and distortion are often noticed by users and result in dissatisfaction that leads many to stop using their hearing aids after a short period of time.

[0006] One artifact results from the latency of the hearing aid, that is, the time delay between when a sound is sensed at the microphone and when it is converted to an acoustical sound wave at the hearing aid's output transducer. For modern digital hearing aids, the latency is 3-7 milliseconds; older analog hearing aids have a latency around 1-2 milliseconds. When both the incident and amplified sounds are similar in level, non-zero latency causes comb filtering, a form of spectral distortion. Comb filtering is characterized by a series of regularly spaced spectral peaks and dips in the sound pressure at the eardrum. For longer latencies, the first dip is at a lower frequency and hence a larger portion of the frequency spectrum is affected. Shorter latencies produce less extensive comb filtering. The human ear is very sensitive to this kind of artifact; latencies shorter than 8 milliseconds are perceived as tone coloration, while longer latencies can be perceived as echos, beating, or tone coloration depending on the relative loudness of the delayed sound.

[0007] Another recombination artifact arises from phase distortion in the amplified sound. This also produces a structure of spectral dips and peaks; wherever frequencies are 180 degrees out of phase, they recombine destructively and create a dip, while those in phase add constructively, creating a peak. Since phase distortions are often spread non-uniformly over the frequency spectrum, this kind of artifact is potentially much less regular than latency artifacts. The source of phase distortion can be any component in the signal path: the microphone, signal processing components, or the output transducer (loudspeaker).

[0008] The above-mentioned artifacts result in spectral distortions to the perceived sound readily apparent to even untrained listeners. In addition to these spectral distortions, hearing aids also distort the phase information when the amplified signal is much louder than the incident signal. It is believed that such phase distortions are themselves noticeable. Recent evidence suggests that phase is used for many tasks, including source localization, speech encoding, and detection of phase modulation.

[0009] The present invention addresses the drawbacks associated with conventional open hear hearing aids. It substantially mitigates the artifacts and distortion problems that exist in open-hear hearing aids, and substantially eliminates the source of user dissatisfaction with this type of hear aid design. The invention allows the user to enjoy the well-known benefits of open-ear designs without suffering the perceptible distractions commonly associated with such designs.

Summary of Invention

[0010] The present invention is directed to an open-ear hearing aid comprised of input means such as a microphone for picking up incident sound to be received by the human ear and converting it to an electrical audio signal, and output means including an output transducer positionable within a human ear canal for producing a sound output in the ear in response to incident sound picked-up by the input means. The output means, which can be in the form of an ear piece or insert having a loudspeaker, is acoustically transparent to allow the transmission of incident sound directly to the eardrum, where it combines with the sound output from the output transducer. The perceived sound heard by the wearer of the hearing aid results from the combination of incident sound and sound output from the output means positioned in the ear.

[0011] The invention further includes a signal processing means for processing the electrical audio signal produced by the input means in order to drive the output transducer of the output means in a desired manner. The signal processing means has a variable gain filter (sometimes referred to herein as a "coherent gate") that causes amplified sound output from the output transducer to have the following characteristics:

i) the sound output is amplified within a frequency band set in accordance with the user's hearing loss characteristics;

ii) the gain of the amplified sound output within said frequency band is dependent on the loudness, i.e. sound pressure levels, of the incident sound; and

iii) the output transducer produces no perceptible sound output when the incident sound pressure level exceeds a pre-established level, whereby the sound perceived by the wearer is almost entirely the result of incident sound.

[0012] In another aspect of the invention the signal processing means produces a sound output from the output transducer characterized in that the gain of the amplified sound output within the set frequency band decreases from a maximum gain at low incident sound pressure levels to a minimum gain at incident sound pressure levels near the set cut-off sound pressure level for incident sound.

[0013] In a further aspect of the invention the input means for picking up incident sound to be received by the human ear converts the incident sound to a digital audio signal, and the signal processing means includes a digital signal processor.

[0014] In still another aspect of the invention the gain of the amplified sound output within the frequency band decreases substantially linearly with increasing low incident sound pressure levels at incident sound pressure levels below the set cut-off sound pressure level for incident sound.

[0015] Still further aspects of the invention include having the gain of the amplified sound output within said frequency band decrease rapidly near the cut-off sound pressure level for incident sound and decrease to below 0 dB at the cut-off sound pressure level for incident sound.

[0016] In yet another aspect of the invention the gain of the amplified sound output within the frequency band decreases monotonically and without discontinuities near the cut-off sound pressure level for incident sound.

[0017] In yet further aspects of the invention the phase distortion of the amplified sound output within the frequency band approaches zero near the cut-off sound pressure level for incident sound, becomes zero when the incident sound pressure level substantially exceeds the cut-off level, and approaches zero monotonically and without discontinuities near the cut-off sound pressure level for incident sound.

[0018] In still another aspect of the invention the signal processing means produces the following additional characteristic in the sound output that combines with incident sound: when transitioning between a state where the sound output is amplified and where the output transducer produces substantially no sound output, the transition is under dynamic control to produce desired attack and release times.

[0019] The present invention is also directed to a method of compensating for hearing loss in an individual having hearing loss. The method generally comprises first determining the frequency dependent hearing loss characteristics of the individual, including a loudness threshold of audibility above which the individual has substantially normal hearing capabilities. Two paths for incident sound to travel to the eardrum of the individual's ear having hearing loss are provided, including a direct open ear path and a processed signal path. The processed signal path delivers a sound output at the individual's eardrum that combines with incident sound arriving at the eardrum through the open ear direct path and more particularly delivers a sound output at the eardrum having the following characteristics:

i) the sound output is amplified within a frequency band set in accordance with the user's hearing loss characteristics;

ii) the gain of the amplified sound output within said frequency band is dependent on the sound pressure levels of the incident sound; and

iii) the output transducer produces substantially no sound output when the incident sound pressure level approximately exceeds the individual's threshold of audibility, whereby the sound perceived by the individual is almost entirely the result of incident sound arriving at the eardrum through the open ear direct path.

[0020] The present invention provides a number of benefits. By attenuating the amplified sound at the user's threshold of audibility, the output transducer of the hearing aid does not need to provide a loud output level, and hence can be used without danger of clipping or limiters. Both limiters and clipping introduce harmonic distortion in the amplified signal; limiters do so by design, to avoid the more extreme artifacts caused by clipping, which is the excitation of nonlinear modes in the diaphragm.

[0021] Furthermore, the invention will increase the number and quality of spatial cues available to the user. Such cues result from the complete head-related transfer function, which is shaped by the external ear anatomy (pinna and concha), the ear canal, and binaural effects caused by the head (such as interaural loudness, timing, and phase differences). Whenever a frequency is amplified, latency and phase distortions are necessarily introduced at that frequency and natural cues are perturbed. The invention, and particularly the coherent gate of the invention, preserves natural cues by judicious amplification of incident sound.

[0022] On a more general level, the invention improves sound quality perceived by the user while preserving natural cues, so that the hearing aid is the least taxing for the user. In complex auditory environments, the brain can use multiple cues to separate sound sources and direct auditory attention. In many cases, loss of such cues results in reduced comprehension or intelligibility. However, recent studies have shown that loss of certain cues may also increase the cognitive effort required to maintain the same performance. This is shown most succinctly by giving the test subject a second, non-auditory task to perform along with the primary auditory task. With hearing loss, degraded input quality, or other factors that increase cognitive load, performance on the second task will drop dramatically and the patient will fatigue much more quickly than normal.

[0023] Other aspects and benefits of the invention will be apparent from the description and claims which follow.

Brief Description of the Drawings

[0024] Fig 1 is a functional block diagram representation of a aid having a coherent gate in accordance with the invention.

[0025] Fig. 2 is a graphical representation of physical components for a hearing aid in accordance with the invention, and component placement both outside and inside the ear canal.

[0026] Fig. 3 is a schematic/graphical representation of the paths taken by amplified and incident sound as they pass through the ear canal to the eardrum of a person fitted with a hearing aid in accordance with the invention; it illustrates how the incident and amplified sound sum and create the perceived sound when they reach the eardrum.

[0027] Fig. 4 is a graph of the sound level produced by a hearing aid in accordance with the invention at the user's eardrum as a function of frequency and for different indicated flat (white noise) input sound pressure levels. It shows the decrease in gain of the hearing aid within a customized frequency band fitted to the user as the flat input SPL rises.

[0028] Fig. 5A is an input/output curve at the peak frequency of 4 kHz for the example shown in Fig 4, showing perceived, amplified, and incident sound measured in dB SPL at the eardrum.

[0029] Fig. 5B is a gain function at the peak frequency of 4 kHz for the example shown in Fig. 4.

[0030] Fig. 6 is a graph showing incident sound and sound amplified by a hearing aid in accordance with the invention and their summation at the eardrum for an input sound with an

SPL level slightly below the cross-over point.

[0031] Fig. 7 is a graph of phase as a function of frequency at the gain levels indicated on Fig 5.

[0032] Fig. 8 is a flow chart illustrating the overall method of the invention.

<u>Best Mode for Carrying out the Invention</u>

[0033] Referring to the drawings, Fig. 1 illustrates in block diagram form an embodiment of a hearing aid in accordance with the invention, generally denoted by the numeral 10, wherein input (incident) sound is transduced by the microphone 11 and digitized by an analog-to-digital converter 13 for digital processing. (It will be understood that the invention is not limited to digital processing, and could be implemented instead with analog components.) The signal is then passed through a signal processing circuit having a coherent gate 15 comprised of a filter 17, a gain control function 19 for providing variable gain, and preferably a later described dynamic control function represented by block 18. The filter's parameters (shape, bandwidth, gain structure, etc.) are set via a settings function within the coherent gate as represented by settings block 20. When in a settings or programming mode, the parameters of the coherent gate (including frequency and gain, among others) can be set to the user's particular hearing loss. These settings function can be controlled by a computer.

[0034] As represented by gain control block 19, the gain supplied by the hearing aid can be determined from the coherent gate's output signal at gate output 21 in a feedback configuration, and can be used to modify the amplitude of the filter as represented by feedback arrow 23. The output signal can then be converted to an analog signal by a digital-to-analog convertor 25, amplified by amplifier 27, and passed to output transducer (loudspeaker) 29. It will be appreciated that gain control could be implemented in ways other than described above, for example, using a feed-forward signal.

[0035] Most suitably, the input transducer (microphone) and output transducer (loudspeaker) will reproduce the audio signal accurately without adding spectral or phase distortion. This requires linear transducers with a flat phase response and no harmonic distortion up to the highest level of gain needed. Since hearing losses appropriate to this invention are mild to moderate, the hearing aid will rarely need to provide levels in excess of 80 dB SPL.

[0036] A physical implementation of a hearing aid in accordance with the invention is shown in Fig 2. The microphone 11 is connected to an electronic package containing coherent gate 15 and the output transducer 29 by a wire 31. While the processing electronics (which includes the coherent gate) and microphone are shown as separate components, it is contemplated that they can be housed together in a single wearable unit. A power supply such as a battery 33 can likewise be housed together with the processing electronics or may be located separately and attached to the circuit by a wire. An ear insertable acoustic output means includes output transducer 29 and an acoustically transparent ear insert 37. Suitably, the transducer is embedded in the ear insert. The insert is held in the outer portion of the ear canal 35, which means that the incident sound is not appreciably attenuated and can still reach the eardrum 39. Such an ear insert is called an 'open-ear' design, in contrast to a hard ear insert that blocks the ear canal completely and attenuates the incident sound. (Where the context requires, reference herein to "ear insert" shall be understood to include the transducer 29.)

[0037] Such an open ear insert allows incident sounds to reach the eardrum, as shown schematically in Fig 3. When the device is worn and inactive, the sound perceived at the eardrum 39 (represented by output arrow 41) is simply the incident sound (represented by input arrow 43). When the device is active, it creates a sound sometimes referred to herein as the "amplified sound." Both the amplified sound (represented by arrow 45) and the incident sound 43 excite the eardrum; the sound perceived by the brain is thus their summation.

[0038] As above-mentioned, the frequency spectrum of the amplified sound is determined by the parameters of the filter 17 of coherent gate 15, which can be controlled by a computer via the coherent gate's setting function 20. (A computer interface can be provided to programmatically determine the filter shape of the coherent gate.) The filter can be thought of as an equalization curve, applying gain separately to narrow bands of frequency. The shape of the filter is highly customizable and can be adapted to most kinds of mild or moderate hearing loss, although ultimately it is limited by the design of the coherent gate algorithm. For instance, the filter may be flat across all frequencies, boosted at particular frequencies (high-pass, low-pass, or band-pass), or bimodal (peaking at two frequencies).

[0039] The characteristics of the coherent gate 15 of the signal processing circuit can first be established by setting a frequency-dependent gain (equalization) curve, hence "filter," tailored to the user's particular measured hearing loss. The filter thusly established is preferably a minimum phase filter, that is, a filter where phase is altered only at those frequencies that are amplified. As the input level (incident sound) in one frequency band increases, the filter gain can gradually be attenuated until the incident sound becomes dominant. The gain can be attenuated in such a way that the phase response also gradually decreases to zero. The precise filter characteristics needed to compensate for the hearing loss for a particular individual can be referred to as a "fitting algorithm."

[0040] Fitting algorithms for a user's particular hearing loss can be determined by testing the hearing of the user. The fitting algorithm can provide customized gain control for the coherent gate (filter) circuit: it amplifies a given frequency band only when below the user's threshold of audibility. When amplifying soft sounds, the phase delay of the filter is acceptable to the user and audibility for low level speech and music is greatly improved. Once the input signal reaches the user's threshold, however, the effects of the filter are removed, preferably rapidly, which also removes distortion. (If the filter remains active above the threshold of audibility, the resulting sound is heard as distorted and unpleasant to the user: the perception can be bright or boomy, depending on the type of hearing loss.)

[0041] Other characteristics of the coherent gate are the dynamic properties of each filter. These include the attack and release times, which are the time required for a filter to fully engage as the loudness of incident sound rises above the person's threshold of audibility and to fully disengage as the loudness of incident sound falls below this threshold. By employing dynamic control, (graphically represented by block 18 in Fig. 1), the attack and release times can be suitably set such that sudden loud events aren't amplified, requiring a fast attack time, and such that soft sounds following a loud event remain audible, which requires a moderately fast release time. If either parameter is too long or too short, there will be tone coloration and noticeable level fluctuation; if the release time is too short, pumping artifacts will be noticed. The values of the dynamics will likely depend on the user's particular hearing loss and subjective feedback from the user during the fitting process. Generally, the filter attack time would suitably be set somewhere between about 15 microseconds and about 10 milliseconds, and preferably less than about 1 millisecond. The filter release time would preferably be in a range of about 200 microseconds to 30 milliseconds. These dynamics would most suitably be set by the manufacturer or trained professional.

