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Review Article

Functionality of hearing aids: state-of-the-art and future model-based solutions

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Abstract

A review about technical and perceptual factors in hearing aid technology, research and development is provided, covering current commercial solutions, underlying models of hearing loss for usage in hearing devices and emerging future technical solutions for hearing aid functionalities. A chain of techniques has provided incremental, but steady increases in user benefit, e.g. in the fields of hearing aid amplification, feedback suppression, dynamic compression, noise reduction and situation adaptation. The models describing the perceptual consequences of sensorineural hearing impairment describe the effects on the acoustical level, the neurosensory level and the cognitive level and provide the framework for compensatory (or even substitutional) functions of hearing aids in terms of the attenuation component, the distortion component and the neural component of the hearing loss. A major factor is the requirement of a strong individualisation of hearing aid solutions calling for an appropriate assessment of the different sensorineural components of a hearing loss, especially with respect to bilateral and binaural hearing aid solutions.

Key Words: Hearing loss; models of hearing; compensation strategies; rehabilitation with hearing aids

Introduction

The objective of hearing aid technology is to support better hearing and speech communication in everyday life for individuals with various degrees of hearing impairment in our ageing, communication-driven society. Effective individualised hearing aids are, therefore, sought that rely on appropriate audiological diagnostic information, efficient fitting techniques and the ability to adapt to the respective acoustic situation and the individual situation-specific requirement by the user.

Hearing aid system technology has undergone a fascinating development in the past decades. After the introduction of the first commercial hearings aids with fully digital signal processing, it took a couple of years to take advantage of the possibilities of the new technology in terms of end user benefit. This is mirrored by MarkeTrak data showing that customer satisfaction with up-to-date hearing aid technology (devices not older than 4 years) has been stable on a level around 60% all over the 1990ies (Kochkin, 2010). But since 2004, 8 years after the advent of fully digital hearing aids, customer satisfaction has gone up steadily and has reached an impressive level of 85% in 2014 (Abrams & Kihm, 2015). However, as shown by Meister et al (2015) residual activity limitation is in the order of 40% for up-to-date hearing aid system technology. That means that there is still quite a bit of opportunity for future improvement.

The assessment of technology can be performed rather straight forward by assessing the degree to which the hearing aid fulfils its specifications under the assumption that the specifications aim at restoring the normal auditory function as closely as possible. Since this assumes a certain understanding of how the normal auditory system operates and how hearing impairment changes these functions (which can be operationalised by models of hearing and of hearing impairment), a model-based construction and assessment of hearing aid technology appears to be the best available guiding principle in the technical evaluation of hearing aids. Hence, the current Issue of the International Journal of Audiology focuses on the usage of auditory models in hearing aid system technology.

It is still open to debate whether a model-based approach to hearing aid functioning will be used in future developments of hearing aids especially given:

(a) the constraints that the real-life conditions a hearing aid has to cope with are outside the development and fabrication site and

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(b) the fundamental impossibility to put all properties of real life acoustics into the models that are used for construction, fitting and evaluation of hearing devices.

For the period of time that can be foreseen now, hearing aid development and evaluation will remain a mixture of half-valid assumptions, half-validated processing schemes and only coarsely fitted parameters to the individual user, resulting in an empirically optimised mix of practical and theoretical methods to produce and fit a hearing aid. Also, the evaluation conditions for hearing devices in the laboratory can only provide a very rough approximation to real life. A realistic evaluation should take into account the fitting method, the acclimatisation of the individual user, and some real-life-simulating conditions in order to produce a realistic mix of, e.g. dynamic range, spatial coherence and the complexity of the situations employed.

This paper, therefore, reviews the state-of-the-art of model-based hearing aid system technology considering both sides, i.e. the technical developments and the user-centred assessment of hearing device benefit achieved. Moreover, this review sets the stage for the current Issue of the International Journal of Audiology (IJA) with a collection of papers on both technical achievements and on the assessment with human listeners using the model-based approach outlined above as a guiding principle. The papers presented in this issue of IJA summarise the outcomes of a German multi-centre research project on model-based hearing aids. An attempt is made to give an overview on the state-of-the-art and to show the current limitations and the potential of the developments achieved so far.

The paper is organised as follows: as an introduction, the current state of the art of commercial hearing aid system technology is briefly reviewed. This is followed by and confronted with the state of current models for hearing impairment and principle compensation measures to be taken in hearing devices. Consequently, the final chapters describe the consequences from the models towards hearing aid functionality which will be analysed in its relation to future developments to be expected in the area of hearing aids.

State of the art of hearing aid system technology

Hardware

A systematic overview on current hearing aid features shall be given based on a solution-oriented classification according to Table 1. In the left column, hearing aid features are grouped in classes such as form factors, amplification schemes, listening programmes, noise reduction approaches and other features. In the next two columns, hearing aid features are specified in detail (column 2) along with their benefits (column 3). A closer look into the benefits shows that most hearing aid features may also be associated with counterproductive effects (column 4) in terms of costs, cosmetics or other aspects. Therefore, appropriate selection of hearing aid features is always a decision making process based on trade-offs driven by customers' needs and demands.

Form factors

Presently, the most common types of hearing aids are mini BTE devices either equipped with micro tube or external receiver in the ear canal (RIC: Receiver-in-Canal). Both types allow small housings and, in the case of RIC instruments, some extra high-frequency gain (when the receivers are equally-sized) due to the fact that there is no need for sound tubing acting as low pass filter. The low pass filter effect increases with decreasing sound tube diameter. However, smaller housings need smaller types of batteries associated with shorter battery lifetimes. Lithium-ion batteries presently introduced into the hearing aid market may change this situation.

All BTE solutions can be fitted as open as possible as long no feedback occurs. Open fittings avoid the occlusion effect resulting in better wearing comfort and natural sound quality as the ear canal is kept open and body conducted sound can drain-off the ear (Kiessling et al, 2005). Open acoustics can be achieved either by open standard domes as well as by open individual ear moulds. From the acoustic point of view, there is no difference between both solutions, but it is the user preference making the difference: some users prefer individual ear moulds over standard domes as they provide better retention in the ear canal, others decide the other way around for other reasons, such as wearing comfort and weight.

Acoustic feedback suppression management will be discussed later, but in this context, it has to be mentioned that it was the advent of efficient feedback suppression that has made totally open fittings possible. Modern feedback suppression schemes have increased the feedback margin by 15–20 dB. That means that about 70% of all hearing aid candidates can benefit from open fittings today. The transparent sound experience provided by open fittings is appreciated by many hearing aid users. But at the same time, it has to be taken into account that unprocessed direct sound through the vent compromises the efficiency of all signal-processing schemes, e.g. noise reduction, directivity, etc.

