

FITTING METHOD AND A HEARING AID FOR SUPPRESSION OF PERCEIVED OCCLUSION

The present invention relates to a fitting method for a hearing aid. The invention further relates to a hearing aid with means for suppression of perceived occlusion.

- 5 The occlusion effect denotes the low frequency enhancement in the perceived loudness level due to occlusion of the ear canal. Some hearing aid users having an earmold or a hearing aid in the ear canal, blocking the canal, complain that they have a perception of the sound environment as that of being in a barrel. In particular, their own voice sounds as if they speak in a barrel.
- 10 Sounds produced in a person's throat propagate to the person's ear canal by air as well as by bone conduction. Sound waves from speech propagated through air from the mouth and around the head to the ear are denoted air conducted speech. Sound waves from speech propagated through the elastic cartilaginous tissue in the ear canal and transformed there into acoustic waves in the ear canal are denoted bone conducted speech.
- 15 It has been suggested to suppress the occlusion effect by inserting the hearing aid earmold or housing deeply in the ear canal, i.e. in the bony part of the ear canal. This reduces the occlusion effect since the sealed volume of the ear canal is isolated from the cartilaginous tissue transforming bone conducted speech to acoustic waves. However, the bony part of the ear canal is typically very sensitive, and positioning of a structural member in this part of the ear canal is,
- 20 in many cases, not comfortable to the user.

It has also been suggested to provide a vent in the earmold or hearing aid housing, allowing bone conducted sound to escape from the ear canal inner portion. The vent is typically a tube extending through the earmold or hearing aid housing, facilitating transmission of acoustic waves from one side to the other so that the ear canal is not completely blocked. However,

25 the vent may give rise to other complications. One problem related to the application of a vent is the inherent sound energy loss to the environment through the vent. This is a drawback, due to the increased power consumption needed to compensate for this energy loss.

Another complication is the tendency of the vent to give rise to acoustic feedback. Acoustic feedback occurs when the microphone of a hearing aid receives the acoustic output signal

30 generated by the receiver (hearing aid parlance for the output transducer, which is a miniature loud-speaker). Amplification in the hearing aid of the signal received by the microphone may

lead to generation of a stronger acoustic output signal, and eventually the hearing aid may oscillate. In hearing aids residing completely in the canal (CIC hearing aids), the short distance between microphone and receiver leads to low attenuation of acoustic waves transmitted from the receiver to the microphone. Generally, the attenuation increases with decreasing vent diameter and increasing vent length. However, prevention of the occlusion imposes the opposite requirements on the vent geometry.

It has been suggested to counter occlusion effect by generally decreasing the gain in the low-frequency bands. Such a solution is, however, not optimal in respect of sound quality as it influences the sound in all environments, including those environments, where there is no occlusion.

WO9934642-A1 provides a multichannel hearing aid with compressors in respective frequency channels. A compressor is suitable for the purpose of compensating a hearing impairment known as recruitment.

It is an object of one embodiment of the invention to provide a hearing aid with signal processing means for suppression of the perceived occlusion effect.

It is a further object of another embodiment of the invention to provide a fitting method leading to a suppression of a hearing aid user's perception of the occlusion effect.

The inventors have discovered that in some listening situations, hearing aid users perceive an improvement in sound quality when low frequency bands are enhanced. However, in some other listening situations, e.g. during conversation, hearing aid users may perceive an improvement in sound quality when low frequency bands are attenuated, probably for reason of avoiding superimposing an amplified acoustic signal on top of bone conducted speech, i.e. the above-mentioned "barrel perception" becomes less noticeable.

In a first aspect, the invention provides a fitting method for a hearing aid with at least one low frequency channel having an individually adjustable compressor, comprising the steps of determining initial values of settings of the hearing aid, including the compression ratio of the compressor, that will compensate the hearing loss, and increasing the compression ratio over the initial value of the compression ratio in the at least one low frequency band.