[0042] While the hearing aid described above is a single channel device for one ear, it shall be understood that an appropriate combination of two such devices could be used for both ears. In such a case, the combination could share a physical enclosure for the electronics and a battery, but each ear would require its own ear insert, and preferably each ear would have its own a dedicated microphone and coherent gate. Separate microphones are recommended to preserve binaural cues, which are different at each ear. The coherent gate will preferably be independently set for each ear because hearing loss in each ear is often different (called asymmetric hearing loss). The microphones will preferably be worn as close to the ear as possible.

[0043] Reference is now made to an exemplary filter shape, which is represented in Fig. 4 and which is a band-pass filter with a peak frequency (F_peak) at 4 kHz. Such a filter corresponds to a typical noise-induced hearing loss of 20 dB at 4 kHz. In accordance with the invention, any filter shape can be realized by the coherent gate. First, the sound arriving and summed at the eardrum must be considered. As illustrated by Fig. 3, this is the combination or sum of incident sound passing through the ear insert directly to the eardrum and the amplified sound. In the filter example shown in Fig. 4, the hearing aid boosts frequencies only around 4 kHz. For a flat input signal of 0 dB SPL, the summed sound at the eardrum at 4 kHz is boosted to 20 dB SPL, making those frequencies now audible to the user. (Other parts of the frequency spectrum, already audible, aren't amplified.) At the higher input levels of 10 dB and 20 dB shown on the graph, it can be see that the degree of boost is progressively reduced. Once the input level reaches 23 dB, the chosen cut-off level or threshold of audibility for a hypothetical wearer, the hearing aid essentially provides no amplified sound at 4kHz (the gain is less than -20 dB). Above this threshold, the incident sound within the hearing loss frequency range will be perceived by the wearer without compensation. In this example, the incident sound producing the input signals for processing are first considered to be static, having a loudness and crest factor that don't vary in time.

[0044] Figs. 5A shows the level of the incident sound (represented by dashed line 49) and the amplified sound (represented by dashed line 51, and the level of sound at the eardrum resulting from the summation of the two (represented by solid line 47), as a function of the input (incident) sound level; Fig. 5B show how the filter gain (represented by dashed line 50) changes as a function of the input sound level. As the input sound level changes, the gain parameters of the filter are made to change, resulting a sound level at the eardrum that changes. At low input sound levels (below approximately 10 dB), the sound arriving at the eardrum and ultimately the perceived sound is seen to be dominated by the amplified sound. In this low input region, the gain of the filter is seen to decrease almost linearly. Above this region is a "cross-over region" (denoted by the numeral 55 in Fig. 5B) where the difference between amplified sound 51 and incident sound 49 are less than about 8 dB. At levels within this cross-over region both incident and amplified sound contribute significantly to the sound arriving at the eardrum and to the perceived sound. As a result, there can be a desirable deviation from linearity in the gain function within this region (this deviation in the cross-over region can be noted in Fig. 5B). Nonetheless, to prevent perceptual artifacts in the cross-over region, changes in the gain function should be gradual; that is, it should be monotonically decreasing, without discontinuities, and smooth (in the mathematical sense, with continuous derivatives). Effectively, such a well-defined gain function maps similar input levels to similar output levels; a small change in input level causes a small change to the output level. While the optimal gain function is nonlinear as shown in Fig. 5B, it should be noted that a linear gain function, which is effective and easier to implement, could also be used.

[0045] Fig 6 shows incident sound (represented by line 57) and amplified sound (represented by line 59) and their summation (represented by line 61) at the eardrum for incident sound having a sound pressure level within the cross-over region (input level at 16 dB). In the cross-over region, the phase and delay characteristics of the amplified sound are particularly important. The frequency-dependent phase must gradually approach zero as the filter gain decreases (just as the gain function changes). As with the amplified sound pressure level, if the phase changes dramatically between slight changes in input level, it will be noticed by the wearer of the device.

[0046] One way to avoid unacceptably large and perceptible phase changes with small changes in input level is illustrated in Fig. 7, which shows the phase perturbation slowly decreasing to zero as the input level rises and the system gain decreases. Fig. 7 shows phase as a function of frequency at the gain levels indicated on Fig. 5B. As the gain decreases to below zero, the phase perturbation also decreases. For example, at +20 dB gain the phase perturbation (represented by graph line 61) is large as compared to the phase perturbation at 0 dB (represented by graph line 63). At -10 dB there is virtually no phase perturbation. Although filters that provide frequency-dependent gain necessarily introduce a phase shift, this shift can be minimized by selecting an appropriate filter implementation (e.g., minimum phase filters).

[0047] The other important parameter of the hearing aid is latency, the time between the incident sound's arrival at the microphone and the output of the amplified sound at the loudspeaker. This delay needs to be kept as small as possible, ideally less than 1 millisecond. Delays longer than ~5 milliseconds create artifacts of coloration, while delays longer than 1 millisecond affect sound localization cues. Thus, preferably, the latency introduced by the coherent gate 15 of the signal processing circuit illustrated in Fig. 1 will be less than 1 millisecond.

[0048] In order to realize the benefits of the above-described processing scheme, the input transducer (microphone) and output transducer (loudspeaker) should be capable of reproducing the audio signal with great fidelity. The equal-phase response of the coherent gate will not be realized unless both the input and output transducers are linear; that is, unless they have a flat phase response and low harmonic distortion (preferably less than 1%) at the loudest expected output level.

[0049] Fig. 8 illustrates the general methodology of the above-described embodiment of the invention, where incident sound, represented by block 101 can arrive at wearer's eardrum via two paths, represented by arrows A and B, where it is summed, as represented by block 103. As shown in Fig. 8, the first path (path A) is a direct open ear path to the eardrum permitted by the open-ear configuration of the ear insert for the hearing aid. Incident sound will always arrive at the eardrum via this path. The path (path B) is a processed signal path that provides to the eardrum amplified sound that is dependent on frequency and incident sound level. Via this path, incident sound that has been converted to an electrical audio signal is processed by a variable gain gating function, shown as a coherent gate 15 in Fig. 1, wherein the incident sound arriving at the eardrum via path A is augmented by amplified sound arriving via path B. The level of the sound arriving from path B not only depends on the band of frequencies where compensation for hearing loss occurs, but also by the level of incident sound at any point of time. The characteristics of the variable gain function for amplifying the audio signal processed through this path will be tailored to the measured hearing loss profile of the wearer, including the wearer's threshold of audibility.

[0050] More particularly, in the processed signal path B, incident sound is introduced to this path via microphone 105, which converts the sound to an electrical audio signal that can be processed by analog circuits or most preferably by digital signal processing. The processing steps include first determining loudness of the incident sound in the frequency band or bands of interest (block 107). If the loudness of the incident sound picked up by the microphone is below the measured threshold of audibility for the wearer (block 109), the gain necessary to compensate for the wearer's measured hearing loss, that is, to bring the below threshold sound up to an audible level for the wearer, is determined such as by a gain calculation (block 111). Based on this determined gain, the filter of the coherent gate is engaged (block 113) to allow the audio signal passing through path B to be amplified to a level determined by the gain. As earlier described, the engagement of the filter can be under dynamic control such that the attack time can be set at desired levels. The resulting amplified sound is used to drive loudspeaker 115 of an ear insert. The output from the loudspeaker produces amplified sound that is summed with incident sound at the eardrum..

[0051] If on the other hand the loudness of the incident sound picked up by the microphone is above the measured threshold of audibility for the wearer (back to block 109), the filter of the coherent gate is disengaged (block 117), thus removing any audio signal that may drive the loudspeaker 115. As with the engagement of the filter, disengagement of the filter can be under dynamic control wherein the release time can be set as earlier described. During release, amplified sound will continue to drive loudspeaker 115 for a very short period of time.

[0052] While the invention has been described in detail in the foregoing specification, it is not intended that the invention be limited to such detail, except as necessitated be the following claims.

WHAT WE CLAIM IS:

1. An open-ear hearing aid for compensating for loss of hearing in the human ear comprising:

input means for picking up incident sound to be received by the human ear and converting it to an electrical audio signal,

output means including an output transducer positionable within a human ear canal for producing a sound output in response to incident sound picked-up by said input means, said output means having a form that allows transmission of incident sound directly to the eardrum, where it combines with the sound output from said output transducer, the combination of incident sound and sound output from said output means resulting in the sound perceived by the wearer of the hearing aid, and

signal processing means for processing the electrical audio signal from said input means, wherein the characteristics of the sound output from said output transducer are at least in part determined by said signal processing means, said signal processing means producing the following characteristics in the sound output that combines with incident sound:

i) the sound output is amplified within a frequency band set in accordance with the user's hearing loss characteristics;

ii) the gain of the amplified sound output within said frequency band is dependent on the sound pressure levels of the incident sound; and

iii) the output transducer produces substantially no sound output when the incident sound pressure level exceeds a set cut-off level, whereby the sound perceived by the wearer is almost entirely the result of incident sound arriving directly at the eardrum.

2. The open-ear hearing aid of claim 1 wherein the signal processing means produces a sound output from said output transducer characterized in that the gain of the amplified sound output within the set frequency band decreases from a maximum gain at low incident sound pressure levels to a minimum gain at incident sound pressure levels near the set cut-off sound pressure level for incident sound.

3. The open-ear hearing aid of claim 1 wherein said input means for picking up incident sound to be received by the human ear converts the incident sound to a digital audio signal, and wherein said signal processing means is a digital signal processor.

4. The open-ear hearing aid of claim 1 wherein said input means for picking up incident sound to be received by the human ear and converting it to an electrical audio signal includes a

microphone worn by the user.

5. The open-ear hearing aid of claim 1 wherein the gain of the amplified sound output within said frequency band decreases substantially linearly with increasing low incident sound pressure levels at incident sound pressure levels below the set cut-off sound pressure level for incident sound.

6. The open-ear hearing aid of claim 1 wherein the gain of the amplified sound output within said frequency band decreases rapidly near the cut-off sound pressure level for incident sound.

7. The open-ear hearing aid of claim 1 wherein the gain of the amplified sound output within said frequency band decreases to below 0 dB at the cut-off sound pressure level for incident sound.

8. The open-ear hearing aid of claim 1 wherein the gain of the amplified sound output within said frequency band decreases monotonically and without discontinuities near the cut-off sound pressure level for incident sound.

9. The open-ear hearing aid of claim 1 wherein the phase distortion of the amplified sound output within said frequency band approaches zero near the cut-off sound pressure level for incident sound and becomes zero when the incident sound pressure level substantially exceeds the cut-off level.

10. The open-ear hearing aid of claim 1 wherein the phase distortion of the amplified sound output within said frequency band approaches zero monotonically and without discontinuities near the cut-off sound pressure level for incident sound.

11. The open-ear hearing aid of claim 1 wherein the signal processing means produces the following additional characteristic in the sound output that combines with incident sound: when transitioning between a state where the sound output is amplified and where the output transducer produces substantially no sound output, the transition is under dynamic control to produce desired attack and release times.

12. An open-ear hearing aid for compensating for loss of hearing in the human ear, comprising:

input means for picking up incident sound to be received by the human ear and converting it to an electrical audio signal, wherein said input means for picking up incident sound to be received by the human ear converts the incident sound to a digital audio signal,

output means including a digital processor and an output transducer positionable within a human ear canal for producing a sound output in response to incident sound picked-up by said input means, said output means having a form that allows transmission of incident sound directly to the eardrum, where it combines with the sound output from said output transducer, the combination of incident sound and sound output from said output means resulting in the sound perceived by the wearer of the hearing aid, and

signal processing means for processing the electrical audio signal produced by said input means, wherein the characteristics of the sound output from said output transducer are at least in part determined by said signal processing means, and wherein the signal processing means produces a sound output from said output transducer characterized in that the gain of the amplified sound output within the set frequency band decreases from a maximum gain at low incident sound pressure levels to a minimum gain at incident sound pressure levels near the cut-off sound pressure level for incident sound,

said signal processing means producing the following characteristics in the sound output that combines with incident sound:

i) the sound output is amplified within a frequency band set in accordance with the user's hearing loss characteristics;

ii) the gain of the amplified sound output within said frequency band is dependent on the sound pressure levels of the incident sound and decreases substantially linearly with increasing low incident sound pressure levels;

iii) the output transducer produces substantially no sound output when the incident sound pressure level exceeds a set cut-off level, whereby the sound perceived by the wearer is almost entirely the result of incident sound arriving directly at the eardrum; and

iv) when transitioning between a state where the sound output is amplified and where the output transducer produces substantially no sound output, the transition is under dynamic control to produce desired attack and release times.

13. The open-ear hearing aid of claim 12 wherein the gain of the amplified sound output within said frequency band decreases substantially linearly with increasing low incident sound pressure levels at incident sound pressure levels below the set cut-off sound pressure level for

incident sound.

14. The open-ear hearing aid of claim 13 wherein the gain of the amplified sound output within said frequency band decreases monotonically and without discontinuities near the cut-off sound pressure level for incident sound.

15. The open-ear hearing aid of claim 14 wherein the phase distortion of the amplified sound output within said frequency band approaches zero near the cut-off sound pressure level for incident sound and becomes zero when the incident sound pressure level substantially exceeds the cut-off level.

16. The open-ear hearing aid of claim 15 wherein the phase distortion of the amplified sound output within said frequency band approaches zero monotonically and without discontinuities near the cut-off sound pressure level for incident sound.

17. An open-ear hearing aid compensating for loss of hearing in the human ear, comprising:

microphone for picking up incident sound to be received by the human ear and converting it to an electrical audio signal,

an ear insert including a loudspeaker positionable within a human ear canal for producing a sound output in response to incident sound picked-up by said microphone, said ear insert having an open ear configuration that allows transmission of incident sound directly to the eardrum, where it combines with the sound output from the loudspeaker of the ear insert, the combination of incident sound and sound output from the loudspeaker of the ear insert resulting in the sound perceived by the wearer of the hearing aid, and

a coherent gate for processing the electrical audio signal from the microphone, said coherent gate having a filter and a gain control function for said filter wherein the characteristics of the sound output from the loudspeaker of said ear insert are at least in part determined by the coherent gate, said coherent gate producing the following characteristics in the sound output that combines with incident sound:

i) the sound output is amplified within a frequency band set in accordance with the user's hearing loss characteristics;

ii) the gain of the amplified sound output within said frequency band is dependent on the sound pressure levels of the incident sound; and

iii) the loudspeaker of the ear insert produces substantially no sound output when the incident sound pressure level exceeds a set cut-off level, whereby the sound perceived by the wearer is almost entirely the result of incident sound arriving directly at the eardrum.

18. The open-ear hearing aid of claim 16 wherein the coherent gate produces the following additional characteristic in the sound output that combines with incident sound: when transitioning between a state where the sound output is amplified and where the output transducer produces substantially no sound output, the transition is under dynamic control to produce desired attack and release times.