In recent years, the market share of ear canal devices has been down to about 10%. But, market share is slowly increasing again as modern technologies allow the building of extremely small instruments to be deeply placed in the bony part of the ear canal. These so-called IIC devices (Invisible-in-Canal) are totally invisible and do not occlude the cartilaginous part of the ear canal so that body conducted sound can drain-off the ear canal. Therefore, IIC devices offer natural sound quality similar to totally open BTE fittings contrary to hearing instruments sitting in the concha or in the cartilaginous part of the ear canal.

Amplification

Amplification is the fundamental concept of hearing aid rehabilitation aiming on the audibility of relevant sound components, as audibility is a necessary – but not a fully sufficient – requirement for better speech intelligibility. Linear amplification providing the same amount of gain at all input levels is appropriate for all hearing impairments not being associated with dynamic range reduction, such as conduction losses or pure inner hair cell loss resulting in the rare cases of sensorineural impairment without recruitment (cf. section Frequency-specific amplification, feedback and acoustic distortion reduction).

As the majority of hearing aid candidates, however, do have sensorineural impairments with reduced dynamic range, predominantly due to outer hair cell loss, it needs non-linear amplification (i.e. compression) to amplify soft sounds with higher gains compared to loud sounds (cf. section Frequency lowering). There are different types of non-linear amplification schemes (Kates, 2005) which can be classified according to different criteria (slow/fast, input-/output controlled, single-/multi-channel compression). Modern non-linear amplification schemes are typically implemented as multi-channel wide dynamic range compression (WDRC) systems to allow frequency-specific restoration of the

Table 1. Solution-oriented overview of current hearing features: benefits and counterproductive aspects.

Classes of solutions	*Hearing aid features*	*Benefits in terms of . . .*	*At the expense of . . .*
Types/form factors with/ without remote control/ smartphone app	Mini-BTEs with micro tube	Cosmetics/size	Size of battery => costs Reduced high-frequency gain
	Mini-BTEs with RIC (Receiver-In-Canal)	Cosmetics/size and extended band width	Size of battery => costs Higher risk of receiver repair => costs
	Ear canal devices (CIC: Completely-In-Canal, IIC: Invisible-In-Canal)	Cosmetics/size IIC: invisibility	Size of battery => costs Occlusion effect (no occlusion in IIC)
	Open fitting	Sound quality and wearing comfort	Signal processing less efficient
Amplification schemes	Linear amplification	Audibility	–
	Non-linear amplification	Audibility and restoration of dynamic range	–
	Frequency lowering	Preservation of high-frequency information	–
Listening programmes	Manual environment-related programme selection	Optimisation of parameter settings	Visibility due to manual control
	Automatic environment-related programme selection based on environment classification	More reliable optimisation of parameter settings	Automatic programme selection possibly not in accordance with users demands
	Automatic optimisation of parameter settings based on environment classification	Improved and more reliable optimisation of parameter settings	Automatic parameter settings possibly not in accordance with users demands
Noise reduction schemes	Single-microphone solutions: noise reduction, impulse noise management, etc.	Ease of listening	SNR improvement only at the expense of artefacts
	Multi-microphone solutions: directional microphones, microphone arrays, beamformer	SNR improvement	Size of housing => Cosmetics
	Induction/telephone coil	SNR improvement	Size of housing => Cosmetics
	Wireless solutions: external microphones, audio streaming (with/ without streamer) from smartphone, telephone, TV, HiFi etc.	SNR improvement	Extra equipment => cosmetics, costs Energy consumption => costs
Other solutions	Feedback suppression	Availability of more stable gain	—
	Ear-to-ear solutions: left/right synchronisation, audio streaming (telephone use, wind, CROS/ BICROS etc.)	SNR improvement (telephone, wind, etc.) Restoration/enhancement of binaural differences	Energy consumption => costs
	Datalogging, self-learning systems	Counselling, fine-tuning, automatic settings	Energy consumption => costs
	Sound generators	Tinnitus treatment, general relaxation	Energy consumption => costs

dynamic range. The user benefit of WDRC systems has been investigated in a meta study by McCreery et al (2012) indicating that WDRC systems improve audibility over linear amplification. In many conditions, this leads to an improvement of speech recognition. In other conditions, speech recognition was never reduced by WDRC. Subjective rating of WDRC systems strongly depended on the individual user profile and the type of acoustic environment (McCreery et al, 2012).

In cases of total loss of inner hair cells in certain areas of the basilar membrane, the so-called ''Cochlear Dead Regions (CDR)'', information transfer from a non-functioning to a functioning frequency range can be taken into consideration (cf. section Frequency lowering). As CDRs are mostly located in the high frequencies, frequency-lowering algorithms are available in many modern hearing aid systems (Kluk & Moore, 2006). For this purpose, a couple of frequency-lowering schemes, such as frequency compression and frequency transposition, have been developed. The benefit of frequency lowering is discussed controversially in the literature. So the ratings reach from ''not useful'' (Robinson et al, 2009) to ''useful, if indicated''(Leifholz et al, 2013). A meta study on this topic based on five single studies indicates that frequency lowering may be useful, but for a more reliable rating. further studies are needed (McCreery et al, 2012). If a broader range in the high frequencies is affected by inner hair loss, then frequency lowering is not an option anymore. Then, electro-acoustic stimulation (EAS), i.e. electric stimulation (cochlea implant) in the high frequencies and acoustic amplification (hearing aid) in the lows, have to be considered (von Ilberg et al, 2011).

Listening programmes

For the optimisation of hearing aid parameter settings (gain, frequency response, compression, noise reduction, directivity, etc.) in different acoustic environments, there are three types of listening programmes available today. A common approach is the implementation of a couple of listening programmes, typically 2–4, for different acoustic conditions, such as speech in quiet, speech in noise, music and telephone. In the most basic version, listening programmes are manually selected by the user either using a toggle switch or button on the hearing instrument or a separate remote control unit. Some manufacturers offer smartphone apps, both for IOS and Android devices, to enable smartphones being used as remote control units.

As studies have shown that manual listening programmes are frequently selected inappropriately by the user (Kiessling et al, 2007), automatic programme selection has been developed to solve this problem. In this case, hearing aids works as acoustic analysers and as situation classifiers to make appropriate programme selections. These systems are functioning quite reliable today, but it still can occur that automatic selection is not in accordance with the demands and preferences of the user. Clinical experience shows that the decision for a manual or an automatic system should be made according to the personality of the user: proper profiling of the personal demands reveals whether the user wants to be in the driver's seat or prefers to get rid of manual switching.

