Digital Signal Processing for Over-the-Counter Hearing Aids

Whitepaper

by Alexander Goldin, PhD,
Alango Technologies CEO

Revision 1.2 (March 2023)

Contents

1 What this whitepaper is about

In this whitepaper, we look at the principles of digital signal processing, algorithms and software that improve hearing for people with hearing loss. The inspiration for this whitepaper comes from the World Health Organization's recent World Report on Hearing (WHO), which shows that 1.5 billion people worldwide have some degree of hearing loss and 430 million people need rehabilitation services. Shockingly, less than 3% (on average worldwide) use hearing aids. Can anything be done to change this? For example, can True Wireless Stereo (TWS) earphones that use the same hardware components serve as over-the-counter hearing aids (OTC)? The answer is yes, if their software is upgraded with algorithms specifically tailored to the needs of the hearing impaired. The main goal of this text is to provide readers with a general understanding of these algorithms and how they can improve hearing. Our goal is to present this information in a simple and intuitive way, without using formulas, to make it accessible to everyone.

2 Hearing loss, hearing thresholds, and hearing tests

In audiology, hearing loss is quantified by measuring hearing thresholds for each ear at different frequencies. During a hearing test, the person being tested typically sits in a soundproof booth wearing headphones. The audiologist then plays a series of tones of varying frequencies and loudness levels, one ear at a time, and the person indicates when they can hear each tone. The test continues at different frequencies until there is a complete picture of the person's hearing thresholds for each ear. These thresholds are expressed relative to those of an average individual with healthy hearing, usually around 0-10dB of Sound Pressure Level (SPL) and then plotted on an audiogram, which is a graph that shows the hearing thresholds relative to those of an average individual with healthy hearing. Audiograms, which illustrate these relative thresholds, are used to summarize the collective data. A sample audiogram is provided in *Figure 1.*

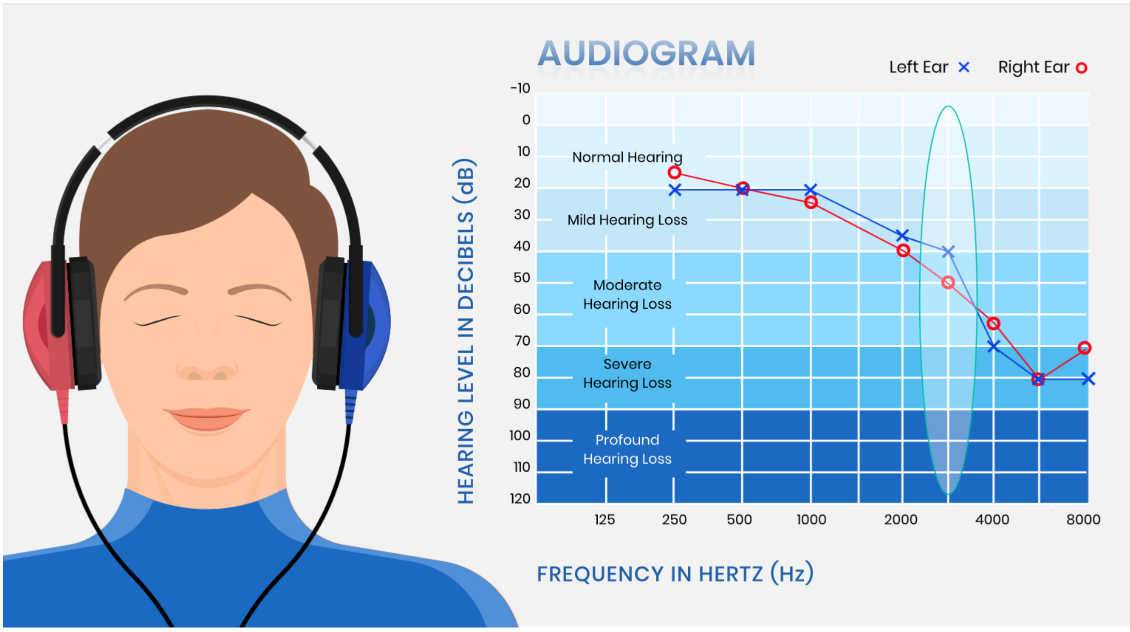

Figure 1: Hearing test and audiogram example

The above audiogram indicates that the tested person has moderate hearing loss in the frequency region around 3000Hz, with a minimal loudness level 40dB higher in the left ear and 50dB higher in the right ear compared to a person with healthy hearing. However, simply amplifying sounds around 3000Hz by 40dB and 50dB respectively will not restore the person's hearing to normal. Hearing loss is not a linear reduction in ear sensitivity that can be compensated for by a simple or even frequency-dependent linear sound amplification. Instead, hearing loss is both frequency and sound level dependent, meaning that the degree and type of hearing loss can vary significantly depending on the specific frequencies and sound levels involved. *Figure 2* illustrates this concept, although the numbers are given for illustration purposes only.

Healthy loudness growth

Impaired loudness growth

Perception	Sound Pressure Level	Perception
Barely heard	10 dB - snowfall	Not heard
Quiet	30 dB - whispering	Not heard
Clear	50 dB - quiet conversation	Not heard
Normal	**60 dB** - normal conversation	Barely heard
Loud	**70 dB** - loud conversation	**Loud**
Very loud	90 dB - jackhammer at 1 m	Very loud
Too loud	100 dB - inside discotheque	Too loud
Painful	110 dB - near propeller plane	Painful

Figure 2: Sound levels and their perception

One of the main characteristics of hearing loss is "loudness recruitment", where the perceived loudness of sounds grows much faster for people with hearing loss than for those with normal hearing. As a result, the difference in sound level between "barely heard" and "normal" becomes very short for people with hearing loss, which can explain why they often have difficulty hearing others speaking to them in a normal voice. This phenomenon is illustrated in *Figure 3*, where the loudness for people with normal hearing grows steadily, while it remains at zero until the personal hearing threshold is reached for people with hearing loss starting to grow abnormally fast afterwards till the point where it becomes equal to people with normal hearing.