19. The open-ear hearing aid of claim 16 wherein said coherent gate has a latency less than 1 millisecond.

20. A method of compensating for hearing loss in an individual having hearing loss comprising:

determining the frequency dependent hearing loss characteristics of the individual, including a loudness threshold of audibility above which the individual has substantially normal hearing capabilities,

providing two paths for incident sound to travel to the eardrum of the individual's ear having hearing loss, including a direct open ear path and a processed signal path,

the processed signal path delivering a sound output at the individual's eardrum that combines with incident sound arriving at the eardrum through the open ear direct path and that delivers a sound output having the following characteristics:

i) the sound output is amplified within a frequency band set in accordance with the user's hearing loss characteristics;

ii) the gain of the amplified sound output within said frequency band is dependent on the sound pressure levels of the incident sound; and

iii) the output transducer produces substantially no sound output when the incident sound pressure level approximately exceeds the individual's threshold of audibility, whereby the sound perceived by the individual is almost entirely the result of incident sound arriving at the eardrum through the open ear direct path.

21. The method of claim 20 wherein, when transitioning between a state where the sound output is amplified and where the output transducer produces substantially no sound output, the transition is under dynamic control to produce desired attack and release times for the amplified sound.

22. The method of claim 20 wherein the gain of the amplified sound output within the set frequency band decreases from a maximum gain at low incident sound pressure levels to a minimum gain at incident sound pressure levels near the individual's threshold of audibility.

23. The method of claim 20 wherein the gain of the amplified sound output within said frequency band decreases substantially linearly with increasing incident sound pressure levels at low incident sound pressure levels below the individual's threshold of audibility.

24. The method of claim 20 wherein the gain of the amplified sound output within said frequency band decreases rapidly near the individual's threshold of audibility.

25. The method of claim 20 wherein the gain of the amplified sound output within said frequency band decreases monotonically and without discontinuities near the individual's threshold of audibility.

26. The method of claim 20 wherein the phase distortion of the amplified sound output within said frequency band approaches zero near the individual's threshold of audibility and becomes zero when the incident sound pressure level substantially exceeds the individual's threshold of audibility.

27. The method of claim 20 wherein the phase distortion of the amplified sound output within said frequency band approaches zero monotonically and without discontinuities near the individual's threshold of audibility.

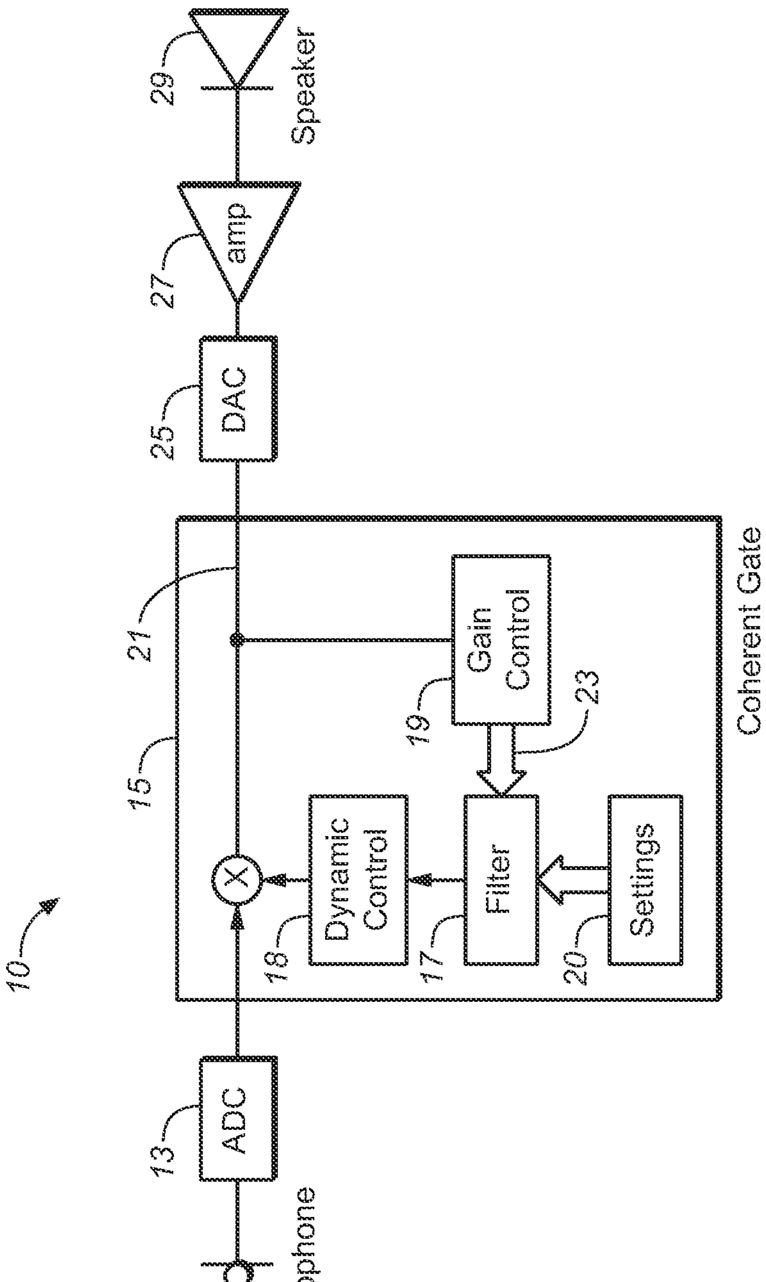
10
11
Microphone
13
ADC
15
18
Dynamic Control
17
Filter
20
Settings
19
Gain Control
23
21
Coherent Gate
25
DAC
27
amp
29
Speaker

FIG. 1

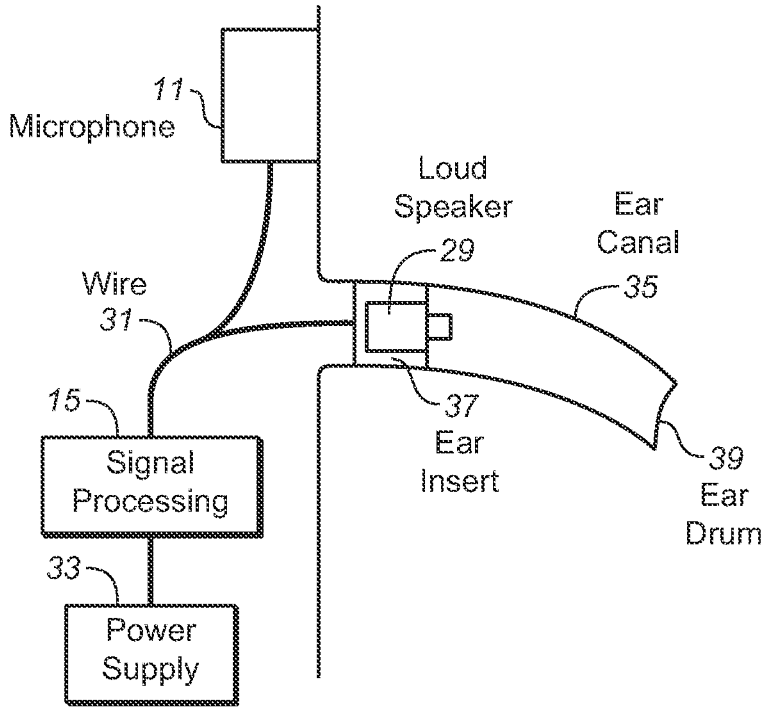
11
Microphone
Loud
Speaker
Ear
Canal
29
Wire
31
35
15
37
Ear
Insert
39
Ear
Drum
Signal
Processing
33
Power
Supply

FIG. 2

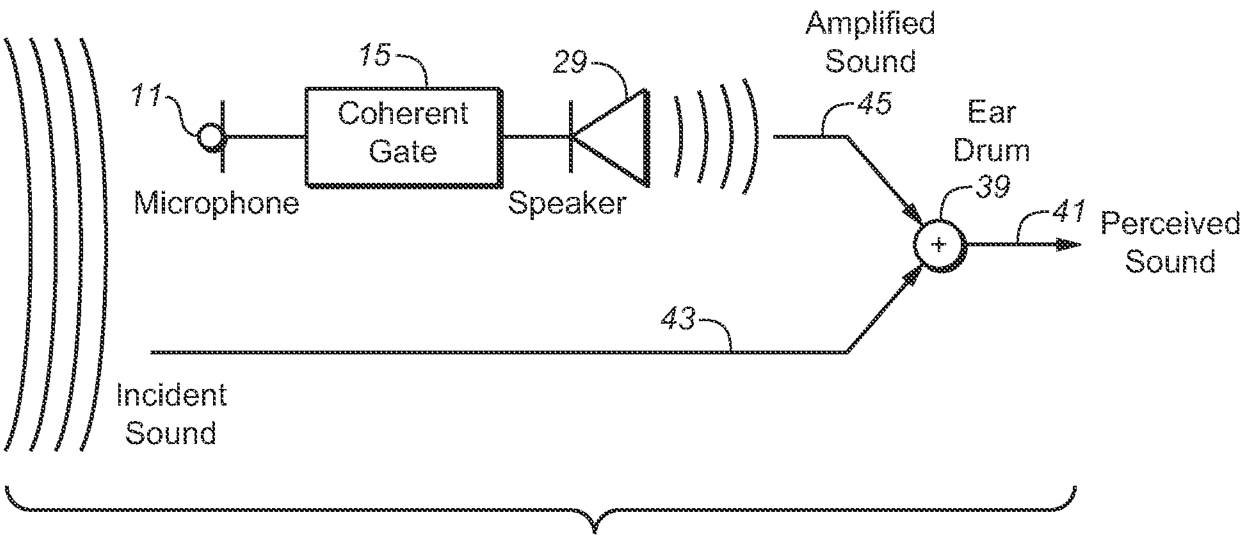
Amplified
Sound
15
29
11
45
Coherent
Gate
Ear
Drum
39
Microphone
Speaker
41
Perceived
Sound
+
43
Incident
Sound