Finally, there is a third level of performance optimisation based on multiple parameter optimisation for each and every listening situation instead of working with a limited number of fixed parameter settings, i.e. listening programmes. On top of this, some modern hearing instruments offer a kind of GPS-based geotagging. This means that locations that are frequently visited by the user can be assigned with appropriate parameter settings (programmes) and the system will automatically switch to this setting, once the hearing aid has identified one of the individually tagged places. This strategy, however, will work appropriately only in places where the acoustic environment is always the same.

Noise reduction schemes

A number of single-microphone, i.e. monaural, solutions are available in hearing instruments today (cf. section (Monaural) noise reduction). If the spectrum of the background noise is not identical with the spectrum of the target sound, mostly speech, an improvement of the signal-to noise ratio (SNR) can be achieved by simple filtering, for instance high-pass filtering in situations with traffic noise. Unfortunately, these listening situations are rare. Particularly, in listening environments, when masking noise happens to be of speech type, simple filtering does not help to improve speech intelligibility.

In listening conditions with stationary maskers, background noise can be separated from speech based on the fact that speech signals are typically characterised by 4 Hz modulations. So the modulation spectrum of the input signal can be continuously analysed in each frequency band. If the signal is dominated by speech-like modulations this band will be amplified, if no speech modulation is detected this band will be given no or less gain. This strategy works very well in principle, but frequently, the spectra of speech and background noise are overlapping to a great extent, i.e. there is a similar mix of speech and noise in all relevant frequency bands and therefore no significant improvement of SNR can be achieved (McCreery et al, 2012). Anyway, modulation-based solutions like this improve the ease of listening which is perceived and appreciated by the users.

Another single-microphone approach in commercial hearing aid systems is based on spectral subtraction. In a first step, speech pauses have to be detected reliably to receive a good estimate of the ''pure'' noise signal. Then, the estimated noise signal is subtracted from the noisy input signal resulting in an improved SNR. Noise estimation is administered continuously to adapt the system to changes of the acoustic situation. More sophisticated noise estimation methods employ special signal features (like the amplitude modulation spectrogram) and machine learning methods (Tchorz & Kollmeier, 2003; Healy et al, 2015) that are not yet available in commercial hearing instruments. The concept of spectral subtraction is actually a subtype of a Wiener filter (Van den Bogaert et al, 2009; Cornelis et al, 2012), as Wiener filtering works with both an estimate of the noise (based on the signal in the speech pauses) and the speech signal (based on the signal in the speech sections). Real binaural signal processing algorithms as presented in this special issue as well as other advanced solutions are not yet available in commercial hearing instruments but are under investigation in research hearing aids.

More efficient than single-microphone solutions are multi-microphone systems in real listening situations (cf. section Directional microphones and beamformers; Beck & Schum, 2006). The concept of SNR improvement by multi-microphone systems is based on the assumption that noise and speech signals come from different directions. This applies for instance in cocktail party situations when the hearing aid users communicate with a single person or a small group of persons in front of him/her in a room with diffuse speech babble around. In listening conditions like this, directional microphones with frontal directivity promise significant SNR improvement. The most simple implementation of a directional microphone is the coupling of two microphones and time-delayed addition of both input signals. Depending on the distance between both microphone ports and choice of the internal time delay different types of directivity pattern (bidirectional, cardioid, hyper-cardioid, super-cardioid) can be realised. Stronger directivity can be achieved by wireless linking of the two microphones of each side (left/right) to form an array of four microphones (Kreikemeier et al, 2013). More directivity is obtained by microphone arrays of higher order, for example, an array of eight microphones being placed in a pair of glasses (Soede et al, 1993; Luts et al, 2004). In addition, adaptive beamforming is available in commercial hearing instruments today. Adaptive beamformers are able to adjust their directivity also to lateral targets or speakers in the back and provide adaptive attenuation of moving noise sources. A literature research on the efficiency of directional microphones and beamformers has shown on a moderate evidence level that directional systems may improve speech intelligibility in particular acoustic situations (McCreery et al, 2012).

In special listening situations (distant speakers, TV, CD/MP3 players, smartphones, cell phones, etc.), an even better SNR improvement can be achieved by direct audio streaming through induction loops or product-specific wireless systems. O one hand, inductive solutions are cheaper than wireless equipment and can be used in a universal, manufacturer-independent way. On the other hand, modern wireless systems provide more reliability and better sound quality free of artefacts or cross talk effects.

The advantages and technical possibilities of audio streaming to hearing aids are illustrated in Figure 1. In conversations on