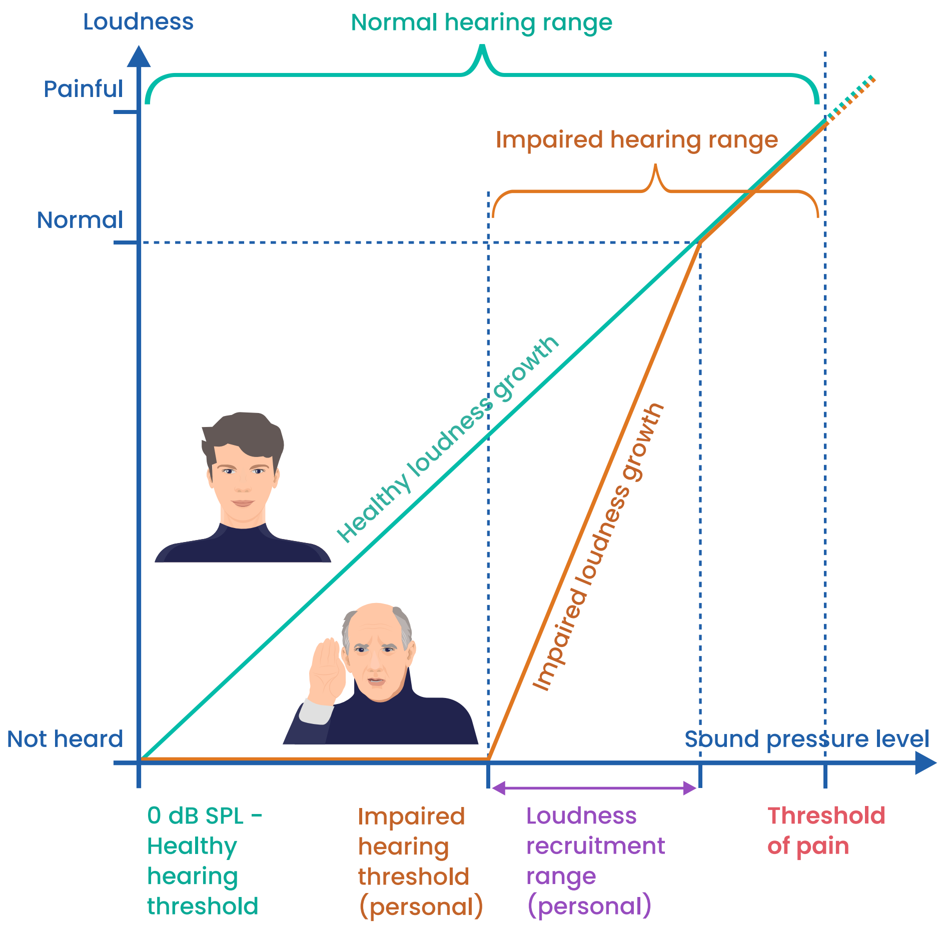

Figure 3: Loudness and hearing recruitment

This phenomenon has led to the development of multi-channel Wide Dynamic Range Compression (WDRC) for hearing loss compensation in hearing aids. However, hearing loss is a complex, non-linear, frequency-dependent, and cognition-related phenomenon that cannot be universally compensated by amplifying sounds. Nonetheless, proper hearing amplification can still benefit most people with hearing loss to some extent. Our goal is to make such amplification as efficient as possible. It's important to note that all thresholds and corresponding plots in *Figure 3* are individual and frequency-dependent.

3 Hearing aids categories

Hearing aids are the most commonly used hearing enhancement devices. Air conduction hearing aids amplify ambient sounds based on the user's specific hearing loss and/or preferences, and functionally consist of a microphone, an amplifier with controlled amplification, and a receiver (speaker) as shown in *Figure 4*.

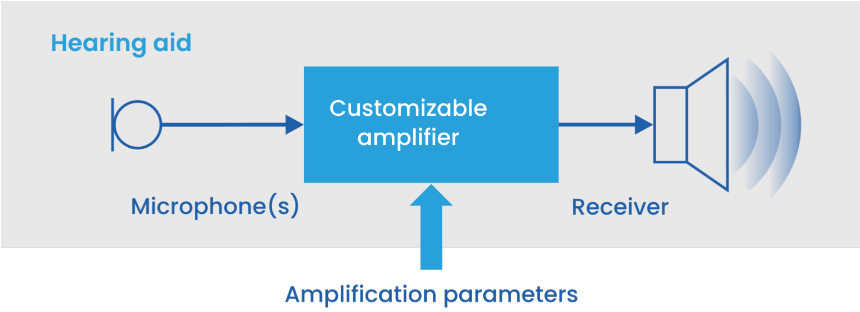

Figure 4: Hearing aid structure

While microphones and receivers are typically high-quality, linear components, the amplification is generally non-linear, with properties and characteristics that define the features and quality of hearing aids. Hearing aids can be divided into two main categories based on the type of amplification control:

1. Prescription hearing aids are Class 1 medical devices distributed through professional hearing healthcare channels, and may be designed to cover all types of hearing loss from mild to profound. The amplification parameters of prescription hearing aids must be changed through professional fitting services and cannot be adjusted by users themselves.
2. Over The Counter (OTC) hearing aids are a new category recently defined by the US Food and Drug Administration (FDA) to provide a solution for people with mild to moderate hearing loss. OTC hearing aids offer similar functionality to prescription hearing aids for people with mild to moderate hearing loss, but can be sold over the counter and administered by users without the help of hearing care professionals. OTC hearing aids are designed to enable users to directly control the amplification characteristics or perform a hearing test and set application parameters accordingly, to provide an optimal user experience.

4 Specifications of OTC hearing aids

According to the FDA, OTC hearing aids can be any ambient sound amplification device that satisfies the following requirements:

1. An air conduction device that does not require implantation or any surgical intervention and does not require the supervision, prescription, physician order, involvement, or intervention of licensed professionals.
2. Must not harm the user’s ears and must be comfortably worn.
3. Customizable and controllable by the user through the use of tools, tests, and/or software with user-controlled volume accessible on device itself, and or via app, remote control
4. Must be labeled as “OTC Hearing Aid for Adults with Perceived Mild-to-Moderate Hearing Loss”

5. May use wireless technology (Bluetooth or otherwise) for connecting to phones, remote microphones, or other sound sources.
6. The tip of the receiver (speaker) is no closer than 10mm from the eardrum.
7. Must pass ANSI/CTA 2051 Electro-Acoustic Quality Standard requirements:
   a. Input/Output distortion < 5%
   b. Self-generated noise levels < 32 dBA
   c. Signal processing latency < 15ms
   d. Frequency bandwidth between 250 Hz to 5 KHz
   e. Frequency response smoothness must not have any single peak on any 1/3 octave band frequency response that exceeds 12 dB relative to the average levels of the 1/3 octave bands.
8. Maximum SPL output at any frequency:
   a. For devices with linear amplification: 111 dB
   b. For devices with input-controlled dynamic range compression: 117 dB

There are no constraints on OTC hearing aid form factors, creating opportunities for innovation, increased competition, and reduced prices. There is a big hope today that OTC hearing aids will lead to much wider use of hearing amplification devices by people who, for whatever reason, do not benefit from prescription hearing aids.

5 Basic digital signal processing in modern hearing aids

All modern hearing aids are digital, i.e., analog electrical microphone signals are first digitized, then processed in the digital domain, and finally converted to an analog signal and reproduced through a receiver (loudspeaker).

The typical digital signal processing blocks of a high-end hearing aid are shown in *Figure 5*.

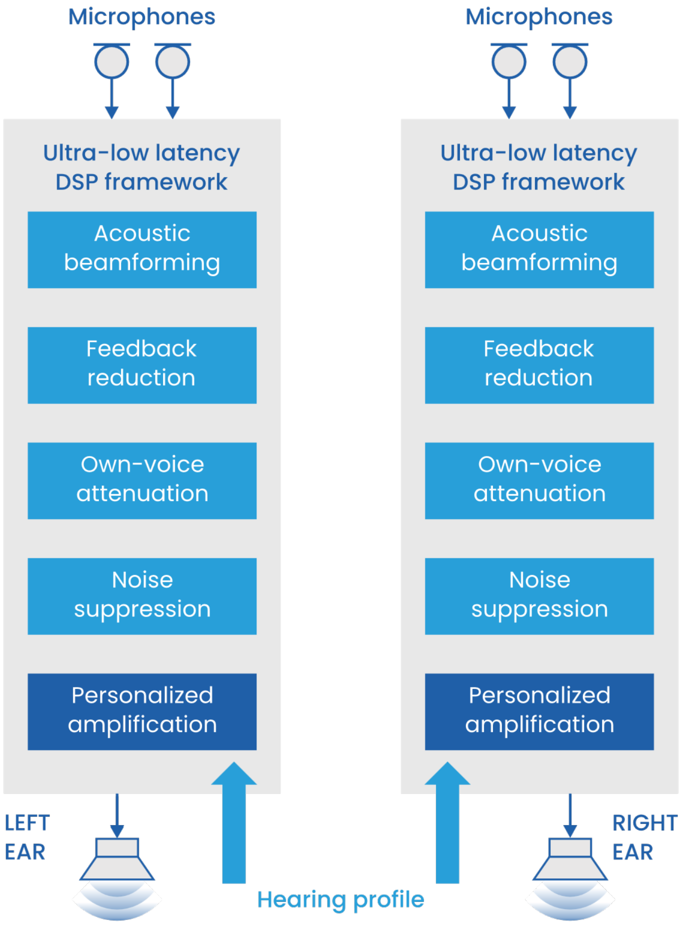

Figure 5: Standard DSP tasks in hearing aids

5.1 Frequency sub-bands signal decomposition

Most of the digital signal processing in hearing aids is done in frequency sub-bands. The full-band input signals are divided into frequency sub-band signals, as shown in *Figure 6*, with each sub-band conveying the information for the corresponding narrow frequency range.