FIG. 3

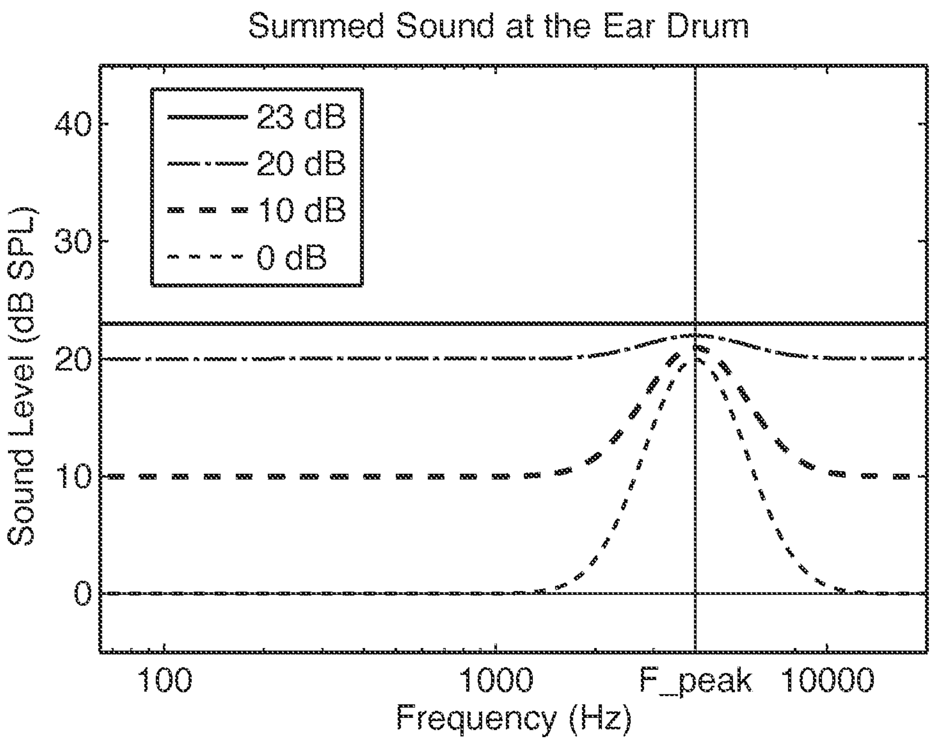

FIG. 4

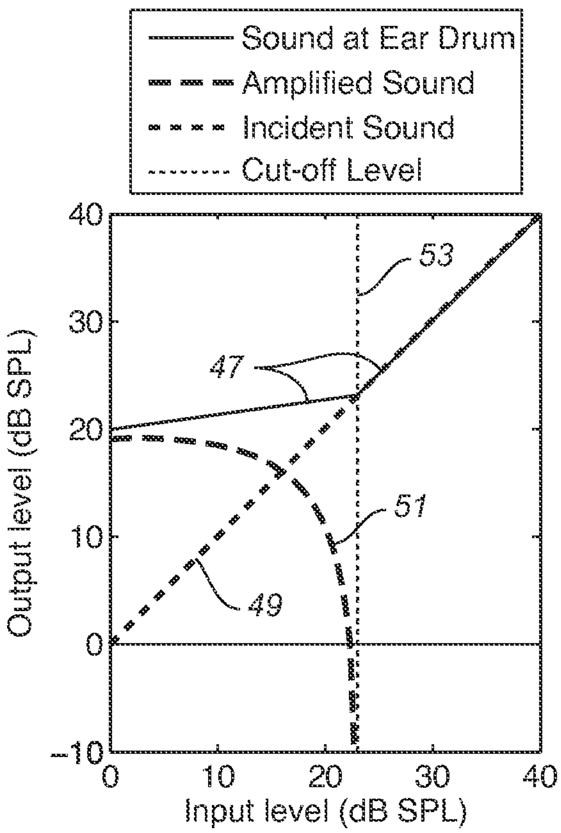

FIG. 5A

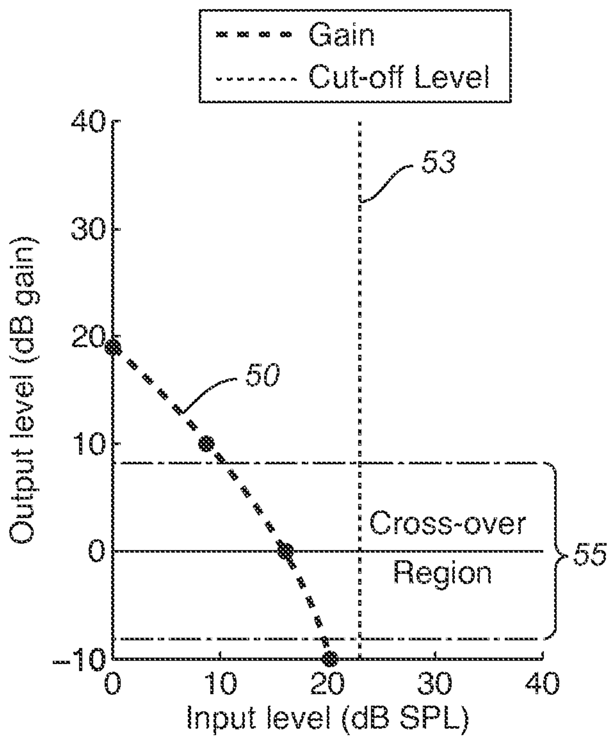

FIG. 5B

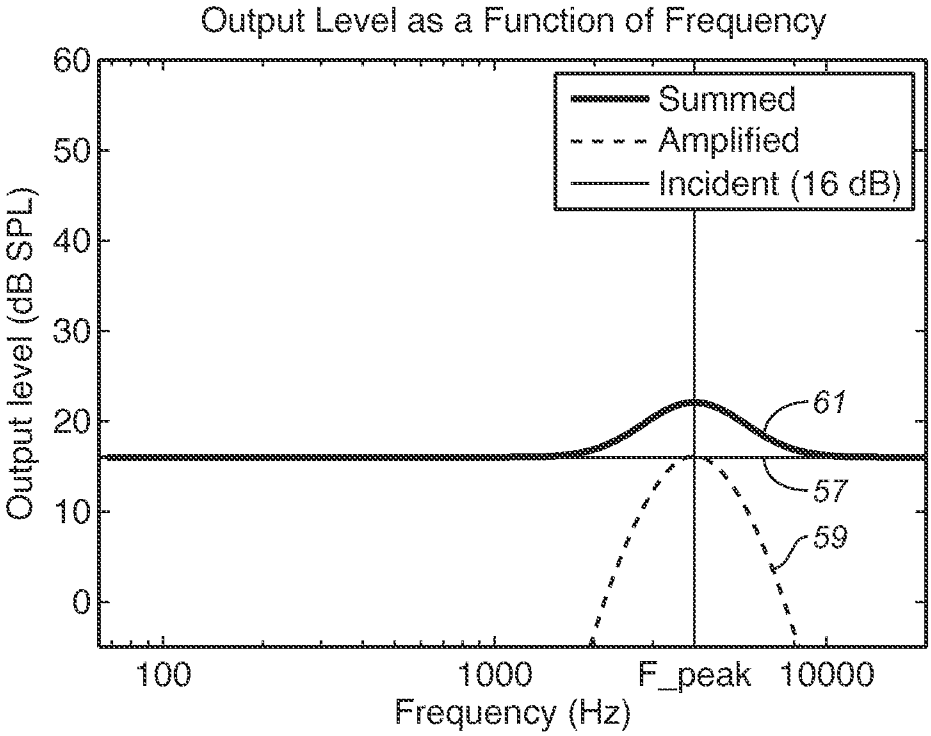
Output Level as a Function of Frequency
Summed
Amplified
Incident (16 dB)
Output level (dB SPL)
60
50
40
30
20
10
0
61
57
59
100
1000
F_peak
10000
Frequency (Hz)

FIG. 6

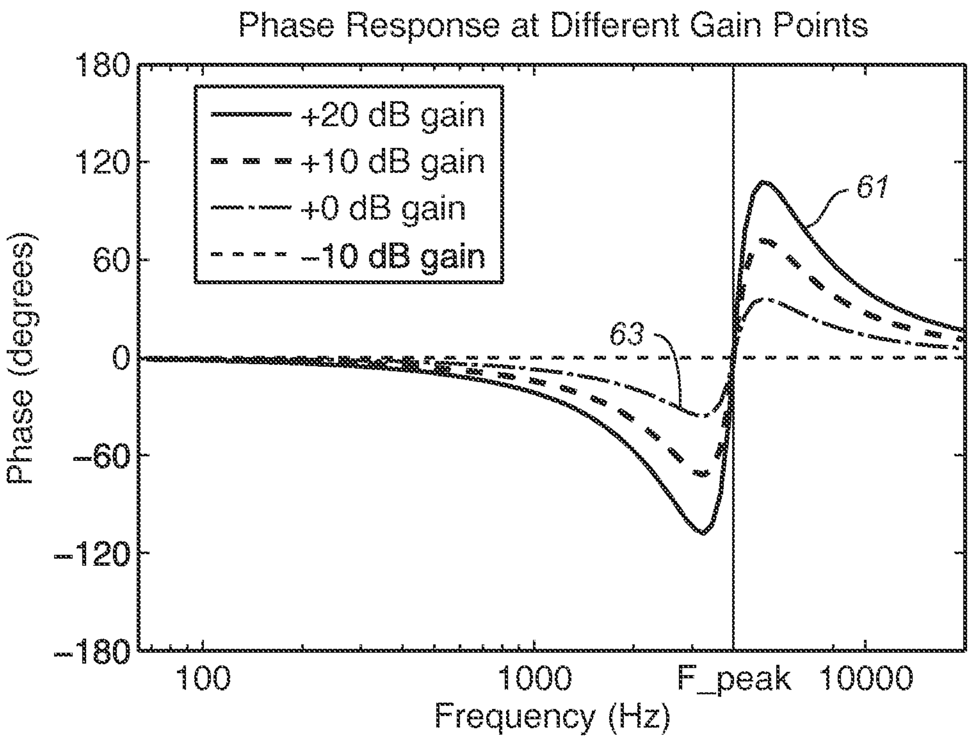
Phase Response at Different Gain Points
+20 dB gain
+10 dB gain
+0 dB gain
−10 dB gain
Phase (degrees)
180
120
60
0
−60
−120
−180
61
63
100
1000
F_peak
10000
Frequency (Hz)

FIG. 7

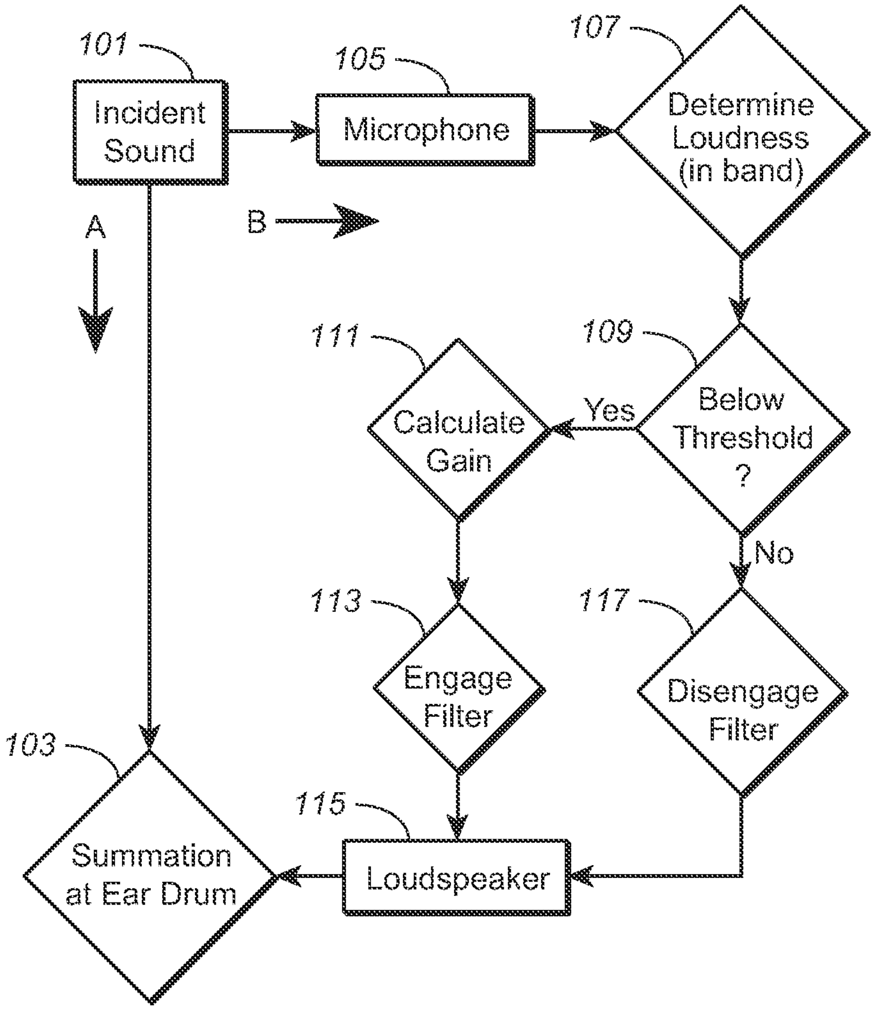
101
Incident Sound
105
Microphone
107
Determine Loudness (in band)
A
B
111
Calculate Gain
Yes
109
Below Threshold ?
No
113
Engage Filter
117
Disengage Filter
103
Summation at Ear Drum
115
Loudspeaker

FIG. 8

(12) INTERNATIONAL APPLICATION PUBLISHED UNDER THE PATENT COOPERATION TREATY (PCT)

(19) **World Intellectual Property Organization**
International Bureau

(43) **International Publication Date**
20 February 2014 (20.02.2014)

(10) **International Publication Number**
WO 2014/028658 A3

(51) **International Patent Classification**:
H03G 9/20 (2006.01) ***G01L 21/02*** (2006.01)
H04R 25/02 (2006.01)

(21) **International Application Number**:
PCT/US2013/055004

(22) **International Filing Date**:
14 August 2013 (14.08.2013)

(25) **Filing Language**: English

(26) **Publication Language**: English

(30) **Priority Data**:
61/683,668 15 August 2012 (15.08.2012) US

(71) **Applicant**: **MEYER SOUND LABORATORIES, INCORPORATED** [US/US]; 2832 San Pablo Avenue, Berkeley, CA 94702 (US).

(72) **Inventors**: **MEYER, John, D.**; 928 Arlington Ave., Berkeley, CA 94707 (US). **SZUTS, Toban, A.**; 7249 Blake Street, El Cerrito, CA 94530 (US).

(74) **Agent**: **BEESON, Donald, L.**; Beeson Skinner Beverly, LLP, One Kaiser Plaza, Suite 750, Oakland, CA 94612 (US).

(81) **Designated States** *(unless otherwise indicated, for every kind of national protection available)*: AE, AG, AL, AM, AO, AT, AU, AZ, BA, BB, BG, BH, BN, BR, BW, BY, BZ, CA, CH, CL, CN, CO, CR, CU, CZ, DE, DK, DM, DO, DZ, EC, EE, EG, ES, FI, GB, GD, GE, GH, GM, GT, HN, HR, HU, ID, IL, IN, IS, JP, KE, KG, KN, KP, KR, KZ, LA, LC, LK, LR, LS, LT, LU, LY, MA, MD, ME, MG, MK, MN, MW, MX, MY, MZ, NA, NG, NI, NO, NZ, OM, PA, PE, PG, PH, PL, PT, QA, RO, RS, RU, RW, SA, SC, SD, SE, SG, SK, SL, SM, ST, SV, SY, TH, TJ, TM, TN, TR, TT, TZ, UA, UG, US, UZ, VC, VN, ZA, ZM, ZW.

(84) **Designated States** *(unless otherwise indicated, for every kind of regional protection available)*: ARIPO (BW, GH, GM, KE, LR, LS, MW, MZ, NA, RW, SD, SL, SZ, TZ, UG, ZM, ZW), Eurasian (AM, AZ, BY, KG, KZ, RU, TJ, TM), European (AL, AT, BE, BG, CH, CY, CZ, DE, DK, EE, ES, FI, FR, GB, GR, HR, HU, IE, IS, IT, LT, LU, LV, MC, MK, MT, NL, NO, PL, PT, RO, RS, SE, SI, SK, SM, TR), OAPI (BF, BJ, CF, CG, CI, CM, GA, GN, GQ, GW, KM, ML, MR, NE, SN, TD, TG).

Published:

— *with international search report (Art. 21(3))*

— *before the expiration of the time limit for amending the claims and to be republished in the event of receipt of amendments (Rule 48.2(h))*

(88) **Date of publication of the international search report**:
30 October 2014

(54) **Title**: HEARING AID HAVING LEVEL AND FREQUENCY-DEPENDENT GAIN

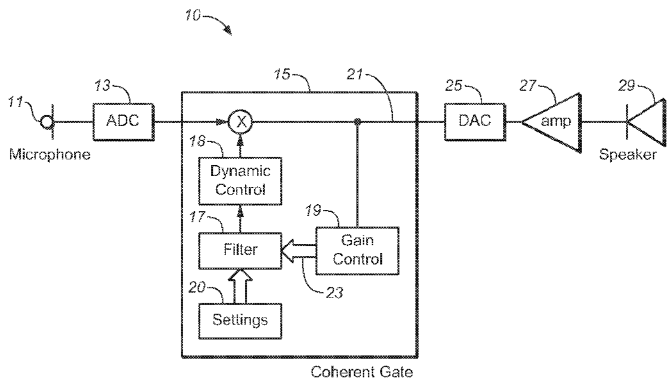

FIG. 1

(57) **Abstract**: An improved open-ear hearing aid to compensate for hearing loss includes a microphone 11 for picking up incident sound and converting it to an electrical audio signal. An ear insert 37 positionable within a human ear canal 35 is provided for producing an output sound amplified within one or more frequency bands in response to incident sound picked up by the microphone. The in-band gain of the amplified sound output of the ear insert's loudspeaker 29 is dependent on the user's hearing loss characteristics and the sound pressure levels of the incident sound. The form of the ear insert allows transmission of incident sound directly to the eardrum 39, where it is summed at the eardrum with the amplified sound output from the ear insert. Sound output is maximum at low incident sound pressure levels and minimum when the incident sound exceeds a set cut-off level.

WO 2014/028658 A3

INTERNATIONAL SEARCH REPORT

International application No.
PCT/US13/55004

A. CLASSIFICATION OF SUBJECT MATTER

IPC(8) - H03G 9/20; H04R 25/02; G10L 21/02 (2013.01)
USPC - 381/320, 106; 704/205
According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)
IPC(8): H03G 9/20; H04R 25/02; G10L 21/02 (2013.01)
USPC: 381/320, 106, 312; 704/205

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practicable, search terms used)
MicroPatent (US-G, US-A, EP-A, EP-B, WO, JP-bib, DE-C,B, DE-A, DE-T, DE-U, GB-A, FR-A); Google; Google Scholar; ProQuest;
KEYWORDS: hearing aid transmit incident sound threshold phase distortion continuous monotonically decrease couple wave

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X — Y	US 5,903,655 A (SALMI, P. et al.) May 11, 1999; figure 1, 2A-2C, 3, 4, 9, 13; column 2, lines 15-30, 52-57; column 3, lines 20-25; column 4, lines 35-50; column 5, lines 19-26; column 6, lines 1-15, 25-40, 50-67; column 7, lines 35-55; column 8, lines 5-15; column 9, lines 30-55	1-9, 11-15, 17, 18, and 20-26 — 10, 16, 19, 27
Y	KASTANOVICH, M. Pass Labs Aleph 3 power amplifier. Datasheet [online]. Stereophile, April 29, 1997 paragraph 5 Retrieved from the Internet: <URL: http://www.stereophile.com/solidpoweramps/674>	10, 16, 27
Y	Advanced Gating Techniques, Part 1. Datasheet [online]. Sound on Sound, (WHITE, P.) April 2001; paragraph 6 Retrieved from the Internet: <URL: http://www.soundonsound.com/sos/apr01/articles/advanced.asp>	19

☐ Further documents are listed in the continuation of Box C. ☐

* Special categories of cited documents:
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"E" earlier application or patent but published on or after the international filing date
"L" document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)
"O" document referring to an oral disclosure, use, exhibition or other means
"P" document published prior to the international filing date but later than the priority date claimed
"T" later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention
"X" document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone
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"&" document member of the same patent family

Date of the actual completion of the international search
30 December 2013 (30.12.2013)

Date of mailing of the international search report
15 AUG 2014

Name and mailing address of the ISA/US
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P.O. Box 1450, Alexandria, Virginia 22313-1450
Facsimile No. 571-273-3201

Authorized officer:
Shane Thomas
PCT Helpdesk: 571-272-4300
PCT OSP: 571-272-7774

(19)中华人民共和国国家知识产权局

(12)发明专利申请

(10)申请公布号 CN 104885360 A

(43)申请公布日 2015.09.02

(21)申请号 201380053923.7

(22)申请日 2013.08.14

(30)优先权数据

61/683,668 2012.08.15 US

(85)PCT国际申请进入国家阶段日

2015.04.15

(86)PCT国际申请的申请数据

PCT/US2013/055004 2013.08.14

(87)PCT国际申请的公布数据

WO2014/028658 EN 2014.02.20

(71)申请人 麦耶声音实验室股份有限公司

地址 美国加利福尼亚州

(72)发明人 J·D·迈耶 T·A·祖茨

(74)专利代理机构 上海专利商标事务所有限公司 31100

代理人 段登新

(51)Int.Cl.