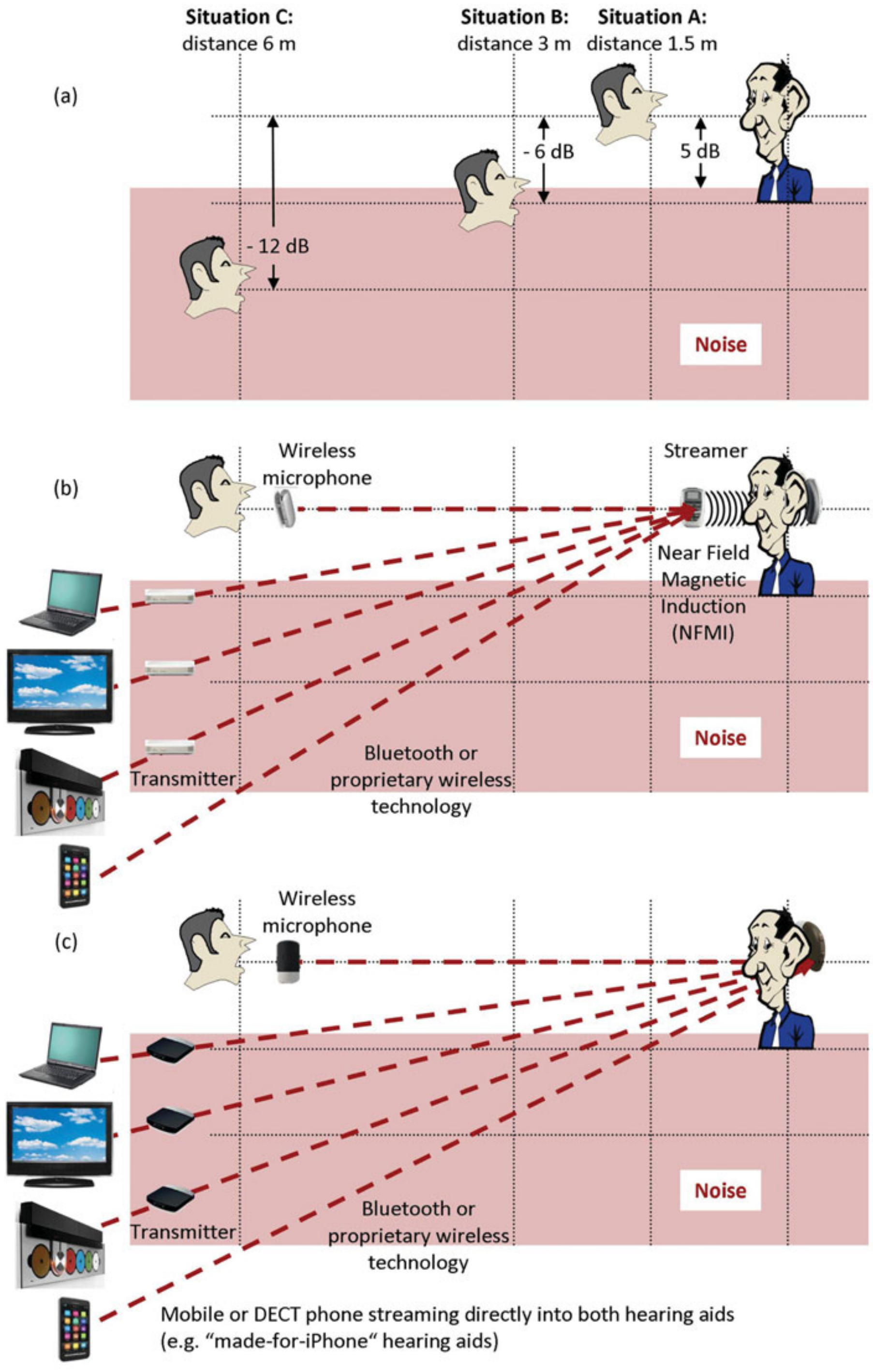

Figure 1. Unaided communication in noise: with increasing distance the speaker disappears in the noise floor (a). Two types of wireless solutions for hearing aids (b and c).

normal social distance, the speaker adjusts his/her voice level typically to about +5 dB SNR (Figure 1a, situation A). Under free field conditions with each doubling of the distance between speaker and listener SNR decreases by 6 dB, in real-life situations, the attenuation is less. But in many listening situations, the voice of the speaker perishes in the noise floor (Figure 1a, situation C).

Currently, there are two types of wireless systems on the market to improve this situation. Some manufactures use Bluetooth or proprietary wireless connection from external microphones and other external sound sources (via transmitter) to a body-worn streamer at the listeners end (Figure 1b) and the streamer sends the signal to the hearing aids by Near Field Magnetic Induction (NFMI).

The other type of products provides direct audio streaming by Bluetooth or other wireless technology in the GHz range from external sound sources (via transmitter) into the hearing aids (Figure 1c). This solution has the advantage that there is no need for an extra streamer device but the power consumption of those hearing aids is much higher than for the technology depicted in Figure 1b, i.e. watching TV for a couple of hours per day will drain the batteries very soon.

Smartphones, both IOS- or Android-based systems, can be directly connected to many hearing aid models today (e.g. ''made-for-iPhone'' hearing aids). This allows phone calls or streaming of other audio signals (music, audiobooks, etc.) at a favourable SNR and perfect sound quality. Some manufacturers offer DECT telephones especially developed for their own hearing instruments. These DECT phones connect themselves automatically to the hearings aids once the phone has been lifted up and brought close (about 50 cm) to the hearing aids. All wireless solutions have proven to be very helpful for hearing aid users (Thibodeau, 2010; Picou & Ricketts, 2011; Picou & Ricketts, 2013). However, their use is limited to special listening situations as described above and, up till now, they work only with the hearing aid models of a single manufacturer. Recently, the major hearing aid manufacturers have agreed to use a standardised 2.4 GHz wireless protocol to achieve compatibility between wireless accessories and hearing aids of all manufacturers.

Other features

In the class of miscellaneous features, feedback suppression schemes seem to be the most important ones. Early feedback reduction strategies worked with notch filters or other types of gain reduction to prevent acoustic feedback. Modern sophisticated approaches analyse the feedback path and add an antiphasic signal to the input signal to increase the feedback margin. Sometimes this strategy is combined with a minor frequency shift between input and output signal to make the system even more stable. Real binaural feedback suppression schemes promise extra feedback margin of up to 20 dB (Grimm et al, 2009). The efficiency of modern feedback suppression algorithms has laid the foundation for the success of open fittings in the last couple of years. Today about 70% of all hearing aid candidates can benefit from open fittings and future technological progress will increase this proportion.

Beyond the features described above wireless technology has made possible several ear-to-ear applications. The most basic one is synchronisation of volume control and programme use between left and right hearing instruments. Another ear-to-ear solution makes sure that natural interaural level differences are restored or enhanced if needed. Some models facilitate telephone calls by streaming the telephone signal to the opposite ear which is beneficial for many users, as making phone calls is one of the major problems reported by hearing-impaired persons. Other ear-to-ear features improve ease of listening in windy environments by copying the input signal from the ear in the wind shadow to the opposite ear exposed to the wind. Wireless audio streaming is also used to implement CROS and BICROS solutions with perfect sound quality.

Datalogging features save relevant usage data such as use time, volume setting, use of listening programmes, etc. in the hearing instruments. These data can be read out by the clinician or audiologist at the next visit and used for counselling and fine-tuning purposes. In more sophisticated (Grimm et al, 2009) implementations, called self-learning devices, saved usage data can be used for automatic fine-tuning by the hearing aid system itself.

Finally, many hearing instruments are equipped with a sound generator for tinnitus treatment or just for relaxation purposes. These models can be programmed either as noise generators only, hearing aids only or tinnitus instruments (hearing aid plus noise generator). The sounds available can be of noise type (white/pink/individually shaped, stationary/modulated), natural sounds (wind, ocean waves, rain, etc.) or of other complex type (e.g. fractal tones for Zen therapy, etc.). Some models allow environmental volume steering meaning that the volume of the therapeutic noise is automatically increased in quiet and turned down in loud environments. Beyond this, various other parameters can be programmed by the clinician and fitted according to the needs of the customer.