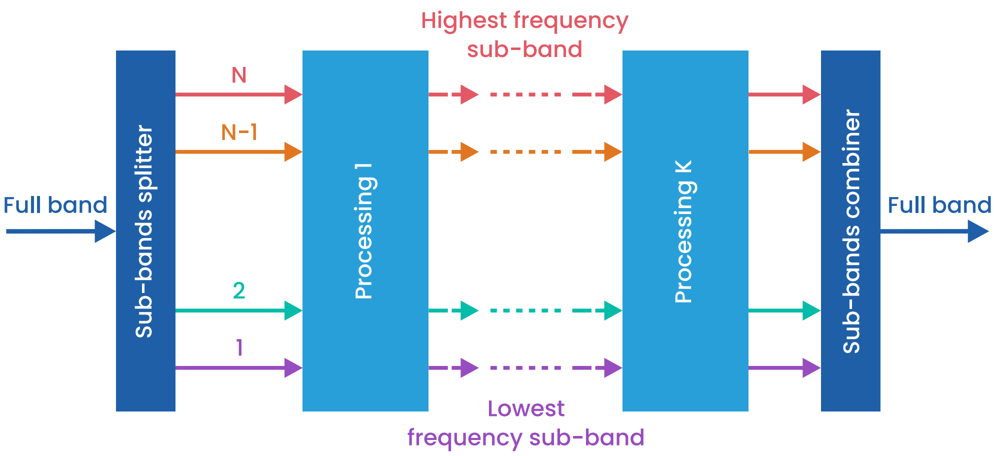

Figure 6: Sub-band processing scheme

Digital signal processing is sequential, block by block. After all the processing is done, the full-band signal is resynthesized (combined) from the processed sub-band signals.

In some cases, the processing blocks can receive sub-band signals from multiple sources and combine them into a set of sub-band signals. Acoustic beamforming is one type of this processing, shown in *Figure 7*.

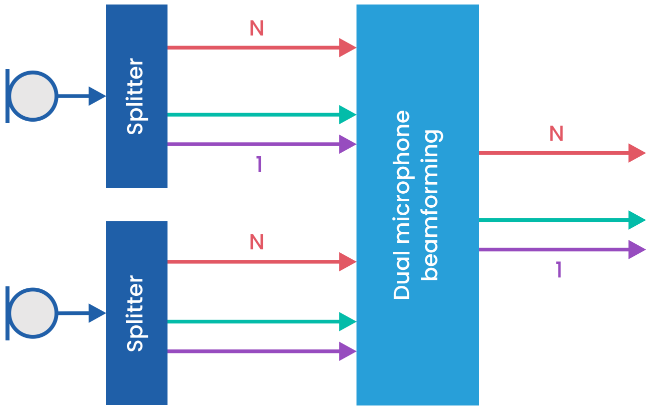

Figure 7: Acoustic beamforming sub-band merging

Splitting into and combining frequency sub-bands can be accomplished using various and well-known methods such as the short-time Fourier transform (STFT) or filter banks. Finite impulse response (FIR) or infinite impulse response (IIR) filter banks can be used, each having its own advantages and disadvantages. For example, IIR filter banks provide the shortest delays but are less suitable for implementing other algorithms such as adaptive beamforming, noise reduction, and others. When STFT is used for decomposition, the sub-bands contain complex numbers and are said to be processed in the "frequency domain."

The number of sub-bands, their frequency distribution, and overlap depend on the method of sub-band decomposition as well as a tradeoff between frequency resolution, processing delay, computational complexity, available data memory on a given chip, and other affected parameters.

5.2 Personalized hearing amplification

Modern digital hearing aids use multi-channel Wide Dynamic Range Compression (WDRC) to achieve personalized amplification. This involves digitally amplifying and compressing the wide dynamic range of sounds heard by a healthy ear into a narrower range. As ambient sound increases, amplification is naturally reduced compensating for the loudness recruitment. For example, the original range of sounds from distant rustling to a pop concert is compressed and reproduced as a range from a vacuum cleaner to a pop concert, as shown in the *Figure 8*.

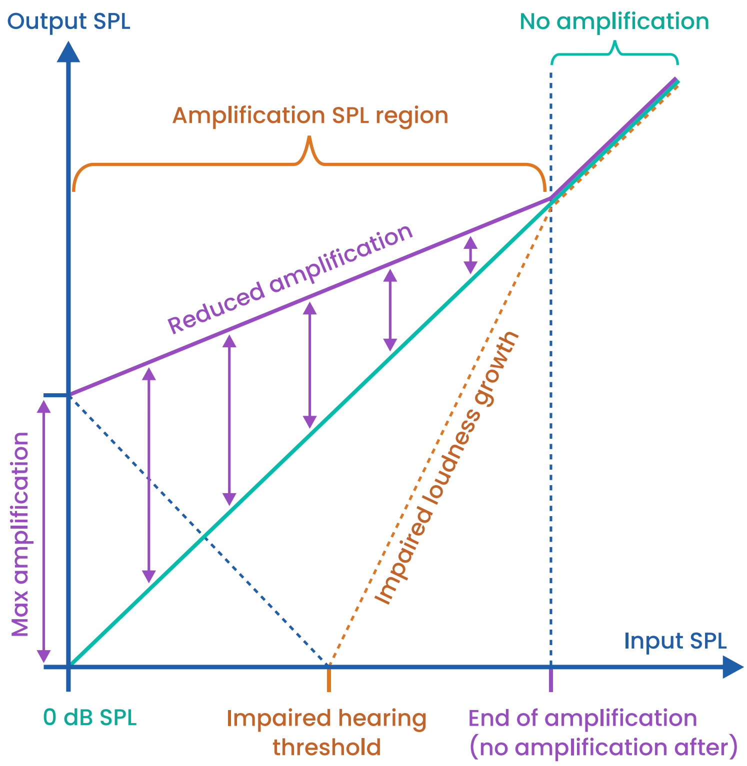

Figure 8: Wide Band Dynamic Compression principles

"Multichannel WDRC" means that the input signals are processed in "frequency channels," as shown in *Figure 9* for a four-channel system.

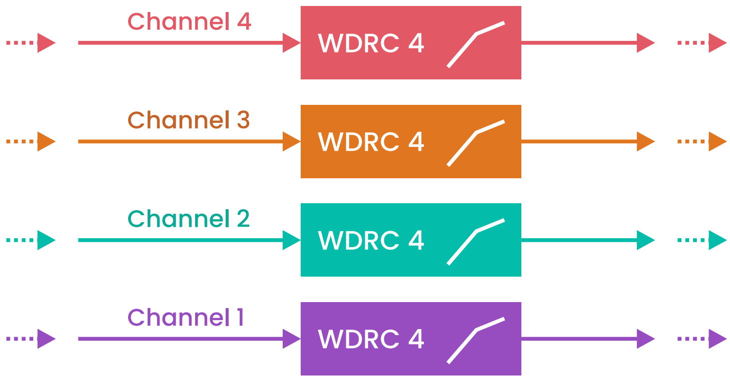

Figure 9: Multi-channel WDRC scheme

The channels are amplified/compressed differently according to a specific hearing loss. Each channel usually contains signal frequencies that correspond to a certain predefined frequency range and partially overlap. The channels can correspond to the frequency sub-bands one-to-one or combine several and even an unequal number of sub-bands, depending on the required frequency resolution. In practical implementations for hearing aids, the number of frequency channels can vary from 4 to 64.

In general, gain/compression parameters are set according to individual hearing thresholds measured directly or indirectly during a hearing test. *Figure 10* illustrates the input/output characteristics of a four-channel WDRC corresponding to a high-frequency loss.