H03G 9/20(2006.01)

H04R 25/02(2006.01)

G01L 21/02(2006.01)

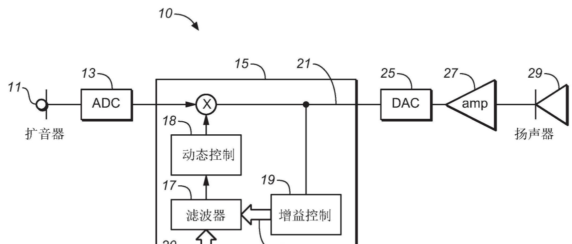
权利要求书4页 说明书8页 附图5页

(54)发明名称

具有声压级和频率依赖性增益的助听器

(57)摘要

一种改良的用于补偿听力损失的开放式助听器，其包括：用于拾取入射声音并且将所述入射声音转换成电音频信号的扩音器11。提供了一种能够放入人耳道35内的耳塞37，所述耳塞响应于由所述扩音器拾取的入射声音而产生声音输出，所述声音输出在一个或多个频带范围内被放大。所述耳塞的扬声器29的已放大的声音输出的带内增益依赖于佩戴者的听力损失特性以及所述入射声音的声压级。所述耳塞的结构允许入射声音直接传送到耳膜39，所述入射声音和所述耳塞的所述已放大的声音输出在耳膜处叠加。于低入射声压级的声音输出是最大值，而当入射声音超出设定的截止声压级时所述声音输出是最小值。



CN 104885360 A

1.一种用于补偿人耳的听力损失的开放式助听器，所述开放式助听器包括：

输入装置，所述输入装置用于拾取人耳接收的入射声音，并且将所述入射声音转换成电音频信号，

输出装置，所述输出装置包括能够放入人耳道内的输出换能器，所述输出换能器响应于由所述输入装置拾取的入射声音而产生声音输出，所述输出装置的形式允许入射声音直接传送到耳膜，所述入射声音与所述输出换能器的声音输出在耳膜处结合，由所述入射声音和所述输出换能器的声音输出结合的结果使所述助听器的佩戴者感知声音，以及

信号处理装置，所述信号处理装置处理所述输入装置的电音频信号，其中，所述信号处理装置至少部分地确定所述输出换能器的声音输出的特征，所述信号处理装置使结合了入射声音的所述声音输出具有以下特征：

i) 所述声音输出在根据佩戴者的听力损失特性而设定的频带范围内被放大；

ii) 在所述频带范围内的已放大的声音输出的增益依赖于所述入射声音的声压级；以及

iii) 当入射声压级超出设定的截止声压级时，所述输出换能器大体上不产生声音输出，佩戴者感知的声音几乎全部是直接到达耳膜的入射声音产生的结果。

2.根据权利要求1所述的开放式助听器，其中，所述信号处理装置产生所述输出换能器的声音输出，所述声音输出的特征是：在设定的频带范围内的已放大的声音输出的增益从低入射声压级的最大增益降低至接近设定的入射声音的截止声压级的入射声压级的最小增益。

3.根据权利要求1所述的开放式助听器，其特征在于，用于拾取人耳接收的入射声音的所述输入装置将所述入射声音转换成数字音频信号，其中，所述信号处理装置是数字信号处理器。

4.根据权利要求1所述的开放式助听器，其特征在于，用于拾取人耳接收的入射声音并且将所述入射声音转换成电音频信号的所述输入装置包括扩音器，所述扩音器由使用者佩戴。

5.根据权利要求1所述的开放式助听器，其特征在于，入射声压级低于设定的入射声音的截止声压级时，随着入射声压级的低入射声压级增大，在所述频带范围内的已放大的声音输出的增益大体上线性降低。

6.根据权利要求1所述的开放式助听器，其特征在于，在所述频带范围内的已放大的声音输出的增益在接近入射声音的截止声压级时迅速降低。

7.根据权利要求1所述的开放式助听器，其特征在于，在所述频带范围内的已放大的声音输出的增益在入射声音的截止声压级降低至低于0dB。

8.根据权利要求1所述的开放式助听器，其特征在于，在所述频带范围内的已放大的声音输出的增益在接近入射声音的截止声压级时单调且非不连续性降低。

9.根据权利要求1所述的开放式助听器，其特征在于，在所述频带范围内的已放大的声音输出的相位失真在接近入射声音的截止声压级时接近零，并且当入射声压级大体上超出截止声压级时变为零。

10.根据权利要求1所述的开放式助听器，其特征在于，在所述频带范围内的已放大的声音输出的相位失真在接近入射声音的截止声压级时单调且非不连续性接近零。

11.根据权利要求1所述的开放式助听器，其特征在于，所述信号处理装置在结合了入射声音的所述声音输出中产生以下附加特征：当在声音输出被放大的状态和在所述输出换能器大体上不产生声音输出的状态之间进行转换时，这种转换处于动态控制，产生所希望的起音时间和释音时间。

12.一种用于补偿人耳的听力损失的开放式助听器，所述开放式助听器包括：

输入装置，所述输入装置用于拾取人耳接收的入射声音，并且将所述入射声音转换成电音频信号，其中，所述用于拾取人耳接收的入射声音的输入装置将所述入射声音转换成数字音频信号，

输出装置，所述输出装置包括能够放入人耳道内的数字处理器和输出换能器，所述输出换能器响应于由所述输入装置拾取的入射声音而产生声音输出，所述输出装置的形式允许入射声音直接传送到耳膜，所述入射声音与所述输出换能器的声音输出在耳膜处结合，由所述入射声音和所述输出换能器的声音输出结合的结果使所述助听器的佩戴者感知声音，以及

信号处理装置，所述信号处理装置处理所述输入装置的电音频信号，其中，所述信号处理装置至少部分地确定所述输出换能器的声音输出的特征，其中所述信号处理装置产生所述输出换能器的声音输出，其特征在于，在设定的频带范围内的已放大的声音输出的增益从低入射声压级的最大增益降低至接近设定的入射声音的截止声压级的入射声压级的最小增益，

所述信号处理装置使结合了入射声音的所述声音输出具有以下特征：

i）所述声音输出在根据佩戴者的听力损失特性而设定的频带范围内被放大；

ii）在所述频带范围内的已放大的声音输出的增益依赖于所述入射声音的声压级，并且随着低入射声压级增大而大体上线性降低；

iii）当入射声压级超出设定的截止声压级时，所述输出换能器大体上不产生声音输出，佩戴者感知的声音几乎全部是直接到达耳膜的入射声音产生的结果；以及

iv）当在声音输出被放大的状态和在所述输出换能器大体上不产生声音输出的状态之间进行转换时，这种转换处于动态控制，产生所希望的起音时间和释音时间。

13.根据权利要求12所述的开放式助听器，其特征在于，随着入射声压级的低入射声压级增大，所述入射声压级低于设定的入射声音的截止声压级，在所述频带范围内的已放大的声音输出的增益大体上线性降低。

14.根据权利要求13所述的开放式助听器，其特征在于，在所述频带范围内的已放大的声音输出的增益在接近入射声音的截止声压级时单调且非不连续性降低。

15.根据权利要求14所述的开放式助听器，其特征在于，在所述频带范围内的已放大的声音输出的相位失真在接近入射声音的截止声压级时接近零，并且当入射声压级大体上超出截止声压级时变为零。

16.根据权利要求15所述的开放式助听器，其特征在于，在所述频带范围内的已放大的声音输出的相位失真在接近入射声音的截止声压级时单调且非不连续性接近零。

17.一种用于补偿人耳的听力损失的开放式助听器，所述开放式助听器包括：

扩音器，所述扩音器用于拾取人耳接收的入射声音，并且将所述入射声音转换成电音频信号，

耳塞，所述耳塞包括能够放入人耳道内的扬声器，所述扬声器响应于由所述扩音器拾取的入射声音而产生声音输出，所述耳塞的开放式构造允许入射声音直接传送到耳膜，所述入射声音与所述扬声器的声音输出在耳膜处结合，由所述入射声音和所述耳塞的所述扬声器的声音输出结合的结果使所述助听器的佩戴者感知声音，以及

相干门，所述相干门处理所述扩音器的电音频信号，其中，所述相干门具有滤波器和用于所述滤波器的增益控制函数，其中，所述相干门至少部分地确定所述耳塞的扬声器的声音输出的特征，所述相干门使结合了入射声音的所述声音输出具有以下特征：

i）所述声音输出在根据佩戴者的听力损失特性而设定的频带范围内被放大；

ii）在所述频带范围内的已放大的声音输出的增益依赖于入射声音的声压级；以及

iii）当所述入射声压级超出设定的截止声压级时，所述耳塞的扬声器大体上不产生声音输出，佩戴者感知的声音几乎全部是直接到达耳膜的入射声音产生的结果。

18. 根据权利要求 16 所述的开放式助听器，其特征在于，所述相干门在结合了入射声音的所述声音输出中产生以下附加特征：在声音输出被放大的状态和在所述输出换能器大体上不产生声音输出的状态之间进行转换时，这种转换处于动态控制，产生所希望的起音时间和释音时间。

19. 根据权利要求 16 所述的开放式助听器，其特征在于，所述相干门的等待时间小于 1 毫秒。

20. 一种为有听力损失的个体补偿听力损失的方法，所述方法包括：

确定个体的频率依赖性听力损失特征，包括可听响度阈值，高于所述可听响度阈值，所述个体具有大体上正常听力能力，

提供两条路径，将入射声音传到患有听力损失的个体的耳朵的耳膜，包括直接开放的耳朵受话路径和处理信号的路径，

所述处理信号的路径在个体的耳膜处送出声音输出，所述声音输出结合了通过所述直接开放的耳朵受话路径到达耳膜的入射声音，并且送出的声音输出具有以下特征：

i）所述声音输出在根据佩戴者的听力损失特性而设定的频带范围内被放大；

ii）在所述频带范围内的已放大的声音输出的增益依赖于所述入射声音的声压级；以及

iii）当所述入射声压级大约超过所述个体的可听阈值时，所述输出换能器大体上不产生声音输出，所述个体感知的声音几乎全部是通过所述直接开放的耳朵受话路径到达耳膜的入射声音产生的结果。

21. 根据权利要求 20 所述的方法，其特征在于，当在所述声音输出被放大的状态和在所述输出换能器大体上不产生声音输出的状态之间进行转换时，这种转换处于动态控制，产生所希望的起音时间和释音时间。

22. 根据权利要求 20 所述的方法，其特征在于，在设定的频带范围内的已放大的声音输出的增益从低入射声压级的最大增益降低至接近所述个体的可听阈值的入射声压级的最小增益。

23. 根据权利要求 20 所述的方法，其特征在于，随着低入射声压级的入射声压级增大，所述低入射声压级低于所述个体的可听阈值，在所述频带范围内的已放大的声音输出的增益大体上线性降低。

24.根据权利要求20所述的方法，其特征在于，在所述频带范围内的已放大的声音输出的增益在接近所述个体的可听阈值时迅速降低。

25.根据权利要求20所述的方法，其特征在于，在所述频带范围内的已放大的声音输出的增益在接近所述个体的可听阈值时单调且非不连续性降低。

26.根据权利要求20所述的方法，其特征在于，在所述频带范围内的已放大的声音输出的相位失真在接近所述个体的可听阈值时接近零，并且当入射声压级大体上超出所述个体的可听阈值时变为零。

27.根据权利要求20所述的方法，其特征在于，在所述频带范围内的已放大的声音输出的相位失真在接近所述个体的可听阈值时单调且非不连续性接近零。

具有声压级和频率依赖性增益的助听器

技术领域

[0001] 本发明大体涉及一种助听器，更具体地说，涉及一种允许入射声音直接到达耳膜的开放式装置。

背景技术

[0002] 助听器通常包括扩音器、信号处理器、以及输出换能器（有时称为“接收器”）。输出换能器被放入耳道内并且能够作为外壳的一部分，其使耳道部分地打开（即透声）或者将耳道完全密封。开放式装置一般比封闭式装置较受使用者欢迎，并且尽可能向患有轻度或中度听力损失的人推荐（开放式助听器在它们可提供的增益量方面具有固有的局限性，因此不适合用于那些患有严重听力损失的人）。

[0003] 开放式装置的一个优点是舒适，开放式设计的柔软尖端的刺激较硬壳的封闭式耳塞的小并且较易适应。被耵聍（耳垢）感染或阻塞的风险也较小。不需要定制耳模，这可大大减小安装时间，仅需对这种现成的助听器稍作修改便可使用。还可避免闭塞影响，其中，封闭的耳道形成共振腔，将由使用者产生的低频率声音（例如说话或咀嚼声）放大，导致使用者的嗓音听起来不自然并且有嗡嗡声。闭塞影响是使用者拒绝使用封闭式助听器的主要原因之一。

[0004] 开放式设计还允许在复杂的声环境中有更好的处理，因为开放式设计允许听到在没有被助听器放大的频率的入射声音。例如，适用于高频率听力损失（高于 1kHz）的助听器不需要放大低频率。值得尽可能维持入射声音，因为入射声音具有确定声源位置和消除背景噪音所需的感知提示。这种感知提示包括耳间计时差、耳间响度差，以及相位效应。

[0005] 尽管开放式助听器有多个优点，但是它们也有显著的缺点。一种缺点是入射声音和已放大的声音的结合会在耳膜处产生人为噪音和失真，所述已放大的声音的频率是被助听器放大的频率。这些人为噪音和失真通常会被使用者注意到并且引起不满意而导致许多使用者在短时间后停用他们的助听器。