Fitting software

Major functions

The process of hearing instrument fitting can be structured according to the following steps:

(1) *Profiling*: Assessment of user needs, demands and preferences by questionnaires and interviews.
(2) *Audiometry*: Unaided measurements like pure tone audiometry PTA, speech audiometry, loudness scaling, etc.
(3) Selection of a set (3–4) of appropriate hearing aid models and type of acoustic coupling (individual earmold/standard ear tip, degree of openness).
(4) Measurement of feedback margin (loop gain).
(5) *Basic fitting*: Initial fit with prescriptive fitting rules:
    (a) Generic PTA-based fitting formulas: NAL-NL2, DSL v5.
    (b) Proprietary PTA-based fitting formulas: manufacturer-specific rules.
    (c) Loudness-based approaches.
(6) *Fine-tuning*: presentation of everyday life sound samples and translation of user complaints into optimised hearing aid settings, either experience-based or by means of a software feature (automatic fine-tuning assistant).
(7) *Programming of optional settings*: functionality of listening programmes, manual controls, warning tones, frequency lowering, datalogging, self-learning/acclimatisation manager, internal sound generator for tinnitus treatment, etc.
(8) *Evaluation of the fit*
    (a) *Verification*: matching aided performance with the targets (coupler/probe microphone measurements, threshold measurement, loudness scaling etc.)
    (b) *Validation*: assessment of benefit (speech audiometry in quiet/noise, questionnaires, diaries, etc.)
(9) Gliding fitting during the acclimatisation process

On all levels process loops may be needed.

Generally, process steps 3–7 are administered by means of the fitting software. Hearing aid programming can be carried out either via a universal hardware interface (Hi-Pro) connecting the hearing aids to the fitting system computer or by wireless connection.

Most of the hearing aid manufacturers offer *in situ* audiometry modules (step 2) as a part of their fitting software, i.e. PTA can be performed through the hearing aid taking properties of the acoustic coupling into account. Fitting software packages can be used either as standalone versions or under NOAH (www.himsa.com), an industry-standard for data storage and management. NOAH handles all data emerging from the fitting process and allows us to connect the fitting system to audiometers and other measurement equipment, e.g. coupler and real ear measurement systems, making all measurement data available for the fitting process and for future use. When working under NOAH hearing aid verification (step 8a) can also be administered within the fitting software environment. A questionnaire module enables NOAH users to administer frequently used questionnaires (COSI, APHAB), store and retrieve data (step 1).

Whereas most of the process stages specified above are straightforward and mainly driven by the hardware available, the process of basic fitting has been is discussed controversially for many years. Therefore, this paper will focus on state-of-the-art of basic fitting strategies. At this point, however, it has to be emphasised that basic hearing aid fitting by means of prescriptive formulas is just one single step which has to be imbedded in a comprehensive procedure as described above.

Basic fitting with prescriptive rules

Basic hearing aid fitting is meant to provide a starting point for further optimisation of parameter settings. As early research (Pascoe, 1988) has shown a correlation between hearing loss data and the sound level being perceived as most comfortable (MCL), historically all basic fitting rules were built on the client's pure tone hearing loss. Starting from the half-gain rule (Lybarger, 1963; Berger et al, 1980), over POGO (McCandless & Lyregaard, 1983) to more sophisticated approaches like NAL for linear instruments (Byrne & Dillon, 1986) target frequency responses have been estimated by PTA based fitting formulas. Once non-linear amplification schemes were available, it became clear that simple threshold based rules cannot be used anymore as they were designed for linear amplification.

At that time, the development of fitting procedures split-up into two different approaches: on the one hand, it has been argued that individual dynamic ranges of the hearing aid users should be taken into account rather than empirical experience based on average data. So loudness-based fitting procedures have been developed and tested (Allen et al, 1990; Kiessling et al, 1996; Pastoors et al, 2001; Moore et al, 2005; Moore & Fullgrabe, 2010; Moore & Sek, 2012). On the other hand, other authors pursued the argument that simple threshold testing is more reliable and less time-consuming compared with loudness estimation or scaling. They developed more sophisticated threshold-based fitting rules for non-linear hearing aid systems such as DSL [i/o] (Cornelisse et al, 1995) or NAL-NL1 (Byrne et al, 2001).

As it could not be demonstrated yet that loudness-based fittings may come closer or quicker to the individual final fit (Elberling, 1999; Pastoors et al, 2001; Wesselkamp et al, 2001) and as threshold measurement is a simple, quick, reliable and worldwide available procedure, threshold-based fitting formulas are dominating the market today. Two types of threshold-based prescription rules have emerged: (1) universal, generic prescription formulas that can be applied to all makes and models and (2) proprietary formulas developed by hearing aid manufacturers especially for their own hearing aid systems. Different from the first generation of threshold-based rules, fitting formulas for non-linear devices prescribe target gain curves for a set of input levels, typically for soft, comfortable, and loud sounds.

Generic PTA-based fitting formulas. Two major families of generic PTA based fitting formulas are available in the fitting software of most manufacturers: the NAL type formulas developed by the National Acoustics Laboratories (Australia) and the DSL family (Desired Sensation Level) developed at the University of Ontario (Canada) (Johnson, 2012; Ching et al, 2013). The latest version of these fitting families are NAL-NL2 (Keidser et al, 2011, 2012) and DSL [i/o] v5 (Scollie et al, 2005; Polonenko et al, 2010). The rationales behind them are summarised in Table 2.

A comparison of frequency responses of NAL-NL2 and DSL v5 shows that NAL-NL2 prescribes more low-frequency gain up to 750 Hz, whereas DSL v5 provides more high-frequency gain above 1000 Hz than NAL-NL2, both at all input levels. An excellent compare and contrast peer-reviewed article has been published by Johnson (Johnson, 2013). Recent studies came to the conclusion that generic fitting formula provide a solid ground for a basic fit to be used as starting point for individual fine-tuning (Abrams et al, 2012; Sanders et al, 2015). At the same time, it has been found that the implementation of generic fitting rules considerably vary from manufacturer to manufacturer, i.e. each manufacturer may come up with different frequency responses for the same prescription formula (Abrams et al, 2012; Leavitt & Flexer, 2012; Sanders et al, 2015). Therefore, real ear measurement (REM) verification of frequency responses is strongly recommended to make sure that the hearing aids are actually fit to target (Sanders et al, 2015).

Proprietary PTA-based fitting formulas. Besides generic fitting formulas, product-specific prescription (Keidser et al, 2003; Mueller, 2005; Abrams et al, 2012; Powers et al, 2014) rules are offered in the fitting software of the hearing aid manufacturers and have gained broad market penetration. These proprietary fitting formulas have the advantage that product-specific features can be taken into account. On the contrary, compared with generic approaches their aims and rationales are not well defined or sometimes not even known and they are frequently not validated systematically. In general, proprietary fitting rules prescribe less gain than generic formulas as manufacturers try to avoid that the initial acceptance of hearing aids may be compromised by providing too much gain. Fitting strategies aiming primarily on spontaneous acceptance definitely need fine-tuning in terms of stepwise enhancement of gain over the acclimatisation period to make sure that hearing aid users eventually end-up with sufficient gain and audibility to optimise speech intelligibility. In practice, however, the fine-tuning process is frequently not pursued appropriately that means that the user will stay with a suboptimal fit.