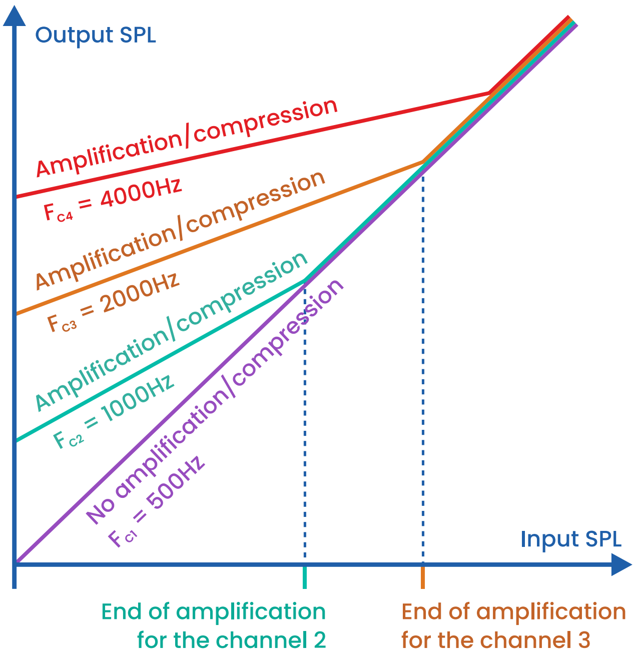

Figure 10: Example compensation for a high tone frequency loss

No hearing loss at 500 Hz corresponds to the magenta "unity gain" line where the input and output sound pressure levels are equal. Hearing loss progresses with frequency and amplification is increased accordingly. For all frequency channels, the amplification is reduced with the signal level. It becomes unity at a specific level, typically near to the end of the loudness recruitment region for the frequency corresponding to the channel.

5.3 Acoustic feedback reduction

Howling, whistling, and squelching are common problems reported by hearing aid users that are due to acoustic feedback. As shown in *Figure 11*, acoustic feedback occurs because the receiver and microphone are not perfectly separated, so the microphone signal reproduced by the receiver is picked up again by the microphone.

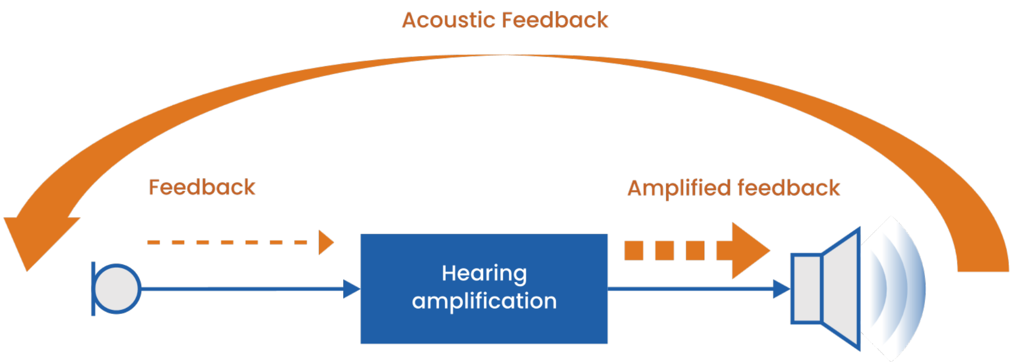

Figure 11: Acoustic feedback

When no amplification is applied, the feedback signal at the microphone is naturally attenuated by the air conduction. However, when the microphone signal is amplified and reproduced by the receiver beyond a certain level, the signal enters an infinite microphone → speaker → microphone loop, with the sound amplified more and more in each circle, making the system unstable and uncontrollable. This devil's loop does not occur simultaneously throughout the entire operating frequency range. Feedback problems begin earlier at frequencies with higher overall system gain. For hearing aids, the frequency range between 3 and 5 kilohertz is the most problematic. The Feedback Reduction block is introduced to reduce the effects of acoustic feedback and allow higher amplification.

There are two major approaches to reducing acoustic feedback: feedback cancellation and feedback suppression.

Feedback cancellation, shown in *Figure 12*, attempts to simulate the acoustic path from the receiver to the microphone, estimate the feedback signal and subtract it from the microphone signal.

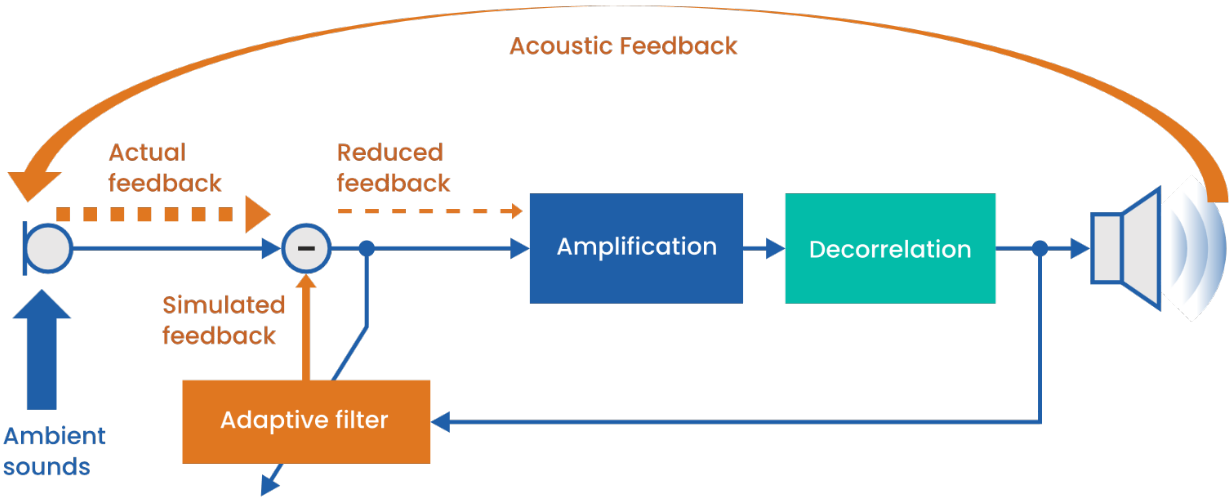

Figure 12: Acoustic feedback cancellation

This is generally done using an adaptive filter controlled by the result of the subtraction. While this problem looks similar to acoustic echo cancellation, there is a fundamental difference that makes it more challenging. In acoustic echo cancellation, the signal to be cancelled (the acoustic echo) and the signal to be preserved (the speech in the near field) are not correlated. Therefore, minimizing the cancellation error (the result of the subtraction) is the valid objective for the adaptive filter coefficients. In acoustic feedback cancellation, both the signal of interest and the feedback are very similar signals with a high degree of correlation, especially at low frequencies. If an acoustic echo cancellation approach is used directly, it will cause the adaptive filter coefficients to converge to incorrect values, resulting in cancellation of the signal of interest. To avoid the erroneous convergence, a decorrelation, in the mathematical sense, is required between the output of the hearing aid and its input. Decorrelation can be accomplished, for example, by a small frequency shift or scaling of the output that is not yet perceived by the user as signal distortion or pitch shift. In general, frequency shifts of up to 3 Hz can be used, which makes the task of decorrelation easier, but does not completely solve it. Ultimately, a compromise must be found between the convergence speed of the filter, its stability and the cancellation of the signal of interest.

Feedback suppression does not attempt to simulate and predict feedback, but works like an adaptive frequency equalizer that depends on the spectrum of the microphone signal, as shown schematically in *Figure 13*.