[0006] 一种人为噪音是由助听器的等待时间而引起，即当扩音器感应到声音时和当该声音在助听器的输出换能器被转换成声波时之间的时间延迟。新式的数字助听器的等待时间为 3-7 毫秒；较旧式的模拟助听器的等待时间为 1-2 毫秒。当入射声音的声压级和已放大声音的声压级是差不多时，非零的等待时间导致梳状滤波，其是一种频谱失真形式。梳状滤波的特征在于在耳膜处的声压出现一系列规则地间隔开的谱峰和谱谷。对于较长的等待时间，第一谱谷处于较低频率，并因此会影响频谱的较大部分。较短的等待时间产生的梳状滤波范围较小。人耳对这种人为噪音非常敏感，等待时间少于 8 毫秒被感知为音调配置（tone coloration），而较长的等待时间被感知为回声、敲打声、或是音调配置，这取决于延迟的声音的相对响度。

[0007] 另一种重组人为噪音产生于已放大的声音的相位失真。这种相位失真也产生了谱谷和谱峰的结构；每当频率相差 180 度相位时，它们合并进行相消性干涉，并产生谱谷，那些同相的频率会相长地叠加而产生谱峰。由于相位失真通常不均匀地散布在频谱上，这种

人为噪音可能较由等待时间引起的人为噪音更不规律。相位失真的来源可以是信号路径中的任何部件 :扩音器、信号处理部件或输出换能器(扬声器)。

[0008] 上述人为噪音的结果使未经训练的收听者也易于感知已感知声音的频谱失真。除了这些频谱失真,当已放大的信号比入射信号响得多时,助听器还使相位信息失真。人们相信这种相位失真本身是可以被察觉到的。近期有证据提出相位用途广泛,包括用于声源定位、语音编码和检测调相。

[0009] 本发明解决了与传统开放式助听器相关的缺点。本发明基本上减轻了在开放式助听器存在的人为噪音和失真的问题,并且基本上消除了使用者对于这种助听器设计的不满。本发明允许使用者可以享受开放式设计众所周知的优点,而不必忍受通常由这种设计带来的感知干扰。

发明内容

[0010] 本发明涉及一种开放式助听器,所述开放式助听器包括诸如扩音器的输入装置,所述输入装置用于拾取人耳接收的入射声音,并且将该入射声音转换成电音频信号 ;以及输出装置,所述输出装置具有能够放入人耳道内的输出换能器,所述输出换能器响应于由所述输入装置拾取的入射声音而产生声音输出。所述输出装置(可以是具有扬声器的听筒或耳塞的形式)是透声的,允许入射声音直接传送到耳膜,所述入射声音与输出换能器的声音输出在耳膜处结合。由所述入射声音与位于耳内的输出装置的声音输出结合的结果使助听器的佩戴者听到感知声音。

[0011] 本发明还包括处理输入装置的电音频信号的信号处理装置,以便能够以期望的方式来驱动输出装置的输出换能器。信号处理装置具有可变增益滤波器(本文有时也称之为“相干门(coherent gate)”),所述可变增益滤波器使输出换能器的已放大的声音输出具有以下特征 :

[0012] i) 声音输出在根据佩戴者的听力损失特性而设定的频带范围内被放大 ;

[0013] ii) 在所述频带范围内的已放大的声音输出的增益依赖于入射声音的响度,即声压级 ;以及

[0014] iii) 当入射声压级超出预设的声压级时,输出换能器不产生可感知的声音输出,佩戴者感知的声音几乎全部是入射声音产生的结果。

[0015] 在本发明的另一方面,信号处理装置产生输出换能器的声音输出,所述声音输出的特征是 :在设定的频带范围内的已放大的声音输出的增益从低入射声压级的最大增益降低至接近设定的入射声音的截止声压级的入射声压级的最小增益。

[0016] 在本发明的又一方面,用于拾取人耳接收的入射声音的输入装置将该入射声音转换成数字音频信号,而信号处理装置包括数字信号处理器。

[0017] 在本发明的再一方面,随着入射声压级的低入射声压级增大,所述入射声压级低于设定的入射声音的截止声压级,在所述频带范围内的已放大的声音输出的增益大体上线性降低。

[0018] 在本发明的进一方面,在所述频带范围内的已放大的声音输出的增益在接近入射声音的截止声压级时迅速降低,并且在入射声音的截止声压级降低至低于 0dB。

[0019] 仍然在本发明的另一方面,在所述频带范围内的已放大的声音输出的增益在接近

入射声音的截止声压级时单调且非不连续性降低。

[0020] 仍然在本发明的又一方面，在所述频带范围内的已放大的声音输出的相位失真在接近入射声音的截止声压级时接近零，当入射声压级大体上超出截止声压级时变为零，并且在接近入射声音的截止声压级时单调且非不连续性接近零。

[0021] 在本发明的再一方面，信号处理装置在结合了入射声音的声音输出中产生以下附加特征：当在声音输出被放大的状态和在输出换能器大体上不产生声音输出的状态之间进行转换时，这种转换处于动态控制，产生所希望的起音时间和释音时间。

[0022] 本发明还涉及一种为有听力损失的个体补偿听力损失的方法。所述方法一般首先包括确定个体的频率依赖性听力损失的特征，包括可听响度阈值，高于所述可听响度阈值，所述个体具有大体上正常听力能力。还提供两条路径，将入射声音传到患有听力损失的个体的耳朵的耳膜，包括直接开放的耳朵受话路径和处理信号的路径。所述处理信号的路径在个体的耳膜处送出声音输出，该声音输出结合了通过所述直接开放的耳朵受话路径到达耳膜的入射声音，更具体地说，在耳膜处送出的声音输出具有以下特征 ：

[0023] i) 声音输出在根据佩戴者的听力损失特性而设定的频带范围内被放大 ；

[0024] ii) 在所述频带范围内的已放大的声音输出的增益依赖于入射声音的声压级 ；以及

[0025] iii) 当所述入射声压级大约超过个体的可听阈值时，所述输出换能器大体上不产生声音输出，所述个体感知的声音几乎全部是通过所述直接开放的耳朵受话路径到达耳膜的入射声音产生的结果。

[0026] 本发明提供了许多益处。通过递减使用者的可听阈值内的已放大的声音，助听器的输出换能器不需要提供高的输出声压级，并因此能够在没有削波或限幅器的风险下使用。限幅器和削波都会在已放大信号中产生谐波失真，而限幅器设计成避免由削波引起的更多极端的人为噪音，削波由隔膜的非线性模式的激励产生。

[0027] 此外，本发明增加使用者的听觉空间提示 (spatial cues) 的数量和质量。这种提示是由完整的头相关变换函数 (head-related transfer function) 引起的，该头相关变换函数由外耳解剖 (耳廓和耳壳)、耳道、和由头引起的双取效应 (binaural effects)(例如两耳响度差、时差、和相位差) 形成。每当频率被放大，必然于该频率引起等待时间和相位失真，自然提示会被干扰。根据本发明，具体地说本发明的相干门，通过入射声音的明智的放大来维持自然提示。

[0028] 更广义地说，本发明在改善由使用者感知的声音的质量的同时保留了自然提示，使用者使用本发明的助听器最轻松。在复杂的听觉环境中，大脑可以使用多个提示来将声源分开并指引听觉注意力。在许多情况下，这种提示的损失会导致理解力或可懂度减小。然而，近期研究已经表明，失去某些提示还可能增加保持相同表现所需的认知努力。最简便是通过给受试者完成除基本听觉任务以外的第二个非听觉任务来说明。由于听力损失、被降级的输入质量、或者提高认知负荷的其它因素，第二个任务的表现会大幅降低，并且患者会较正常情况更快感到疲劳。

[0029] 通过以下叙述和权利要求书，本发明的其它方面和优点将变得显而易见。

附图说明

[0030] 图 1 是具有根据本发明的相干门的助听器的功能方块图。

[0031] 图 2 是根据本发明的助听器的物理组件、以及置于耳道内和外的组件的图示。

[0032] 图 3 是已放大的声音和入射声音通过佩戴了根据本发明的助听器的佩戴者的耳道传到该佩戴者的耳膜所沿的路径的示意图 / 图示 ;图中示出了入射声音和已放大的声音是如何叠加以及当它们到达耳膜时如何产生感知声音。

[0033] 图 4 是由根据本发明的助听器在使用者的耳膜处产生的声音声压级作为频率的函数以及处于不同的指示扁平 (白噪音) 输入声压级的曲线图。图中示出了随着扁平输入声压级上升,在与使用者相配的自定义的频带范围内的助听器的增益会降低。

[0034] 图 5A 是图 4 所示的例子处于 4kHz 的峰值频率时的输入 / 输出曲线,图中示出了在耳膜处以 dB 声压级测量的感知的声音、已放大的声音和入射声音。

[0035] 图 5B 是图 4 所示的例子处于 4kHz 的峰值频率时的增益函数。

[0036] 图 6 是示出输入声音的入射声音和由根据本发明的助听器放大的声音,以及它们在耳膜处的叠加的曲线图,其中输入声音的声压级略低于交叉点。

[0037] 图 7 是相位作为处于图 5 所示的增益声压级的频率的函数的曲线图。

[0038] 图 8 是示出根据本发明的总体方法的流程图。

具体实施方式

[0039] 参照附图,图 1 示出了形成根据本发明的助听器的实施例的方块图,助听器整体以数字 10 表示,其中输入 (入射) 声音由扩音器 11 换能、并且由模拟 - 数字转换器 13 数字化以进行数字处理 (应当理解的是本发明并不限于数字处理,也可以改为用模拟部件执行)。所述信号然后通过具有相干门 15 的信号处理电路,所述相干门 15 由滤波器 17、提供可变增益的增益控制函数 19、以及优选地由方块 18 表示的动态控制函数组成,所述动态控制函数 18 将在稍后进行叙述。滤波器的参数 (形状、带宽、增益结构等) 通过所述相干门内的设定函数来设定,所述设定函数由设定方块 20 表示。当在设定模式或编程模式时,可以根据使用者具体的听力损失来设定所述相干门的参数 (包括频率和增益等等)。这些设定函数可以由计算机控制。

[0040] 如增益控制方块 19 所表示,由助听器提供的增益可以反馈配置的方式从处于门输出 21 的所述相干门的输出信号确定,并且可以用于修改由反馈箭头 23 表示的滤波器的振幅。然后可以通过数字 - 模拟转换器 25 将所述输出信号转换成模拟信号、所述输出信号由放大器 27 放大,并且被传到输出换能器 (扬声器)29。应当意识到的是,也可以用上述方法以外的其它方法来执行增益控制,例如,可以采用前馈信号。

[0041] 最合适地,输入换能器 (扩音器) 和输出换能器 (扬声器) 在不产生增加频谱或相位失真的情况下精确地重现音频信号。这需要线性换能器具有平相响应并且在高至所需增益的最高声压级时都没有谐波失真。由于本发明适用于轻度至中度听力损失,本发明的助听器很少需要提供超过 80dB 声压级的声压级。

[0042] 图 2 示出了根据本发明的助听器的物理实现。扩音器 11 通过电线 31 连接到包含相干门 15 和输出换能器 29 的电子软件包。处理电子器件 (包括相干门) 和扩音器以独立的部件示出,可以预期的是它们能够一起容纳在单个可配戴单元内。诸如电池 33 的电源可以同样地与处理电子器件容纳在一起、或者可以分开放置并由电线连接至电路。耳塞式声

输出装置包括输出换能器 29 和透声耳塞 37。该换能器适当地嵌入在耳塞内。耳塞容纳在耳道 35 的外部分中，这意味着入射声音非略微递减并仍能够到达耳膜 39。这种耳塞被称为"开放式"设计，这与完全封闭耳道并递减入射声音的硬耳塞不同（按上下文需要，本文的术语"耳塞"应当被理解为包括换能器 29）。

[0043] 如图 3 中示意性地示出了这种开放式耳塞允许入射声音到达耳膜。当装置已耗损或处于不活动的状态时，在耳膜 39 处感知的声音（由输出箭头 41 表示）即是入射声音（由输入箭头 43 表示）。当装置处于活动状态时会产生声音，在本文有时称为"已放大的声音"。所述已放大的声音（由箭头 45 表示）和入射声音 43 均刺激耳膜；因此由大脑感知的声音是入射声音和已放大的声音的叠加。

[0044] 如上所述，已放大的声音的频谱由相干门 15 的滤波器 17 的参数确定，能够通过计算机经由相干门的设定函数 20 控制相干门 15（可以提供计算机界面以通过编程来确定相干门的滤波器形状）。可将滤波器视作为均衡曲线，分别向频率的窄带提供增益。滤波器的形状可以高度自定义，并能够适应于各种轻度或中度的听力损失，尽管最终滤波器的形状是由相干门算法的设计限制。例如，滤波器在横跨所有频率时可以是扁平的、在特定频率（高通、低通、或带通）升压、或双模态（在两个频率出现峰值）。

[0045] 可以首先通过设置频率依赖性增益（均衡）曲线建立信号处理电路的相干门 15 的特性，因此"滤波器"根据使用者的具体测得的听力损失进行调整。因此，已建立的滤波器优选地是最小相位滤波器，即仅于那些已放大的频率改变该滤波器的相位。随着一个频带的输入声压级（入射声音）增大，滤波器增益被逐渐递减直到入射声音占优势。可以以这种方式递减增益，以致于相位响应也逐渐降低至零。补偿特定个体的听力损失所需的这种精确的滤波器特性可被称为"拟合算法"。

[0046] 可以通过测试使用者的听力来确定使用者具体的听力损失的拟合算法。拟合算法能够为相干门（滤波器）电路提供自定义的增益控制：该拟合算法仅当频率低于使用者的可听阈值时放大给定的频率。当放大柔和的声音时，滤波器的相位延迟是可被使用者接受的，并且大大改善低声压级的讲话和音乐的可听度。然而，一旦输入信号到达使用者的阈值，滤波器的作用被移除，优选地迅速被移除，这也移除了失真（如果滤波器在上述可听阈值之上保持有效，所产生的声音在使用者听起来是失真的且令人不愉快的：依赖于听力损失的类别，感知可以是响亮的或是有嗡嗡声的）。

[0047] 相干门的其它特性是每一个滤波器的动态属性。这些动态属性包括起音时间和释音时间，是当入射声音的响度上升到高于使用者的可听阈值以上滤波器完全启动所需的时间，以及是当入射声音的响度下降到低于该阈值滤波器完全失效所需的时间。通过采用动态控制（图 1 中的方块 18 表示）可以适当地设定起音时间和释音时间，以致于突然响亮的事件不会被放大，这需要快的起音时间，并使得在响亮的事件之后的柔和声音仍然能被听到，这需要适度地快的释音时间。如何任一个参数太长或太短，会出现音调配置和明显的声压级波动；如果释音时间太短，将会注意到跳动的人为噪音。动态的值可能依赖于使用者具体的听力损失以及在拟合过程中使用者的主观反馈。通常，滤波器的起音时间适当地设定在大约 15 微秒至大约 10 毫秒之间，并且优选地小于大约 1 毫秒。滤波器释音时间优选地在大约 200 微秒至 30 毫秒的范围内。这些动态由制造商或受训练的专业人士设定最为适宜。