Models of the effects of hearing impairment on auditory functions

Models relating to hearing impairment and its potential compensation by hearing aids aim at collecting and ordering the available knowledge in a way that is – among others – useful for the treatment of hearing impairment with hearing aids. Numerous models for predicting psychoacoustical effects, speech perception and audio quality have been developed in the past. Under the assumption that

Table 2. Generic versus proprietary fitting rules: rationales and properties.

Generic rules		*Proprietary rules*
- Are based on known rationales - Are appropriately validated - Generally, provide more gain, i.e. come closer to optimised settings in terms of speech intelligibility - Are sometimes not exactly implemented in the fitting software packages of the manufacturers, i.e. manufacturers may prescribe different targets for the same hearing loss		- Are mostly based on unknown or not well-defined rationales - Are not or less validated - Provide less gain, i.e. come closer to optimised settings in terms of initial acceptance and feedback avoidance
NAL-NL2	DSL v5	
- Equalizes loudness in the mid frequencies and limits overall loudness of speech at or below normal - maximises effective audibility, i.e. maximises speech intelligibility for each input level - Does not allow for individual loudness discomfort levels - Takes gender and level of experience into account - Offers different formulas for children and adults	- Normalizes loudness for each frequency and each input level, i.e. restores audibility - Avoids loudness discomfort - Allows for individual loudness discomfort levels - Offers different formulas for infants, children and adults	

such perception models represent the most relevant aspects of auditory signal processing in the normal and hearing-impaired auditory system, they can eventually be used for fitting hearing systems to individual users or/and specific use cases and potentially for steering them (‘‘model in the loop’’). Moreover, instrumental methods have been developed on their basis as tools for the objective assessment of audio systems (e.g. audio reproduction systems, algorithms, and hearing devices) that can be used during development and optimisation.[1]

In general, perceptual methods share fundamental auditory processing mechanisms like frequency decomposition and temporal integration, similar to and motivated by effective auditory models such as Dau et al (1997) and Derleth et al (2001). Recently, more physiologically motivated stages were introduced (Jepsen et al, 2008) and adapted to individual hearing loss (Jepsen & Dau, 2011). Figure 2 shows the generic layout of such ‘‘effective’’ models: an important element of such a functional description is a filter bank as a first stage that distributes the sound according to its frequency into different frequency bands and hence mimics some aspects of the basilar membrane. On one hand, it may be preceded by a broadband pass filter that mimics the ‘‘effective’’ frequency shaping by the middle ear. In each frequency channel, the instantaneous energy is then obtained by extracting the temporal envelope which can be modelled by a half-wave rectifier with successive low-pass filtering. This process and the subsequent adaptation stage should simulate the function of the hair cell and the auditory nerve where the respective sensitivity is actually adapted to the average of the respective input signal and the exact fine structure of the input signal is lost for high frequencies of the signal. The adaptation is followed by a binaural comparison as well as a separation into different modulation frequencies by a modulation filter bank. Hence, the output of these basic processing structures can be thought of as a two-dimensional pattern for each point in time that represents audio frequency on one axis and modulation frequency on the other hand. In addition, for each combination of modulation frequency and centre frequency, a binaural, interaural relation can be assumed to be present (cf. Kollmeier & Koch, 1994).

The time course of this multidimensional pattern can be thought of as representing the ‘‘internal representation’’ of an acoustical stimulus presented to the auditory system. To account for processing inaccuracies of the auditory system, an additive processing noise (‘‘internal noise’’) has to be added. Hence, most models of the ‘‘effective’’ signal processing in the auditory system assume that these most important processing steps of the auditory system can be represented with comparatively simple processing circuits that require very few parameters for their functioning. The actual detection of any signals in noise or of any changes in the acoustic input signal can be modelled then by an optimum detector or matched filter (cf. Green & Swets, 1966; Dau et al, 1997). In other words, the current models assume that auditory detection and discrimination is not necessarily limited by the cognitive information processing, or by the previous knowledge about the stimulus and the training in the auditory task, but rather by any inaccuracies in representing the external acoustical stimulus in the internal representation

Such a model can also be used to characterise the effect of hearing impairment which may be caused a by a number of different ‘‘primary’’ factors (i.e. effects with a defined physical or physiological cause) that influence the functioning of the whole auditory system in a complex and interrelated way which is not yet clear. At least, four major factors can be characterised (indicated by red stars in the figure):

(1) *Attenuation loss*: The ‘‘classical’’ view of hearing loss as a simple attenuation of the input sound, caused, e.g. by conductive hearing loss or a loss of sensitivity due to a pure loss of the inner hair cells.
(2) *Compression loss*: The loss of input dynamic range of the auditory system usually associated with a loss of outer hair cell functioning resulting in the recruitment phenomenon. Note that this reduction or even failure of ‘‘active processes’’ in the inner ear causes many ‘‘secondary’’ effects (such as, e.g. effective flattening of frequency tuning, broader auditory filter bandwidths and non-linear distortion)

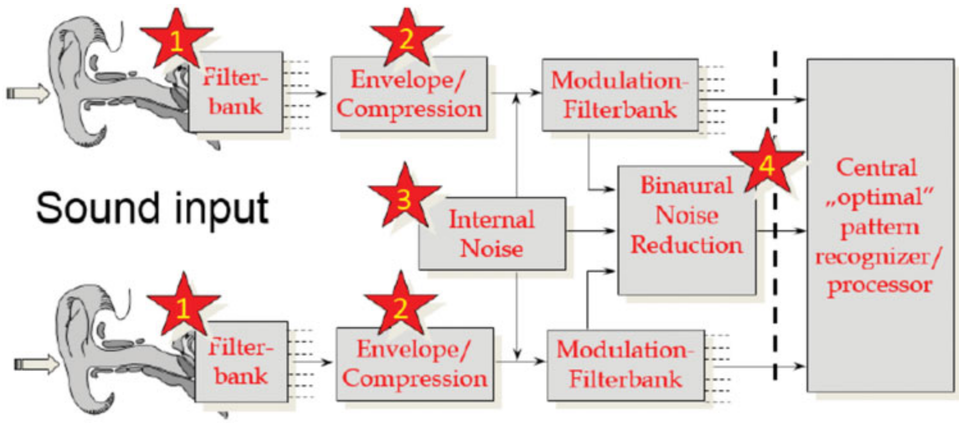

Figure 2. Generic model of the ''effective'' signal processing in the auditory system and possible places where degradations in the signal processing occur due to hearing impairment (marked by stars 1...4). The dashed vertical line indicates the ''internal representation'' of the acoustic input signal.

that cannot be completely separated from a pure compression loss.