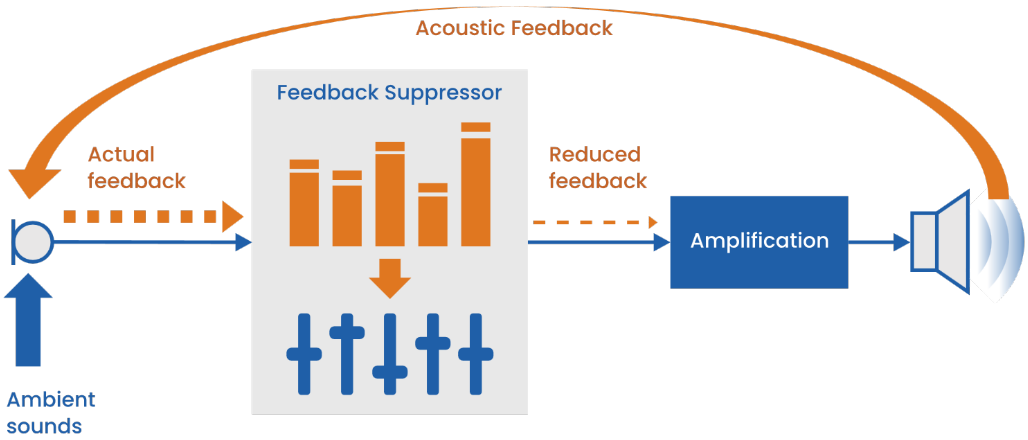

Figure 13: Acoustic feedback suppression

Acoustic feedback often manifests as a sudden and unrealistic increase in spectral amplitude within narrow frequency ranges. When detected, corresponding frequencies are attenuated to mitigate the unpleasant artifacts. However, acoustic feedback occurs rapidly in hearing aids, providing limited time for feedback suppression algorithms to analyze the microphone signal spectrum and respond before the feedback becomes perceptible. Additionally, a challenge in feedback suppression is determining the appropriate time to release the suppressed frequency range and provide the required amplification.

To achieve the best performance, the feedback suppressor and canceller can be combined in one algorithm. In addition, binaural decisions can be made by comparing the signal analysis in the two hearing aids to detect feedback. In-ear microphones can be used to analyze the differences in arrival time, and many other "inventions" can be made. The field of acoustic feedback cancelation offers much scope for specific heuristic approaches and know-how for differentiation.

5.4 Acoustic beamforming

Most modern digital hearing aids are equipped with two microphones, positioned apart from each other (front and back) for acoustic beamforming. This technique generates a directional "sensitivity beam" towards the desired sound, preserving sounds within the beam while attenuating others. Typically, sounds from the front are preserved, while sounds from the sides and rear are considered unwanted and suppressed. However, different beamforming strategies can be applied in specific environments. Acoustic beamforming relies on a slight time delay, typically in tens of microseconds, for sounds to reach the two microphones. This time difference is dependent on the direction of the sound and the distance between the microphones, as illustrated in *Figure 14*.

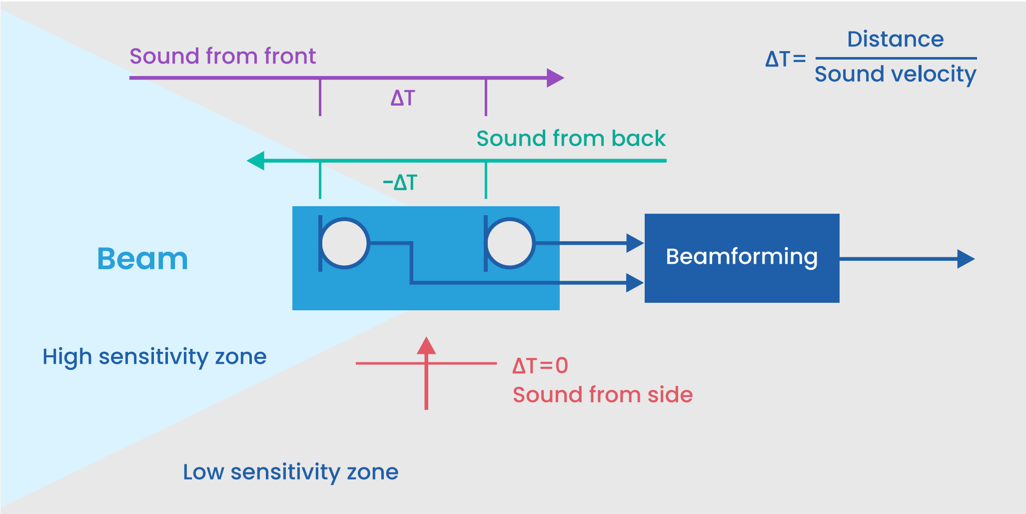

Figure 14: Acoustic beamforming principles

Front sounds reach the front microphone first, side sounds arrive simultaneously, and rear sounds reach the rear microphone first. Acoustic beamforming utilizes this time difference to create a variable sensitivity called a "polar pattern" or "polar sensitivity pattern" that is directionally dependent. This is achieved through digital signal processing techniques. *Figure 15* shows the four well-known theoretical polar patterns that can be achieved in a free field with dual microphone beamforming.

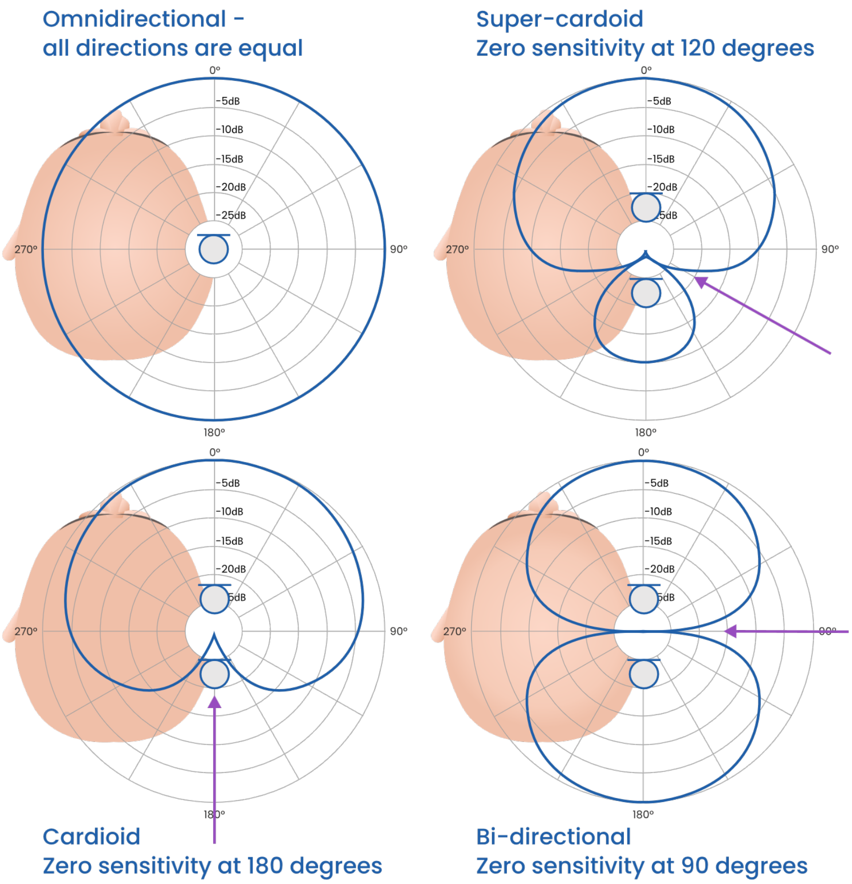

Figure 15: Typical microphone polar patterns

While sensitivity to front sounds stays consistent, sensitivity to sounds from other directions decreases with one angle having zero sensitivity. However, these theoretical polar patterns only apply in a free field and are not realistic in practical situations due to acoustic distortion caused by the user's head and torso.

Acoustic beamforming can be either fixed or adaptive. Fixed beamforming has a static polar pattern that is independent of the microphone signals and acoustic environment, making it easy to implement and computationally efficient. Adaptive beamforming, on the other hand, changes the polar pattern in real time to steer the null of the sensitivity polar pattern toward the direction of the strongest noise, optimizing array performance.