[0048] 尽管上述助听器是用于一只耳朵的单通道装置，应当理解的是，可以通过将两个这样的装置适当组合来用于双耳。在这种情况下，该组合装置可以共用装有电子器件和电池的物理附件，但是每只耳朵需要有属于自己的耳塞，并且优选地每只耳朵都具有属于自己的专用扩音器和相干门。建议采用分开的扩音器以维持双耳声音差异，双耳声音差异在每只耳朵是不同的。优选地独立设定每只耳朵的相干门，因为每只耳朵的听力损失通常都是不同的（称为非对称听力损失）。优选地，越靠近耳朵佩戴扩音器越好。

[0049] 现参照图 4 所示的示例性滤波器形状，其是峰值频率 (F_peak) 为 4kHz 的带通滤波器。这种滤波器对应于处于 4kHz 的 20dB 的典型噪音型听力损失。根据本发明，相干门可以实现任何滤波器形状。首先，必须考虑到达耳膜的声音以及在耳膜处叠加的声音。图 3 示出通过耳塞直接到达耳膜的入射声音和已放大的声音的组合或叠加。在图 4 所示的滤波器例子中，助听器仅增强大约 4kHz 的频率。对于 0dB 声压级的扁平输入信号，在耳膜处将在 4kHz 的叠加声音增强到 20dB 声压级，使这些频率现能够被使用者听到（已经可以被听到频谱的其它部分没有被放大）。如图所示，当在 10dB 和 20dB 的较高输入声压级时，可以看到增强的程度在逐渐减小。一旦输入声压级到达 23dB，即假设佩戴者所选定的截止声压级或可听阈值，助听器基本上在 4kHz 时不提供已放大的声音（增益小于 -20dB）。在这个阈值之上，在听力损失频率范围内的入射声音将被佩戴者感知而不需要补偿。在这个例子中，产生用于处理的输入信号的入射声音首先被认为是静态的，该入射声音的响度音量和波峰因数不随时间而变化。

[0050] 图 5A 示出了入射声音的声压级（由虚线 49 表示）、已放大的声音（由虚线 51 表示）、以及由这两种声音在耳膜处叠加所产生的声音的声压级（由实线 47 表示）作为输入（入射）声音声压级的函数 ；图 5B 示出了滤波器增益（由虚线 50 表示）随着输入声音声压级的变化如何改变。随着输入声音声压级改变，滤波器的增益参数也被改变，导致在耳膜处的声音声压级也改变。在低输入声音声压级（大约低于 10dB)，可以看到到达耳膜的声音和最终被感知的声音由已放大的声音主导。在该低输入区域，可以看到滤波器的增益几乎线性降低。在该低输入区域上面是“交叉区域”（在图 5B 中由数字 55 表示），其中已放大的声音 51 和入射声音 49 之间的差小于 8dB。在这个交叉区域内的声压级中，入射声音和已放大的声音都显著地对到达耳膜处的声音和被感知的声音作出贡献。因此，从该交叉区域内可以按所需偏离增益函数的线性（交叉区域内的偏离可在图 5B 看到）。尽管如此，为了防止在交叉区域中感知人为噪音，必须逐步改变增益函数 ；即它必须单调且非不连续性降低，并且是平滑的（在数学的意义上，具有连续的导数）。有效地，这种定义明确的增益函数将类似的输入声压级映射到类似的输出声压级 ；输入声压级的细小变化会引起输出声压级的细小变化。如图 5B 所示，虽然最佳的增益函数是非线性的，应当注意的是，也可以使用有效且易于实现的线性增益函数。

[0051] 图 6 示出了入射声音（由线 57 表示）和已放大的声音（由线 59 表示）以及它们在耳膜处的叠加（由线 61 表示），其中，入射声音的声压级在交叉区域内（输入声压级在 16dB)。在交叉区域中，已放大的声音的相位特征和延迟特征尤其重要。随着滤波器增益降低（正如增益函数的变化），频率依赖性相位必须逐步接近零。因为具有已放大的声压级，如果输入声压级细微变化而相位急剧地变化，这将会被装置的佩戴者注意到。

[0052] 图 7 示出了一种避免在输入声压级的细小变化出现不可接受的大且可感知的相

位变化的方法，图中示出了随着输入声压级升高和系统增益降低，相位扰动缓慢降低到零。图 7 示出了相位作为处于图 5B 所示的增益声压级的频率的函数。当增益降低至低于零时相位扰动也会降低。例如，处于 +20dB 增益的相位扰动（由线 61 表示）比处于 0dB 的相位扰动（由线 63 表示）大。在 -10dB 时实际上没有相位扰动。尽管提供频率依赖性增益的滤波器必定引起相移，可以通过选择合适的滤波器装置（即最小相位滤波器）使这种相移最小化。

[0053] 助听器的其它重要参数是等待时间，即入射声音到达扩音器和已放大的声音在扩音器处输出之间的时间。这种延迟需要尽可能小，理想是小于 1 毫秒。延迟超过大约 5 毫秒产生配置的人为噪音，而延迟超过 1 毫秒影响声音定位提示。因此，优选地，由图 1 所示的信号处理电路的相干门 15 引起的等待时间小于 1 毫秒。

[0054] 为了实现上述处理方案的优点，输入换能器（扩音器）和输出换能器（扬声器）必须能够以最大的逼真度再现音频信号。除非输入和输出换能器是线性的，否则不会实现相干门的等相响应；即是，除非它们在最响亮的预期输出声压级处具有扁平相位响应以及低谐波失真（优选地小于 1%）。

[0055] 图 8 示出本发明的上述实施例的总体方法，其中由方块 101 表示的入射声音能够通过两条路径到达佩戴者的耳膜，分别由箭头 A 和箭头 B 表示，其中入射声音在方块 103 进行叠加。如图 8 所示，第一路径（路径 A）是由助听器的耳塞的开放式构造实现的传到耳膜的直接开放的耳朵受话路径。入射声音总是通过这条路径到达耳膜。另一条路径（路径 B）是处理信号的路径，所述处理信号的路径将依赖于频率和入射声音声压级的已放大的声音提供到耳膜。通过这个路径，被转换成电音频信号的入射声音由可变增益选通功能器件处理（如图 1 所示的相干门 15），其中通过路径 A 到达耳膜处的入射声音由通过路径 B 到达的已放大的声音增强。从路径 B 到达的声音的声压级不仅依赖于补偿听力损失的频带，还依赖于在任何时间点的入射声音的声压级。用于放大由该路径处理的音频信号的可变增益功能器件的特征根据测得的佩戴者的听力损失而定制，包括佩戴者的可听阈值。

[0056] 更具体地说，在处理信号的路径 B 中，入射声音经由扩音器 105 进入该路径，扩音器将声音转换成能够由模拟电路或更优选地由数字信号处理的电音频信号。处理步骤包括首先确定在频带或感兴趣的频带范围内（方块 107）的入射声音的响度。如果扩音器拾取的入射声音的响度低于测得的佩戴者的可听阈值（方块 109），增益必须补偿测得的佩戴者的听力损失，即将低于阈值的声音提升到佩戴者的可听声压级，该增益由例如增益计算（方块 111）确定。基于这已确定的增益，启动相干门的滤波器（方块 113）以允许通过路径 B 的音频信号被放大至由增益确定的声压级。如较早前所述，滤波器的启动可以处于动态控制，使起音时间可设定成所希望的声压级。所产生的已放大声音被用作驱动耳塞的扬声器 115。扬声器的输出产生已放大的声音，该已放大的声音在耳膜处与入射声音叠加。

[0057] 另一方面，如果由扩音器拾取的入射声音的响度在上述测得的佩戴者的可听阈值以上（返回方块 109）时，相干门的过滤器失效（方块 117），从而将可驱动扬声器 115 的任何音频信号移除。与滤波器的启动一样，滤波器的失效可以处于动态控制，其中释音时间能够以上述的方式设定。在释音过程中，已放大的声音会在非常短的时间内继续驱动扬声器 115。

[0058] 虽然本发明在上述说明书中已详细地叙述，但是说明书的细节并不旨在限制本发

明，本发明仅受以下权利要求书的限制。

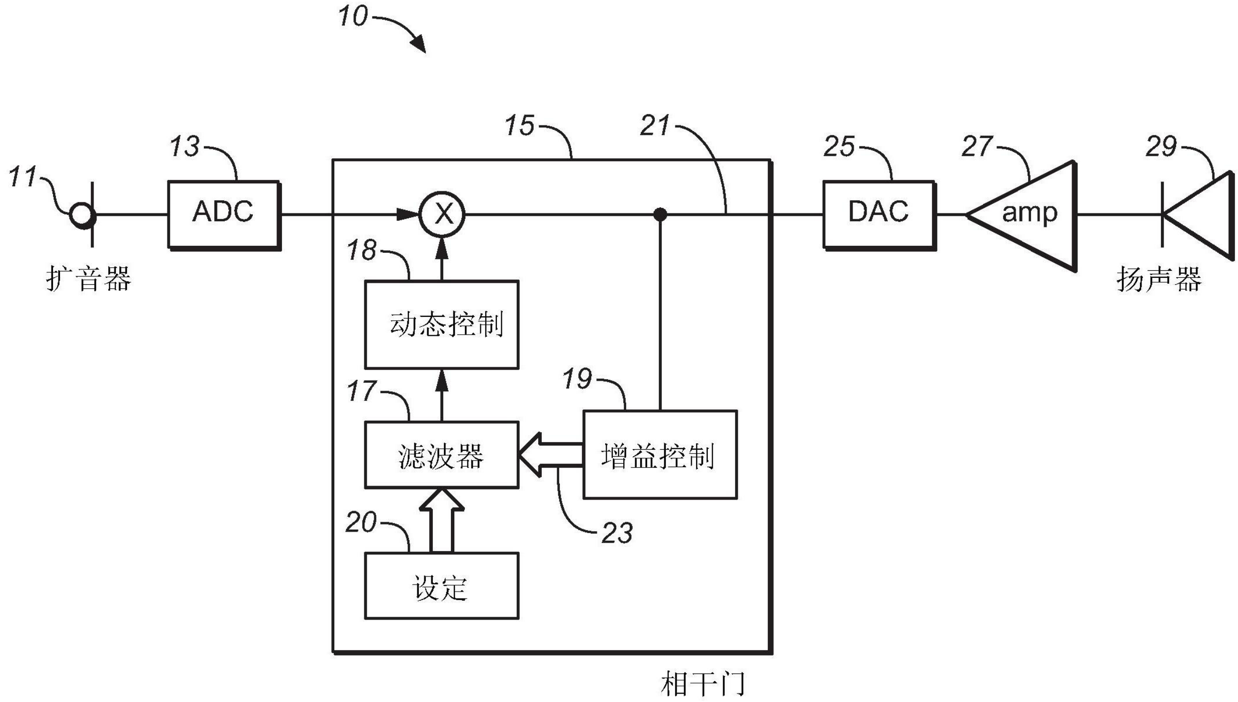

10
11
13
15
21
25
27
29
ADC
X
DAC
amp
扩音器
扬声器
18
动态控制
17
19
滤波器
增益控制
20
23
设定
相干门