In a second aspect, the invention provides a hearing aid having a microphone, a signal processor and an output transducer, wherein the signal processor comprises a filter bank having at least a low frequency band and a high frequency band with respective compressors

and with at least one offset amplifier in respect of the low frequency band compressor for increasing the compression.

In a third aspect, the invention provides a multichannel hearing aid comprising at least one input transducer for transforming an acoustic input signal into a first electrical signal, a filter bank with bandpass filters for dividing the first electrical signal into a set of bandpass filtered first electrical signals, a processor for generation of a second electrical signal by individual processing of each of the bandpass filtered first electrical signals and adding together the processed electrical signals into the second electrical signal, and an output transducer for transforming the second electrical signal into an acoustic output signal, wherein the processor comprises a set of compressors, each of which is connected to a respective bandpass filter for compression of the corresponding bandpass filtered signal, and wherein at least one of the compressors is a low frequency compressor adapted to compress low frequency, high level signals.

Accordingly, one embodiment of a hearing aid that has been fitted with the fitting method according to the invention is provided with a compressor in a low frequency channel that compresses signals with a larger compression ratio than would have been set according to known fitting methods.

The increased compression ratio according to an embodiment of the invention attenuates occlusion effects without generally attenuating low frequency signals, whereby the sound quality in situations with no occlusion effect is improved.

Thus, according to an embodiment of the invention, a fitting method is provided for a hearing aid with at least one low frequency channel having an individually adjustable compressor. The method may comprise the first step of adjusting the characteristic of the compressor according to the hearing loss to be compensated by the hearing aid. The method may further comprise the step of increasing the compression ratio of the characteristic of the compressor in the low frequency band.

According to another embodiment of the invention, a hearing aid is adjusted with a specific emphasis on attenuating a user's own speech at low frequencies. As the hearing aid processes air conducted sound signals, this will attenuate solely the air conducted part of the sound. However, this improves the user comfort as the sum of air conducted and bone conducted speech is attenuated. A suppression of the occlusion effect during conversation is hereby obtained since the sum of bone and air conducted speech has been reduced to a level that is

closer to the sum level in a nonoccluded ear canal. The user's own speech is discriminated from another person's speech by the signal level at low frequencies.

In another embodiment, a multichannel hearing aid comprises at least one input transducer for transforming an acoustic input signal into a first electrical signal, a filter bank with bandpass filters for dividing the first electrical signal into a set of bandpass filtered first electrical signals, a processor for generation of a second electrical signal by individual processing of each of the bandpass filtered first electrical signals, e.g. for amplification with different gains, and adding the processed electrical signals into the second electrical signal, and an output transducer for transforming the second electrical signal into an acoustic output signal. In this hearing aid the processor comprises a set of compressors, each of which is connected to a respective bandpass filter for compression of the corresponding bandpass filtered signal. The frequency ranges of the bandpass filters are also denoted channels.

In a simple embodiment of the invention, the hearing aid is a dual channel hearing aid, i.e. the hearing aid processes incoming signals in two frequency bands only. Thus, the first filter bank consists of a single bandpass filter, and the single bandpass filter may be constituted by the bandpass filter that is inherent in the electronic circuit, i.e. no special circuitry provides the bandpass filter. In this embodiment of the hearing aid, the single processed electrical signal is presented at the output of the processor.

It is presently preferred that the compression ratio is increased to at least 1.4, and more preferred to increase the compression ratio to approximately 2.

The low frequency channel may further comprise an offset amplifier adding an offset gain to the compressor characteristic, and the method may further comprise the step of adjusting the offset gain in the range from -20 dB to 20 dB.

It is a characteristic feature of a compressor characteristic having been adjusted in accordance with the fitting method according to one embodiment of the invention that the compression ratio, e.g. a compression ratio equal to 2, is maintained for a large range of the signal level at the input of the compressor. It is preferred that the signal level range starts at 30 dB SPL, more preferred at 25 dB SPL, still more preferred at 20 dB SPL, and even more preferred below 20 dB SPL. Preferably the range ends at 60 dB SPL, preferably at 70 dB SPL, more preferred at 80 dB SPL, and even more preferred above 80 dB SPL. The range may vary from one frequency band to another.