Adaptive acoustic beamforming can be implemented in either the full frequency band or in frequency sub-bands. The full band scheme optimizes an average performance with a uniform polar pattern across all frequencies in the hearing aid frequency range, which may not be optimal for individual frequencies. In the sub-band scheme, polar patterns are tailored to different frequency regions to optimize performance across frequencies. This can be more efficient for practical scenarios where noise comes from different directions in different frequency regions.

5.5 Noise reduction

The primary concern of individuals with hearing loss is difficulty communicating in noisy environments. While acoustic beamforming can effectively reduce noise from specific directions, it has a limited ability to mitigate diffuse noise - noise that emanates from multiple sources without a distinct direction. Diffuse noise arises from multiple noise reflections off hard surfaces in a reverberant space, which further complicates noise directionality. For example, noise inside a car is often diffuse, as uncorrelated noises from various sources such as the engine, tires, floor, and roof are diffused by reflections off the windows.

Compensating for hearing loss with Wide Dynamic Range Compression (WDRC) amplifies sounds that are below the user's hearing threshold, but it can also amplify noise and reduce the contrast between noise and the desired sound. Noise reduction technologies aim to mitigate these issues by suppressing noise prior to amplification by WDRC, thereby improving the output Signal to Noise Ratio (SNR).

Today, most noise reduction technologies used in hearing aids are based on spectral subtraction techniques. The input signal is divided into frequency sub-bands or channels, and the noise is assumed to have relatively low amplitude and be stationary across sub-bands, while speech exhibits dynamic spectral changes with rapid fluctuations in sub-band amplitude. If the average noise amplitude spectrum is known or estimated, a simple "spectral subtraction" strategy can be employed to suppress noise, as shown in *Figure 16*.

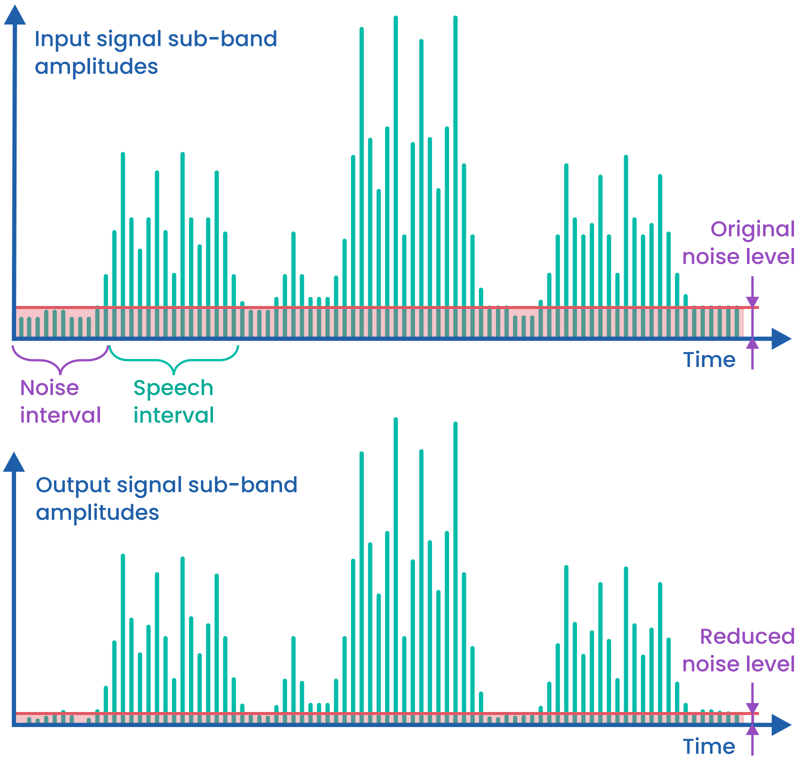

Figure 16: Spectral subtraction principles

The upper plot illustrates the sub-band signal amplitudes of a speech in noise signal across different frequency sub-bands. The relatively stationary noise amplitudes are averaged by the red line, while the speech signal amplitudes are significantly higher. In the lower plot, the sub-band noise amplitude shown by the red line in the upper plot has been subtracted from the sub-band input amplitudes using the spectral subtraction technique, which is applied to all sub-bands. Finally, the output full-band signal is reconstructed using the same method used for splitting, with the noise effectively reduced.

6 Alango digital signal processing for hearing aids

Alango Technologies has developed a full range of state-of-the-art DSP algorithms, software, and turn-key reference designs that can be scaled for various types of hearing enhancement devices, including over-the-counter (OTC) hearing aids, as well as true wireless stereo (TWS) earbuds and headsets with amplified transparency and conversation boost.

Figure 17 displays the DSP technologies that are integrated into the Hearing Enhancement Package (HEP) developed by Alango.

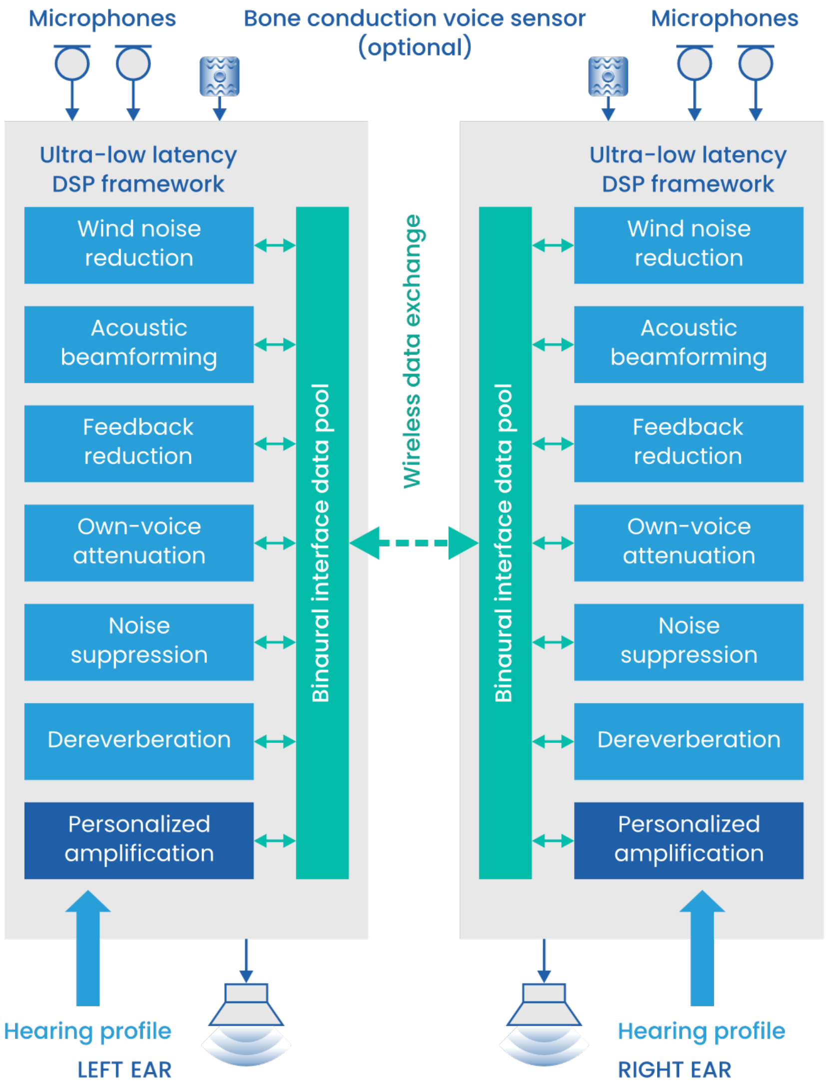

Figure 17: Alango DSP block for hearing aids

While some of the technologies share names with those shown in Figure 4, they are based on Alango's proprietary algorithms, which have been refined based on our experience and feedback received from customers.