图 1

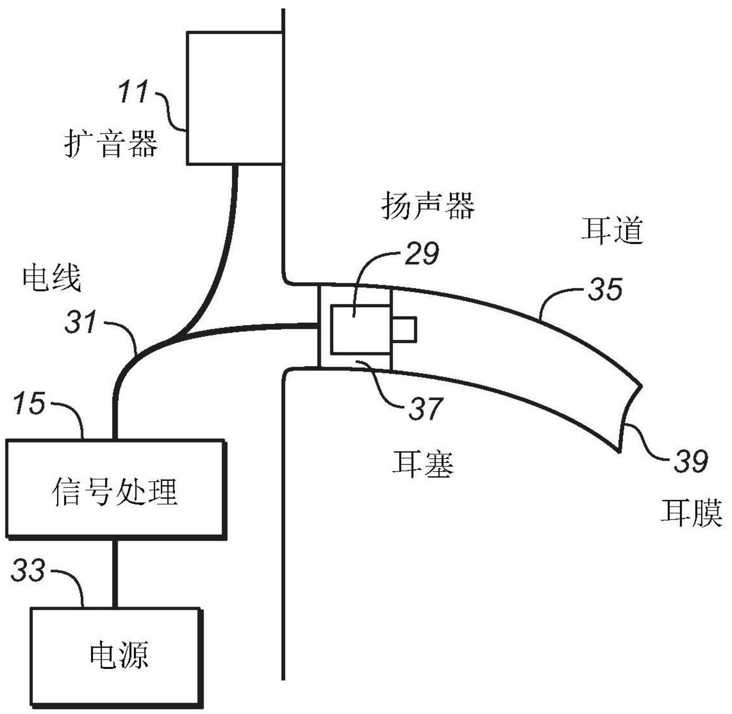

11
扩音器
扬声器
耳道
29
35
电线
31
15
37
39
信号处理
耳塞
耳膜
33
电源

图 2

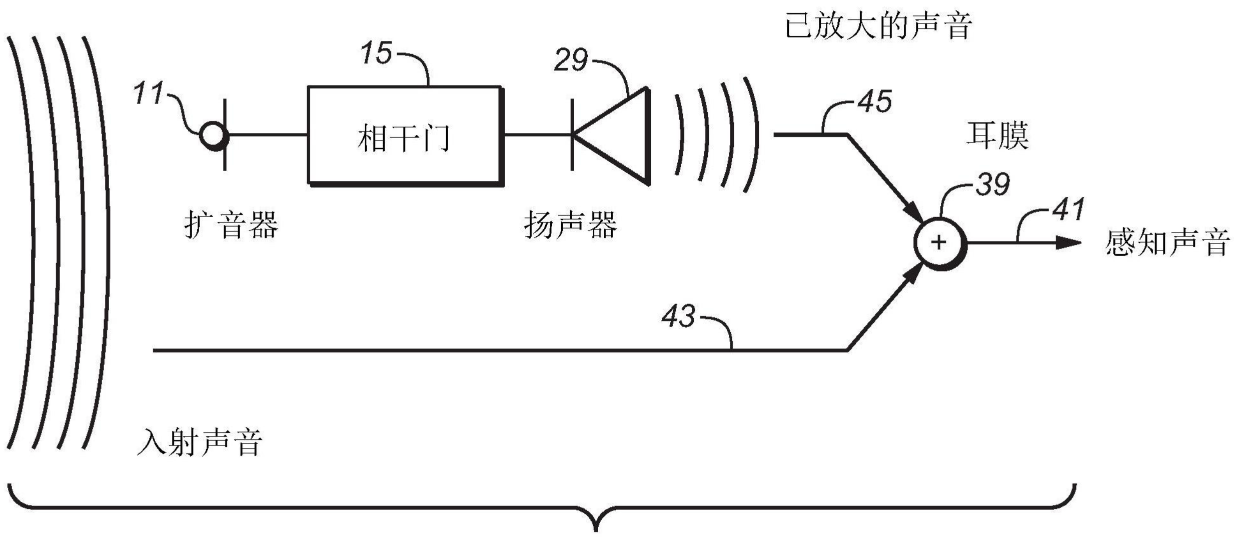

已放大的声音
15
29
11
相干门
45
耳膜
39
扩音器
扬声器
41
感知声音
43
入射声音

图 3

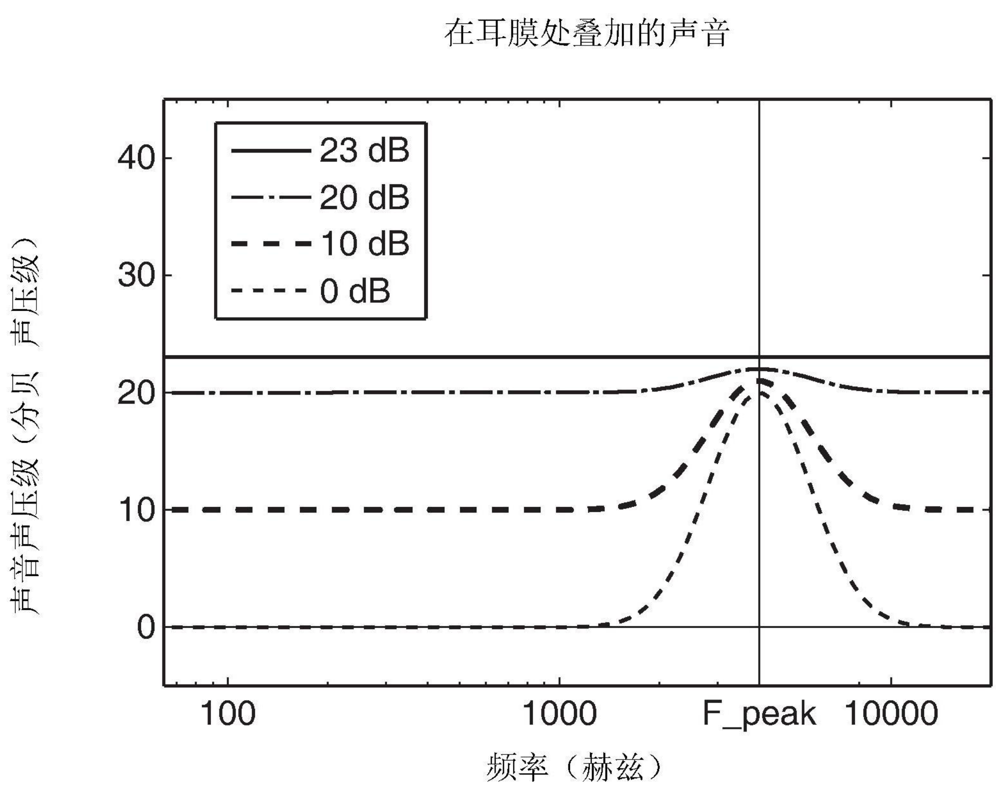

在耳膜处叠加的声音
23 dB
20 dB
10 dB
0 dB
40
30
20
10
0
声音声压级（分贝 声压级）
100
1000
F_peak
10000
频率（赫兹）

图 4

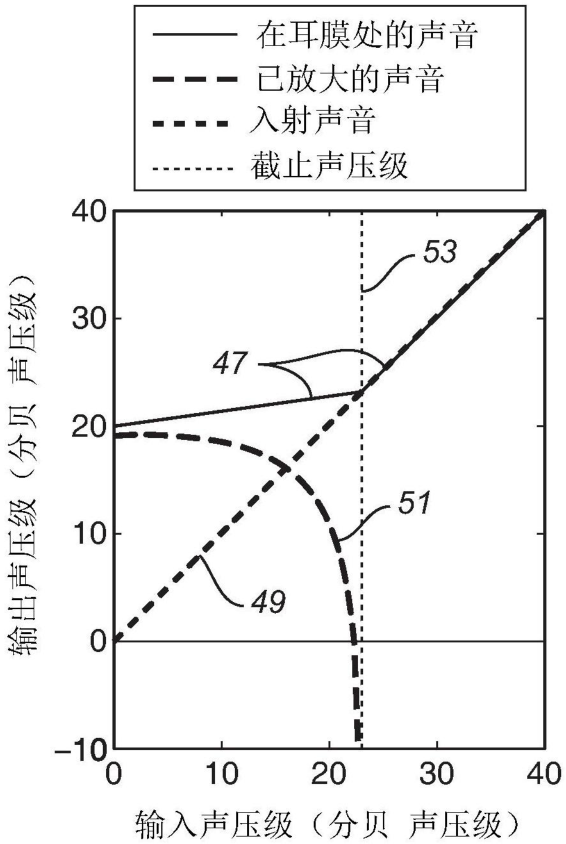

在耳膜处的声音
已放大的声音
入射声音
截止声压级
40
30
20
10
0
−10
0
10
20
30
40
53
47
51
49
输出声压级（分贝 声压级）
输入声压级（分贝 声压级）

图 5A

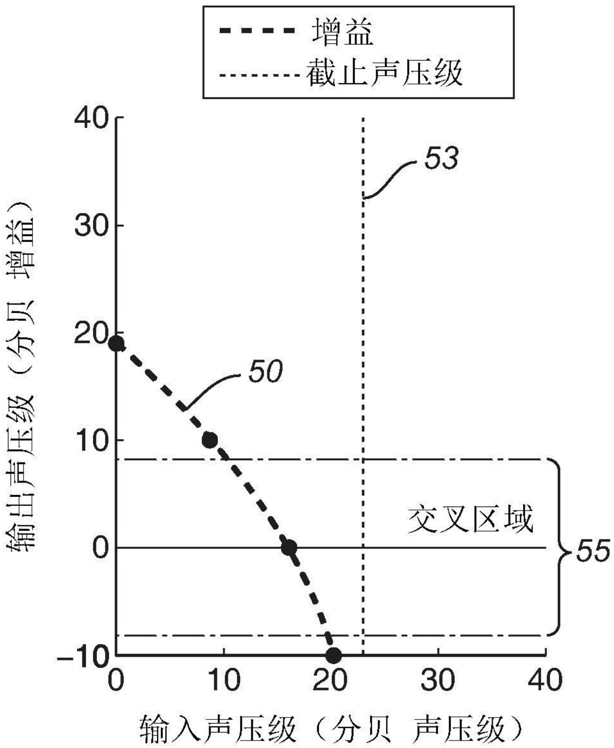

增益
截止声压级
40
30
20
10
0
−10
0
10
20
30
40
53
50
交叉区域
55
输出声压级（分贝 增益）
输入声压级（分贝 声压级）

图 5B

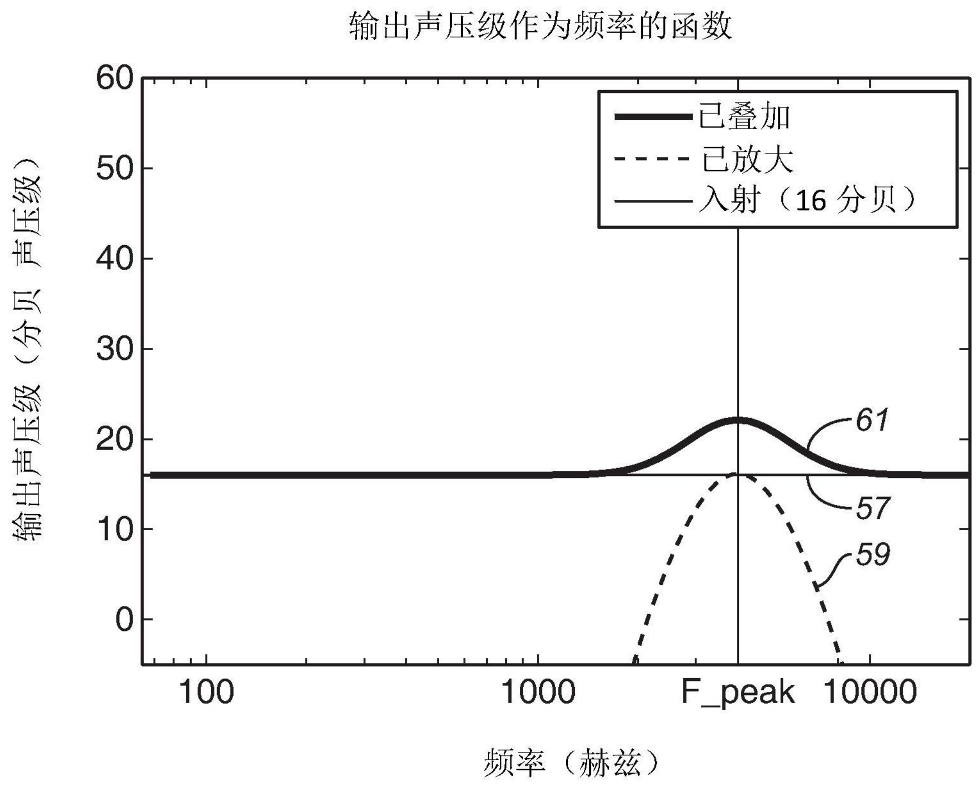

输出声压级作为频率的函数
60
50
40
30
20
10
0
已叠加
已放大
入射（16 分贝）
61
57
59
输出声压级（分贝 声压级）
100
1000
F_peak
10000
频率（赫兹）

图 6

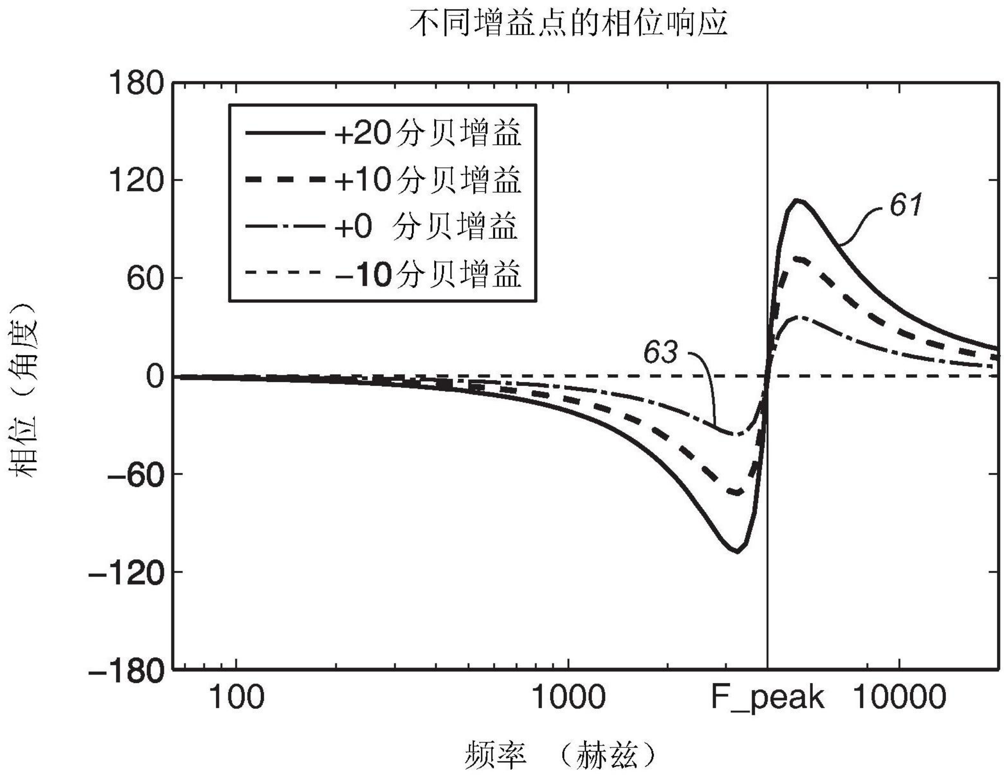

不同增益点的相位响应
180
120
60
0
−60
−120
−180
+20分贝增益
+10分贝增益
+0 分贝增益
−10分贝增益
61
63
相位（角度）
100
1000
F_peak
10000
频率 （赫兹）

图 7

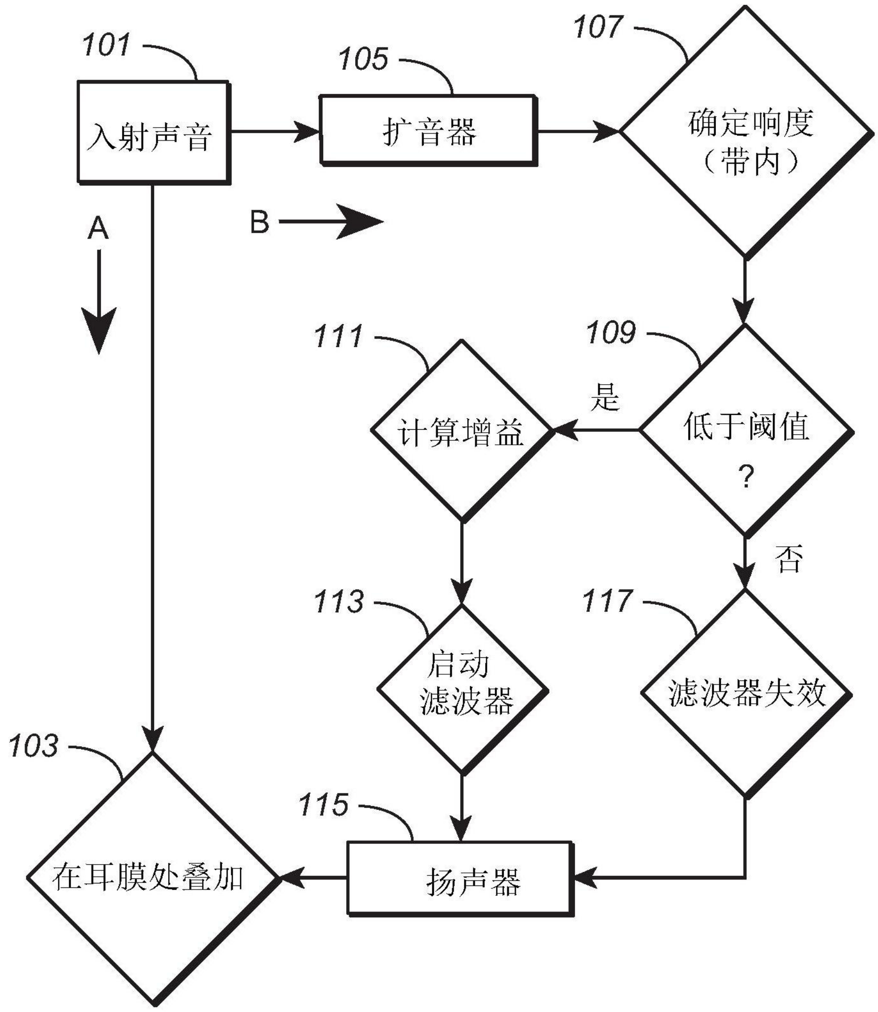

101
入射声音
105
扩音器
107
确定响度
（带内）
A
B
111
计算增益
是
109
低于阈值
？
否
113
启动
滤波器
117
滤波器失效
103
在耳膜处叠加
115
扬声器

图 8