In accordance with another embodiment of the invention, it has been recognized that the perception of the occlusion effect is caused by signals at low frequencies, such as frequencies below 1600 Hz, more pronounced below 1000 Hz, even more pronounced below 800 Hz, still more pronounced below 500 Hz. Thus, according to this embodiment of the invention, a low frequency band comprises frequencies below 1600 Hz, preferably below 1000 Hz, more preferred below 800 Hz, and most preferred below 500 Hz.

Still other objects of the invention will become apparent to those skilled in the art from the following description wherein the invention will be explained in greater detail. By way of example, there is shown and described a preferred embodiment of this invention. As will be realized, the invention is capable of other different embodiments, and its several details are capable of modification in various, obvious aspects all without departing from the invention. Accordingly, the drawings and descriptions will be regarded as illustrative in nature and not as restrictive. In the drawing:

Fig. 1 shows plots of sound pressure level (SPL) in the occluded and nonoccluded ear canal, respectively, as a function of frequency for a specific sound,

Fig. 2 shows plots of SPL of speech generated by the person himself and of speech generated by another person, respectively, in a nonoccluded ear canal as a function of frequency, for a long-term speech spectrum,

Fig. 3 shows a prior art compressor characteristic,

Fig. 4 shows a compressor characteristic according to the invention,

Fig. 5 illustrates fine tuning of the compressor characteristic according to the invention, and

Fig. 6 shows a block diagram of a hearing aid according to the invention.

- 5 Fig. 1 shows plots of sound pressure level (SPL) in the ear canal as a function of frequency for a sound with a specific frequency spectrum. SPL is the quantity of sound energy relative to a reference pressure: $20 \mu\text{Pa}$. The plotted SPL is measured in two situations. Curve 1 shows SPL measured in the occluded ear canal, and curve 2 shows SPL measured in the non-occluded ear canal. By comparison of these curves, it may be seen that for low frequencies, the SPL is
- 10 approximately 10-30 dB higher for an occluded ear canal than for a nonoccluded ear canal. The plotted curves are adopted from "The hollow voice occlusion effect", M. Killion, Fig. 6, "Hearing aid fitting", J. Jensen, p.231, 13th Danavox Symposium, 1988.

The invention exploits the fact that the eardrum of a user receives the user's own speech from two different propagation paths. In addition to the bone conducted speech mentioned above,

15 sound waves also propagate through air from the mouth and around the head to the ear where it is received by the hearing aid. For air conducted speech, low frequencies are enhanced since the curved path of propagation through the air around the head attenuates high frequencies, favoring low frequencies.

This is further illustrated in Fig. 2 showing plots of SPL in a non-occluded ear canal as a

20 function of frequency for a sound with a specific frequency spectrum. Curve 3 is SPL of speech generated by the person himself or herself, while curve 4 is SPL of speech generated by another person. At low frequencies, there is a difference in SPL of approximately 10-15 dB between a person's own speech and the speech of another person.

From these two figures the following deductions can be made:

- 25 In Figure 2, there is a level separation between the two curves 3 and 4 on 10-15 dB in the low frequency range (0-1000 Hz.). The frequency distribution characteristic of the curve 3 of Figure 2 generally resembles that of curve 1 of Figure 1, in that there is an emphasis of low frequency signals.

There is a signal level separation at low frequencies between the users own speech and the speech level of a conversation partner of some 10 -15 dB. Accordingly, a reduction in amplification of high level sounds - and thus in the sound level of sounds generated by the user and propagated around the users head into the hearing aid for amplification - will reduce the level of sound amplified and propagated by the output transducer. Although this may not dampen the bone conducted speech, it will reduce the total sound level at the users eardrum, whereby the users perception of occlusion of the ear canal will be reduced.