6.1 Alango frequency sub-bands

Alango's DSP software blocks use a uniform frequency sub-band approach. Sub-band splitting and combining is achieved through Weighted Overlap Add (WOLA) methods based on Short Time Fourier Transform (STFT), where the STFT window and frame sizes are parameters. Optimal values for these parameters depend on the product, as they involve tradeoffs between processing latency, frequency resolution, computational complexity, memory size, and power consumption. Two main versions exist, with STFT window lengths of 4 and 8 milliseconds, providing frequency resolutions of 250Hz and 125Hz, respectively. The corresponding algorithmic/processing delays are 3 and 6 milliseconds, respectively, for the 4 and 8 millisecond STFT windows.

6.2 Binaural synchronization of Alango DSP technologies

The human brain processes information from both ears in a way that is not fully independent. This allows us to identify the direction of sound arrival and filter out sounds that come from directions other than the direction of interest. However, if the sound processing in two hearing aids is not synchronized, it can reduce the brain's ability to locate sounds accurately, especially in noisy or reverberant environments. As a result, it can create an unnatural and confusing perception of sounds.

To address this challenge, Alango's DSP hearing enhancement technologies support two types of binaural synchronization: full and partial. Full synchronization requires that microphone signals from both ears be available in each device. This can be achieved through either wired or wireless (e.g., Near Field Magnetic Induction) connections. However, in practical situations where dedicated hardware is not available, wireless (e.g., Bluetooth) connections between the two sides of the head may be unreliable or associated with transmission delays. In such cases, Alango supports partial synchronization via a messaging mechanism, in which the processing blocks on the two sides exchange limited information about their states.

All information exchanged during the operation is encapsulated into a single data chunk that is formed on one side and accommodated continuously on the other side. The device software developer only needs to query the relevant APIs and transmit the data between the devices. This approach allows the two hearing aids to work together seamlessly, despite the lack of full synchronization. It ensures that the brain can effectively process sound information from both ears, improving the user's ability to locate and identify sounds in their environment.

6.3 Alango hearing personalization

The basic principles of personalizing hearing with Alango technology are not different from those commonly used, as outlined in Section 5.2. In Alango's processing approach, frequency bins resulting from the Short-Time Fourier Transform (STFT) are combined into "channels," as shown in *Figure 18* for an example with five channels.

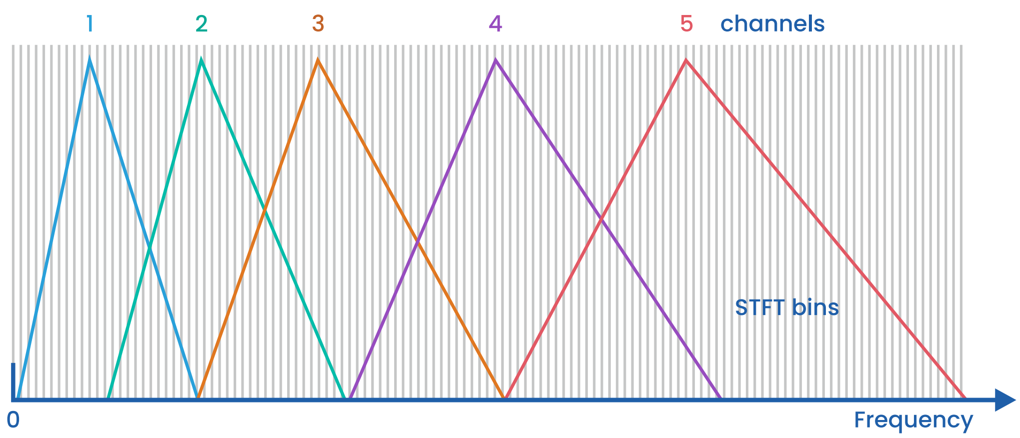

Figure 18: 5 frequency channels system example

Frequency channels are created based on the number and positions of their central frequencies, which are configured according to the product requirements. Each channel can have its own dynamic processing characteristic that corresponds to the user's frequency loss at the central frequency of the channel. With Alango technology, the dynamic characteristics of each channel are highly flexible and defined by six points. An example characteristic for one channel is illustrated in *Figure 19* by the magenta line. The plot displays the output Sound Pressure Level (SPL) generated by the device in relation to the input SPL of sound in the frequency range that corresponds to the channel's central frequency. The green line signifies the "unity line" of no amplification, where input and output levels are equal. The difference between the green and magenta lines represents the amplification.

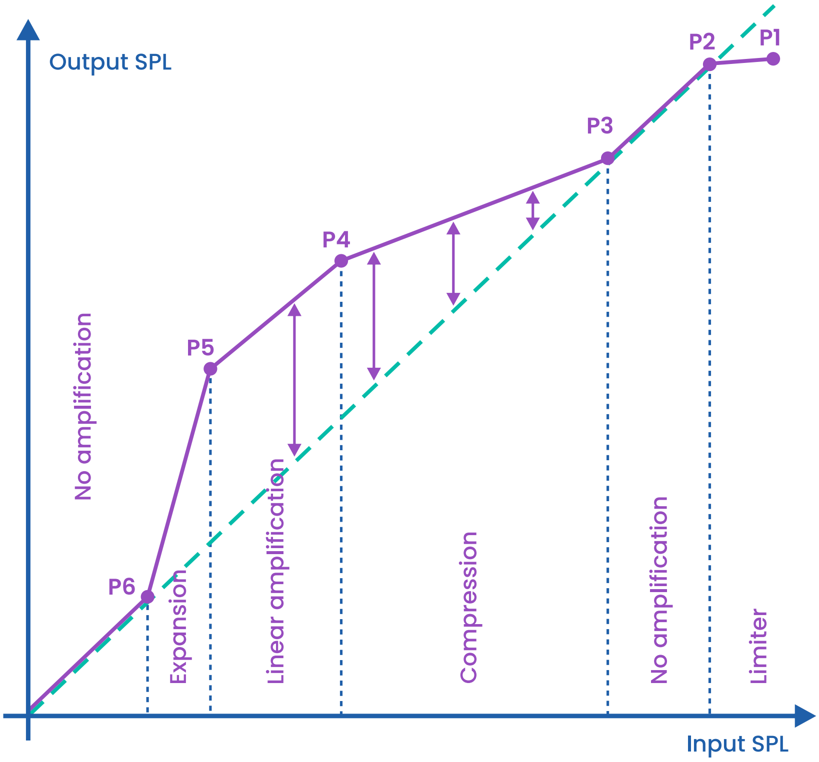

Figure 19: Alango 6 point dynamics processing

To prevent the amplification of ultra-low-level sounds, no amplification is specified before point P6. Amplification begins at point P6 and increases (expands) to a maximum at point P5. In the P5-P4 region, the amplification is linear, preserving the original Signal-to-Noise Ratio (SNR). The P4-P3 region corresponds to "loudness recruitment," after which hearing is normal, and no amplification is required. Consequently, amplification gradually decreases (compresses) from the maximum at point P4 to no amplification at point P3. After P2, the output sound pressure is limited to prevent potential ear damage. It's important to note that this is just an example, and other characteristics can be defined.

Although specifying points P1-P6 directly is possible, the Alango auxiliary software simplifies the process by transforming the results of a hearing test into optimal values of P1-P6 for the user.

Optimal hearing amplification is non-linear, and the difference between input and output levels (amplification) changes depending on the input level. As a result, natural sound perception and localization can only be achieved when amplification on both sides is coordinated (binaural amplification). Alango technology accomplishes this by exchanging the sub-band signal levels used to calculate amplification between two devices.