(3) *Resolution loss*: An increased level of ''internal noise'' is associated with a central hearing loss that disables the effective readout of the ''internal representation''. Such a hearing loss leads, e.g. to an increased intensity discrimination threshold in hearing-impaired listeners characterised by this factor. Pathologies in hearing impairment such as deteriorated transmission of temporal fine structure to the central auditory system or a ''hidden hearing loss'' connected with a loss of afferent auditory nerve fibres (Kujawa & Liberman, 2009) may contribute to this factor.

(4) *Binaural loss*: The ability of the auditory system to suppress an ''undesired'' sound source emanating from a certain angle in space is largely determined by binaural interaction across both ears. This allows, e.g. for spatial listening, localisation and reverberation suppression in rooms and the so-called ''cocktail party phenomenon'', i.e. the ability to concentrate on a ''desired'' speaker in a lively party and to suppress any undesired sound sources. A specific deficit of this binaural interaction can be a consequence, e.g. of an asymmetric hearing loss or non-linear distortion in the processing of the stimulus-evoked neural activity on either side of the head.

Even though the current status of such models of the signal processing in the auditory system does not allow a very detailed description of the auditory system, it appears that a technically oriented model of interpreting the action of the auditory system holds promise in order to better ''understand'' the auditory system and to derive applications from it (see Kollmeier, 1999, for a review).

The model framework given above allows for a hierarchical modelling of acoustic information processing (and its various impairments) at different levels of this processing: the *acoustical level* (covering the influence of acoustic quality degradations on the acoustic input to the ear), the *(neuro-) sensory level* (covering cochlear and retrocochlear processing up to the primary auditory cortex), and the *cognitive processing level* covering effects of familiarity and training with the stimulus, working memory and attention deficits.

Acoustical level

Existing speech intelligibility models have primarily considered the *acoustical level* (in terms of effective signal-to-noise ratio across auditory filter bands) to quantitatively evaluate the impact of disturbing factors, especially reverberation and noise, on speech intelligibility. State-of-the-art models go clearly beyond current standards such as SII (ANSI, 1997) and STI (Houtgast & Steeneken, 1985) by, e.g. accounting for binaural hearing (van Wijngaarden & Drullman, 2008); (Leclere et al, 2015; Beutelmann et al, 2010; Rennies et al, 2011), amplitude modulation masking (Jorgensen et al, 2013). Even though the SII and STI models can account for individual hearing thresholds (Horwitz et al, 2007), the prediction accuracy for assessing the effect of hearing impairment primarily by accounting for the increased threshold in quiet and a distortion/desensitisation factor for high-sound levels is limited. This is primarily because neurosensory and cognitive factors that vary across listeners are not accounted for (see below).

Neurosensory level

The *neurosensory level* for modelling speech intelligibility, on the contrary, has been considered by ''microscopic'' models that use an ''effective'' signal-processing model as a front end to derive auditory-system-inspired speech features to be employed by a backend derived from Automatic Speech Recognition (ASR). (Holube & Kollmeier, 1996) and (Jurgens et al, 2014) used the perception model by Dau et al (1996) as a front end and a simple Dynamic Time Warping Algorithm (DTW) as an ASR backend, whereas Meyer et al (2011) and Schadler et al (2015) employed a mel-frequency-scaled-cepstral-coefficient (MFCC)-based feature set (which is typically employed in ASR applications while only roughly representing auditory signal processing) in combination with a Hidden Markov Model (HMM) backend. This approach can be extended to include the attenuation loss effect and a more central hearing loss component (represented by an internal level uncertainty or multiplicative noise) to increase the prediction accuracy for SRT data in 198 ears with different degrees of hearing impairment (see Figure 3 taken from Schädler et al, 2016). Hence, by taking an estimate of the individual central distortion component into account, a correlation coefficient of 0.85 between predicted and empirical

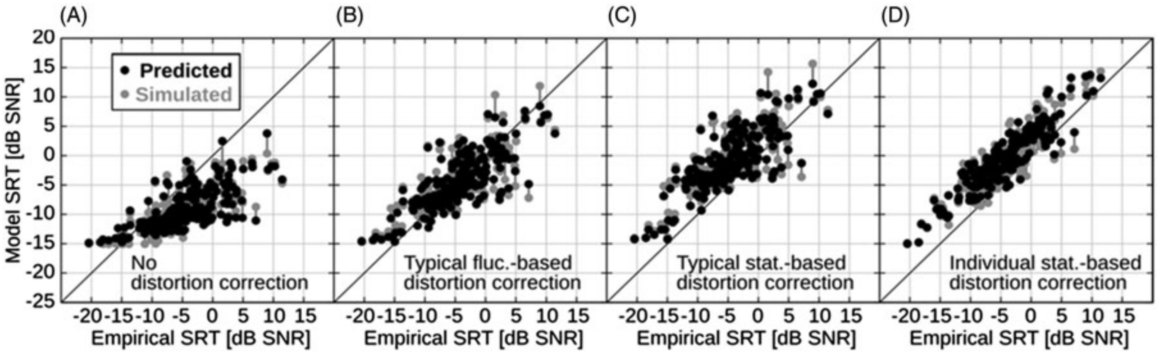

Figure 3. (From Kollmeier et al., 2016): Modelled Speech Recognition Threshold (SRT) for 198 ears from 99 test persons plotted against the empirical data (*x*-axis) for fluctuating noise. The predicted SRT-values (employing typical audiograms) are given as black dots, and the simulated data (employing the individual audiogram) are given as grey dots. The simulations are performed for different degree of distortion correction: No distortion correction (panel A), stationary-noise-, typical-audiogram-based distortion correction (panel B), fluctuating-noise-, typical-audiogram-based distortion correction (panel C), stationary noise-based, individual distortion correction (panel D).

SRT values could be reached which is clearly superior to any prediction approach based on the *acoustical level* only (e.g. a short-term SII) approach by Meyer and Brand (2013) who achieved a maximum of 0.52 for the same empirical data or similar approaches by Rhebergen et al (2006). Note that the SII prediction accuracy might be increased by including an individual estimate of distortion in addition to the hearing threshold as well, but such a distortion has to be independent of stimulus level and is clearly different from the desensitisation factor of the AI and SII that sets in at high levels.