This fact points to the solution according to the invention, by which the problem of perceived occlusion of the ear canal is solved by providing signal processing to effectively provide one amplification setting for low level signals and another, lower, amplification setting for higher level signals. A processing device with such capability is a compressor. A compressor has been suggested in the prior art for the purpose of compensating a hearing impairment known as recruitment. However, the prior art has not suggested the use of a compressor for defeating problems related to perceived occlusion.

Therefore, according to the invention the solution to the problem of subjectively perceived occlusion effect - in the conversation situation - is to decrease gain in the high-level part of the compressor characteristic.

Fig. 3 shows a plot of a prior art compressor characteristic, i.e. a plot of the compressor output level as a function of the input level, both in SPL. The characteristic comprises two segments 5, 6, that are interconnected at a knee-point 10, typically positioned at 50 dB SPL input level. Below the knee point 10, there is substantially no compression, i.e. the gain is a constant gain, as illustrated by the linear segment 5, suitable for compensating the hearing loss at low input signal levels. Above the knee point 10, there is a compression ratio exceeding unity, e.g. 1.2:1, as illustrated by the segment 6, suitable for compensating for recruitment. Recruitment denotes the effect of a sensorineural hearing loss that the perception of loudness rises steeply with increasing sound pressure just above the hearing threshold, while it rises normally at high sound pressures. The hearing threshold is the lowest sound level at which sound is perceived. The compression ratio of a segment is equal to the reciprocal value of the slope of the respective segment.

In Fig. 3, segment 5 has a compression ratio of 1, and segment 6 has a compression ratio in the range of approximately 1.0 to 1.4:1, e.g. 1.2:1. Generally, the gain and the compression characteristic will be adapted in order that signals at higher levels will not be compressed to levels below the input level, except for a safety capping at very high levels.

Fig. 4 shows a compressor characteristic of a compressor for use according to the invention. In accordance with the invention, signals at a high level, i.e. above signal levels of speech from another person, are compressed. In Fig. 4, the segments 5, 6 are identical to the segments 5, 6 shown in Fig. 3. Preferably, segment 6 has a compression ratio that is greater than 1.4, and, more preferred, a compression ratio substantially equal to 2. Other values of the compression ratio may be used if appropriate. It is the gist of the invention that compressors operating at low frequencies enhance low level signals and attenuate high level signals, whereby perception of occlusion is suppressed. The compression ratio is constant in a wide range of signal levels, in the present example from 20 dB SPL to 100 dB SPL.

- 10 Users bothered by perceived occlusion effects are typically users with a mild or moderate hearing loss in a low frequency range. For such patients, the generally favored fitting methods will prescribe a compressor characteristic with a low compression ratio, i.e. a compression ratio close to 1. This applies a high level of gain to high level signals whereby the above-mentioned perceived occlusion effect may arise. The invention, however, adds a further step to a known
- 15 fitting method, that leads to an increase of the compression ratio, e.g. to a compression ratio that is greater than 1.4, e.g. equal to 2, whereby low frequency, high level signals are attenuated, alleviating the perceived occlusion effect. Hereby high level signals, i.e. signal levels exceeding the signal level of normal speech produced by a conversation partner in a typical conversation situation, is compressed in the low frequency range, whereby a reduction
- 20 in the level of perceived occlusion of the ear canal is obtained, as explained above.

For the programming of a hearing aid the following steps need to be performed:

Measuring the hearing loss at a number of frequencies

Programming the compressors (fitting them) with parameters corresponding to the characteristic needed to compensate the measured hearing loss

- 25 *Possibly fine-tuning the compensation characteristic*

In the multichannel hearing aid the compressor characteristics will be adjusted in a channel-by-channel approach in accordance with the measured auditory characteristics, and possibly modified in accordance with the audiologists suggestions and the users preferences. In case of a simple compressor, this would comprise entering the value of the knee-point and the

30 compression ratio in every channel. Fine-tuning may be done by performing one or more of the following steps: adjusting the output level of the knee-point, adjusting the input level of the

knee-point, or adjusting the over-all gain of the compressor characteristic (i.e. changing gain at all signal levels with the same amount). Reference is made to WO99/34642-A1 concerning further guidance about types and settings of hearing aid compressors.