6.4 Alango feedback cancellation

Alango's acoustic feedback prevention algorithms leverage both feedback cancellation and feedback suppression principles outlined in Section 5.3. To achieve feedback cancellation, the proprietary adaptive filters based on LMS principles are utilized. To avoid unnecessary calculations, the range of frequencies where the adaptive filters are active is a parameter. Decorrelation is performed by frequency shifting through multiplication of each complex sub-band signal by a complex factor of unit amplitude rotating with the speed as required for the frequency shift. For uniform frequency shift across all frequencies, the same factors are used in all sub-bands, while different factors with phases proportional to the sub-band central frequency are used for frequency proportional shift. In practice, a combination of both is used to obtain the best results, which can also depend on the application profile, such as conversation, live music, nature sounds, and others.

Normal acoustic signals, such as speech, are generally wideband, while feedback typically occurs at a specific frequency where the amplitude quickly rises. To prevent feedback, Alango's algorithm analyzes microphone signal sub-bands and identifies those with amplitudes significantly exceeding adjacent bands, which often occurs at a specific frequency. If this situation persists beyond a predefined time (e.g., 50 milliseconds), the algorithm attenuates the signal in that sub-band. Alango's binaural feedback suppression algorithm takes advantage of the fact that feedback usually does not occur simultaneously on both sides of the head. It analyzes the spectra differences of microphone signals from both sides and attenuates sub-bands with significant and lasting differences. This binaural feedback suppression works for both full and partial synchronizations.

6.5 Alango acoustic beamforming

Alango's acoustic beamforming technology supports multiple microphones, but the dual microphone configuration is more common for hearing aids and TWS earbuds due to space and interaural connection limitations. The algorithm operates in independent frequency sub-bands, allowing each sub-band to have its own polar sensitivity pattern. This outperforms full band beamformers in practical scenarios with multiple noise sources or quasi diffused noise, and near acoustic barriers like the human head. Alango supports both full and partial beamforming synchronization between ears, improving directivity in the front direction and sound source localization for comprehension.

6.6 Alango wind noise reduction

Wind noise can be a significant issue for hearing aid users, as turbulence affecting the microphones is amplified and worsened by hearing loss. To mitigate this problem, Alango has developed Wind Noise Reduction (WNR) technology, which uses information from two or more device microphones to detect and reduce wind noise. The algorithm estimates the correlation between device microphone sub-signals in each frequency band, and identifies and attenuates sub-bands affected by wind.

If the microphone signals from both sides are available in each device (full synchronization), signals from the less windy side can be used for both ears, potentially improving speech intelligibility during conversations in windy conditions. However, this approach may reduce the ability to locate sound sources. When full synchronization is not possible and the microphone signals from the opposite side are not available, partial wind noise reduction synchronization can still be effective in reducing wind artifacts.

6.7 Alango noise suppression

"Alango's noise reduction technology operates in frequency sub-bands and uses a "spectral noise gating" approach to classify sub-band signals as "noise" or "signal" based on a proprietary voice activity detector detecting human speech. Noise reduction is then applied to the "noise" sub-bands, while the "signal" sub-bands are passed unmodified. This technology adapts quickly to changes in noise level and helps to reduce "musical noise," a common problem with spectral subtraction. For the best user experience, noise reduction on both ears should be synchronized, which Alango supports through both full and partial synchronization options.

6.8 Alango own voice attenuation

Hearing one's own voice amplified is one of the main complaints of hearing aid users. It is particularly acute when the ear canal is sealed to enhance the low-frequency range and reduce the likelihood of acoustic feedback. Ideally, the user's voice must be detected and removed from the microphone signal. However, distinguishing the user's voice from other voices in the microphone signal is very difficult and unreliable, to say the least, and requires full synchronization between the two ears.

Alango's approach uses an additional bone conduction sensor integrated into a hearing aid that is attached to the user's ear. The user's speech causes tiny vibrations of the skull bones, which are picked up by the sensor. The signal from the bone-conducted sensor sounds very similar to the user's speech picked up by the microphones, but is limited to about 1000 KHz or less. However, the bone-conducted signal is completely isolated from all ambient noise and can therefore be reliably used to detect and attenuate the user's voice.

While adding an additional sensor may seem costly and space consuming, it actually saves both compared to the NFMI chip and its antenna used in hearing aids for binaural synchronization. In addition, the same bone conduction signal greatly improves the quality of the outgoing speech signal for voice communication via a Bluetooth connection to a phone, which is becoming a "must have" for hearing aids.

Small, inexpensive MEMS bone conduction sensors are now being manufactured by several companies for use in TWS earbuds.

6.9 Alango dereverberation technology

Speech intelligibility for people with hearing loss is drastically reduced by reverberation in the room. Although the problem of restoring the original signal from a reverberant signal is generally unsolvable, some improvements are possible in practice. The Alango Hearing Enhancement Package includes reverberation suppression technology that significantly improves perceptual quality and reduces listener fatigue from additional mental effort by people with hearing loss.

7 Voice communication with hearing aids

Modern hearing aids are also communication devices similar to TWS earbuds. This functionality requires DSP tasks such as acoustic echo cancelation, microphone beamforming, noise reduction, automatic gain control. In addition, the incoming voice during a phone call needs to be modified in the hearing aids to improve intelligibility and compensate for the user's hearing loss. All of these tasks DSP are offered by Alango in its basic Voice Communication Package (VCP). In addition, Alango offers advanced proprietary voice communication DSP technologies such as OnlyVoice™ and EasyListen™.

OnlyVoice technology uses the bone conduction voice sensor, which is also used to reduce the user's own voice. However, the opposite is used for voice communication. Instead of attenuation, the user's voice is emphasized and isolated from ambient noise to be transmitted to the other side of the connection.

EasyListen significantly improves the intelligibility of incoming speech from the far end by dynamically slowing it down in real time. The accumulated delay is quickly recovered during pauses.

8 Hearing aids for streaming audio

With improved speaker technology and Bluetooth connectivity, hearing aids are becoming suitable for listening to streaming audio. As with voice communication, incoming audio must be personalized in terms of frequency response and dynamic range, or the user will miss much of the content, which cannot be compensated for by changing the volume. The Alango Audio Enhancement Package (AEP) enables such customization by applying multichannel WDRC with user-specific parameters to the audio signal and reducing the spectral difference so that soft, typically high-frequency sounds are not masked by strong low frequencies.

When audio streaming is used to listen to podcasts or news, the incoming voices of participants are modified in a manner similar to telephone conversations to improve their intelligibility for hearing-impaired users. For such use cases, Alango EasyListen technology can further improve speech intelligibility by slowing down the often rapid speech of podcasters and newscasters.

9 DSP core choice for hearing aids applications

To select or design a DSP for hearing aids, one need to compare different DSP cores and rank them according to certain criteria. The most important criteria are:

- Selection of available software algorithms

- Ease of porting unavailable algorithms
- Ease of debugging and optimization
- Possible power consumption to achieve the target uptime before recharging.

While we may have our own preferences, Alango partners with all major DSP IP provider and can deliver optimized solutions for them. Please, contact us for more details.

10 The roadmap of the hearing enhancement

The possibilities for DSP technologies in the realm of hearing are endless. In addition to the solutions we have already discussed, there are many other exciting opportunities to explore. For example, we could harness the power of neural networks to develop even more advanced noise reduction techniques, or leverage low latency wireless connectivity to create smart remote microphones with broadcasting capabilities. Additionally, active occlusion effect management and dynamic vent technologies offer promising avenues for addressing common hearing problems. Beyond this, we can even consider innovative approaches like hearing sense substitution via tactile sense and real-time speech tempo manipulation for improved intelligibility. While we couldn't cover everything in this whitepaper, we are always excited to hear from our readers about their specific needs, ideas and interests. We look forward to hearing from you!

11 Contact information

HQ Address: 2 Etgar St., PO Box 62, Tirat Carmel 3903213 Israel

Web site: www.alango.com, www.alango-behear.com

Email: info@alango.com

Tel: +972-4-858-0743, Fax: +972-4-858-0621