5 When fitting a compressor, the audiologist will take into account the measured auditory characteristics, e.g. the hearing threshold and the upper comfort level (UCL). The upper comfort level is the maximum signal level the user can accept without any pain or damage. Of course, the maximum output level can be limited by other characteristics, such as the limited dynamic range of the input or output converter.

10 In suggesting the hearing aid parameters, the audiologist will rely on miscellaneous standards (sometimes denoted fitting rules) that suggest suitable hearing aid transfer functions in dependency of the character of the actual hearing loss.

The compressor characteristic may be established by entering the audiometric data in a fitting program whereby the fitting equipment transforms this information to the required compressor characteristic parameters to be programmed in the hearing aid.

15 Now, in order to suppress the perception of occlusion of the ear canal, for such a system, it is required to adjust the setting of the high level compression ratio in a number of channels. Preferably, the adjustments for occlusion reduction is done as the first process. This may be done by adjusting the compression ratio for a range of signal levels in each of the relevant channels, but in a preferred embodiment the compression ratio of the high level section, e.g.
20 over 50 dB, is pre-set to a compression ratio of e.g. 2:1. In this preferred embodiment the final fitting to the users preferences is done by adjustment in each channel of the over-all channel gain (the gain scale) of the compressor characteristic, as shown in Figure 5, by adjusting the amplification at each signal level with an equal amount. As shown in the figure this has the effect of changing the position of the curve by adding (or subtracting) equal amounts of gain to
25 any point on the curve. Of course, it must be ensured that the UCL is still respected, e.g. by providing a limiter at the output of the compressor or by limiting the range of gain adjustments.

In this respect, it is important to note, that there are several ways of fitting a compressor characteristic. As mentioned above, one preferred way is to enter the amplification needed at three points (fix-points), whereby the fitting equipment translates these settings to suitable
30 parameters to be programmed in the hearing aid. It will be obvious to a skilled person that several other methods for this programming can be applied without departing from the scope of the invention. Regarding a block-mode fitting approach, it is possible to specify that changes to

a compressor setting is specified as a change to the gain at each of the fix-points. In this way, even different settings in a block of channels will be changed with an equal amount - contrary to be changed to the same amount - such that the relative difference is maintained even though the absolute settings is changed with the same amounts.

- 5 Research has shown that two frequency ranges, approximately 0-350 Hz and 350-500 Hz, have distinctly different effects on the perception of the occlusion effect. Therefore, it is an advantage to operate with filter bands and compressors to separately cover these ranges. In addition, it might be advantageous to avoid selecting widely different gain settings in neighboring frequency bands. This may be addressed by implementing a number of frequency
10 bands with respective compressors for permitting a fine graduation of amplification settings.

- In one embodiment, offset amplifiers are provided for adjusting the compressor characteristic in each of the low frequency channels subjected to compression, for reduction of the perception of occlusion of the ear canal. Fig. 5 shows such compressor characteristic adjustments as a displacement of the compressor characteristic. In Fig. 5, characteristic 13 corresponds to the
15 characteristic shown in Fig. 4, and the characteristics 14 and 15 illustrate possible displacements in response to gain adjustments. It is preferred to provide compressor characteristic adjustment in the range from - 20 dB to + 20 dB.

- It should be noted from Fig. 4 that the illustrated fine tuning of the compressor characteristic provides an adjustment of the balance between enhancement of low level signals and
20 attenuation of high level signals at the frequencies at which the compressor in question operates.

- Fig. 6 shows a schematic block diagram of a hearing aid 20 according to the invention. It will be obvious for the person skilled in the art that the circuits indicated in Fig. 6 may be implemented using digital or analogue circuitry or any combination hereof. In the present
25 embodiment, digital signal processing is employed and thus, the processor 28 consists of digital signal processing circuits. In different implementations of this embodiment, all the digital circuitry of the hearing aid 20 may be provided on a single digital signal processing chip or, the circuitry may be distributed on a plurality of integrated circuit chips in an appropriate way.

- In the hearing aid microphone 22 is provided for reception of a sound signal and conversion of
30 the sound signal into an electrical signal representing the received sound signal. The hearing aid microphone 22 may comprise a plurality of input transducers, e.g. for the purpose of implementing directional sensitivity characteristics. The microphone 22 converts the sound

signal to an analogue signal. The analogue signal is sampled and digitized by an A/D converter 24 into a digital signal 26 for digital signal processing in the hearing aid 20. The digital signal 26 is fed to a digital signal processor 28 for amplification of the microphone output signal 26 according to a desired frequency characteristic and compressor function to provide an output
5 signal 30 suitable for compensating the hearing deficiency of the user. The output signal 30 is fed to an D/A converter 32 and further to an output transducer 34 that converts the output signal 30 to an acoustic output signal.

The signal processor 28 comprises a filter bank 36 with bandpass filters 36_i for dividing the electrical signal 26 into a set of bandpass filtered first electrical signals $26_1, 26_2, \dots, 26_i$. Further,
10 the signal processor 28 comprises a set 38 of compressors and offset amplifiers $38_1, 38_2, \dots, 38_i$ each of which is connected to a different bandpass filter $36_1, 36_2, \dots, 36_i$ for individual compression of the corresponding bandpass filtered signal $26_1, 26_2, \dots, 26_i$, the compressor and offset amplifiers 38_1 and 38_2 in the low frequency bands 36_1 and 36_2 having compression ratios that have been increased in accordance with the invention.

15 The illustrated compressor characteristics 38_1 and 38_2 correspond to the characteristic shown in Fig. 4, and the characteristic 38_i corresponds to the characteristic shown in Fig. 3. Filters 36_1 and 36_2 are low frequency bandpass filters or low-pass filters, e.g. filters with passbands below 500 Hz. Filter 36_1 may have a passband below 300 Hz and 36_2 may have a passband between 300 Hz and 500 Hz.

20 In another embodiment of the invention, the set of compressors comprises four compressors, each with a compressor characteristic of the type shown in Fig. 4, i.e. each of the two bands described in the previous segment is divided further into two frequency bands with a compressor operating in each band.

During fitting and/or fine tuning, compressors in neighboring bands may be grouped together
25 for simultaneous adjustment of respective parameters. For simplicity, it is preferred that corresponding parameters of compressors in a specific group are adjusted by the same value.

As mentioned above, the use of an occlusion manager according to the invention is particularly advantageous in conversation situations. On the other hand, since the dynamic compression is enhanced when the occlusion manager is active there may be situations where it is desirable not
30 to use the manager. Since multi-program hearing aids is well-known it will thus be an obvious solution to include the occlusion manager in at least one of these programs, such that the user has at least one program optimized for conversation - with occlusion reduction - available.

EXAMPLE

One specific example of the fitting according to the invention shall be given as follows:

The hearing loss in the left ear of one patient had been measured at a number of frequencies as follows:

Frequency Hz	250	500	1000	1500	2000	3000	4000	6000	8000
Hearing loss dB	20	20	20	30	40	45	50	55	60

- 5 A first fitting rule, identified as NAL-NL1, 2000 Scott Brewer National Acoustic Laboratories Australian Hearing Services, was applied on this set of data. Set-points for frequencies, compression thresholds and compression ratios as suggested by this fitting rule were:

Frequency Hz	400	1250	2000	4000
Comp. Threshold	52	51	57	59
Comp. Ratio	1.16	2.03	2.68	1.11

A second fitting rule, identified as DSL Hearing Health Care Research Unit, The University of Western Ontario, was applied to the same set of data. Suggested set-points regarding

- 10 frequencies and compression ratios as suggested by this fitting rule were:

Frequency Hz	250	500	750	1000	1500	2000	3000	4000	6000
Comp. Ratio	1.4	1.4	1.4	1.4	1.6	1.9	2.1	2.4	2.9

A third fitting rule, identified as CamFit, Brian C. Moore 2002, was applied to the same set of data. Suggested set-points regarding frequencies and compression ratios as suggested by this fitting rule were:

Frequency Hz	125	250	500	750	1000	1500	2000	3000	4000	6000
Comp. Ratio	1.00	1.00	1.11	1.26	1.41	1.64	1.73	1.98	2.31	2.61

- 15 Common to all of these settings is, that the compression ratios suggested are moderate in the low frequency range, where the hearing loss of this patient is moderate.

These settings have all been tested. When interviewed, the patient complained about an annoying perceived occlusion effect.

Additional tests have been conducted wherein the settings as suggested according to each of the fitting rules were modified by increasing the compression in the low frequency range, i.e. the range below approximately 1000 Hz, to a compression ratio setting of 2.0. The patient was again interviewed. The patient reported, in respect of each of the additional tests, that the

5 hearing aid operated well with no problems of perceived occlusion.

CLAIMS

1. A fitting method for a hearing aid with at least one low frequency channel having an individually adjustable compressor, comprising the steps of determining initial values of settings of the hearing aid, including the compression ratio of the compressor, that will
5 compensate the hearing loss, and increasing the compression ratio over the initial value of the compression ratio in the at least one low frequency band.
2. The fitting method according to claim 1, wherein the at least one low frequency channel further comprises an offset amplifier adding an offset gain to the compressor characteristic, and further comprising the step of adjusting the offset gain in the range from -20 dB to 20
10 dB.
3. The fitting method according to claim 1 or 2, wherein the increased compression ratio is greater than 1.4.
4. The fitting method according to claim 1 or 2, wherein the increased compression ratio is substantially equal to 2.
- 15 5. The fitting method according to any of the preceding claims, wherein each of the hearing aid channels comprises an individually adjustable compressor and an offset amplifier.
6. The fitting method according to claim 1, wherein the range of signal levels at which the compression is active extends from 30 dB SPL to 60 dB SPL.
7. The fitting method according to claim 1, wherein the range of signal levels at which the
20 compression is active extends from 25 dB SPL, and more preferred from below 20 dB SPL and to 70 dB SPL, more preferred to 80 dB SPL.
8. The fitting method according to claim 1, wherein the range of signal levels at which the compression is active is selected in respective frequency bands.
9. The fitting method according to claim 1, wherein the low frequency band comprises
25 frequencies below 1600 Hz, preferably below 1000 Hz, more preferred below 800 Hz, and most preferred below 500 Hz.
10. A hearing aid having a microphone, a signal processor and an output transducer, wherein the signal processor comprises a filter bank having at least a low frequency band and a high

frequency band with respective compressors and with at least one offset amplifier in respect of the low frequency band compressor for increasing the compression.

11. The hearing aid according to claim 10, wherein the low frequency band extends below 500 Hz.
- 5 12. The hearing aid according to claim 10, wherein the filter bank has a passband below 300 Hz, a passband extending from 300 Hz to 500 Hz and at least one passband extending above 500 Hz, and with a compressor in respect of each passband.
- 10 13. A multichannel hearing aid comprising at least one input transducer for transforming an acoustic input signal into a first electrical signal, a filter bank with bandpass filters for dividing the first electrical signal into a set of bandpass filtered first electrical signals, a processor for generation of a second electrical signal by individual processing of each of the bandpass filtered first electrical signals and adding together the processed electrical signals into the second electrical signal, and an output transducer for transforming the second electrical signal into an acoustic output signal, wherein the processor comprises a set of compressors, each of which is connected to a respective bandpass filter for
15 compression of the corresponding bandpass filtered signal, and wherein at least one of the compressors is a low frequency compressor adapted to compress low frequency, high level signals.