Another important aspect of modelling (impaired) speech perception on the neurosensory level is reflected by *binaural processing*, i.e. the human ability to utilise the input from both ears to effectively improve the internal signal-to-noise ratio and hence to increase speech intelligibility. Such an improvement can occur if the target speech exhibits another interaural relation than the masker, for example, if the target speech comes from the front and arrives at both ears simultaneously without an interaural level difference, whereas the masker comes from the side and exhibits an interaural difference both in time and in level. Even though a large body of the literature exists on binaural signal processing and binaural models primarily for normal listeners (see, e.g. Colburn & Durlach, 1978; Durlach et al, 1981; Blauert, 1983), only few model approaches have been suggested so far to quantitatively describe the effect of binaural interaction on improving speech intelligibility which is equivalent to lowering the speech reception threshold in binaural conditions (vom Hövel, 1984; Zurek, 1990). Moreover, the effect of hearing impairment on this binaural advantage has been described (Beutelmann & Brand, 2006; Beutelmann et al, 2010; Lavandier & Culling, 2010).

All these models utilise a comparatively simple binaural circuit, the *equalisation and cancellation* (*EC*)*-mechanism* according to Durlach (1963) as the central binaural noise reduction mechanism: in the equalisation (E) stage, the noise (or in some implementations the complete mixture of signal and noise) is matched on both binaural channels by an appropriate interaural amplification (factor alpha) and an interaural time shift (parameter tau). In the subsequent cancellation (C) step, both channels are subtracted from each other, thus yielding a more or less complete elimination of the background noise (especially, if the same noise is present at both ears) and an amplification of the target signal. The improvement in signal-to-noise ratio and, equivalently in SRT is dependent on the binaural configuration. It is limited by some statistical fluctuations of the parameters alpha and tau. These statistical inaccuracies effectively form some kind of residual noise at the central/neural level according to (vom Hövel, 1984). The combination of the EC mechanism and the speech intelligibility prediction method (such as, the articulation index or the speech intelligibility index) can then be used to predict SRT values for different binaural conditions in normal listeners (vom Hövel, 1984; Zurek, 1990; Beutelmann & Brand, 2006).

Figure 4 shows a schematic diagram of the model version by Beutelmann et al (2010) and Figure 5 shows the prediction results for babble noise in different spatial environments in comparison with empirical data for normal and hearing-impaired listeners. Hearing loss has simply been modelled by an increased peripheral, internal noise, uncorrelated at both ears, simulating the auditory threshold. Overall, this kind of models provides a remarkably good prediction of the binaural noise reduction effect across a number of conditions. This includes complex acoustical conditions such as several interfering noise sources and reverberation (Warzybok et al, 2013; Rennies et al, 2011), even though a very rough functional description of the binaural noise reduction processes is employed by the EC mechanism. Interestingly, most of the attempts to model the effect of hearing impairment on binaural hearing have primarily required to model the attenuation component of the hearing loss (i.e. an internal peripheral noise), whereas a specific ''binaural hearing loss'' (assuming that a specific binaural processing deficit is present) has not been clearly identified so far.

The neurosensory level has also been considered for the modelling and prediction of perceived audio quality for normal and hearing-impaired listeners with and without hearing devices. Existing perceptual quality models have been mainly applied for assessing the transmission quality of lossy audio systems (e.g. audio codecs such as mp3) which introduce distortions. State-of-the-art models are ''double-ended'', i.e. based on the comparison of input and output signals of the system under test. The input signal (or its internal representation obtained by an auditory model) serves as a reference, defining the optimal quality (e.g. Huber & Kollmeier, 2006; ITU-T, 2011). ''Single-ended'' quality models that work only on the system's output signals also exist (e.g. ITU-T, 2004; Suelzle et al, 2013; Falk et al, 2015) but have not yet been applied to hearing impairment. They are not considered here.

The requirement of a reference signal poses a yet unsolved problem for the evaluation of systems such as hearing devices which introduce signal alterations intended to improve the perceived quality of the input signal for the individual user. The Hearing Aid Speech Quality Index (HASQI; Arehart et al, 2011), for example, builds a reference signal from the input signal of the assessed hearing device by filtering it according to a standard prescription rule used to fit hearing aids to individual hearing losses. Obviously, this approach is problematic when new prescription rules or dynamic compression algorithms are evaluated, as by definition any deviation from the reference is interpreted as quality degradation.

A potential solution is to use the unprocessed input signal's internal representation obtained by a normal hearing auditory model. This approach is followed by a new speech intelligibility metric for hearing aids, the ''Hearing Aid Speech Perception Index'' (HASPI) (Falk et al, 2015), but to our knowledge, has not been used by any quality metric for hearing-impaired. This appears worthwhile to be investigated in more detail. An alternative approach to the aided user performance prediction including properties of the individual's impaired auditory system utilises the ''Framework for Auditory Discrimination Experiments'' FADE approach outlined above (Kollmeier et al, 2016). The individual SRT is estimated using the hearing aid output processed with an automatic speech recogniser (ASR) that utilises a front end adapted to the individual impairment and a backend trained with a normal-hearing front end (Kollmeier et al, 2016). The data support the high SRT prediction performance of FADE for listeners with normal hearing ($R^2 = 0.92$) and for the median of listeners with impaired hearing ($R^2 = 0.80$). The benefit from (binaural) noise suppression schemes for both groups of listeners is accurately predicted based on audiogram and better-ear listening information only (i.e. no specific binaural interaction prediction, see Figure 6). For hearing-impaired listeners, the individual SRTs are slightly overestimated and the benefit show high variability. Hence, extra suprathreshold processing deficit parameters are required for a more accurate auditory model.

Binaural quality models are subject of ongoing research and development (e.g. Schäfer et al, 2013; Fleßner et al, 2014). So far, there is no international standard or recommendation like in the area of monaural speech and audio quality models (ITU-T, 2011).

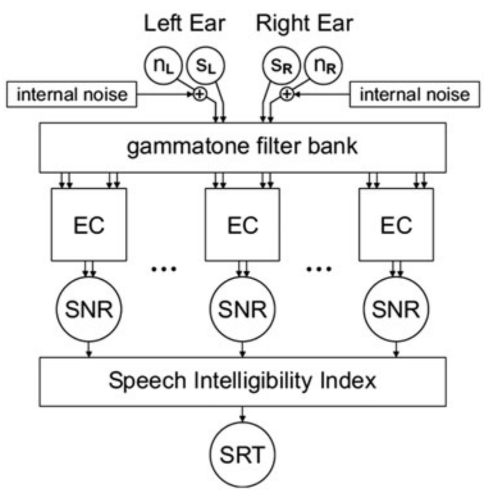

Figure 4. Schematic diagram of the BSIM-Model (Beutelmann et al., 2010): The abbreviations s_L *and* n_L denote the speech and noise signals at the left ear of the subject, respectively, and s_R and n_R the same for the right ear. EC stands for the equalisation–cancellation process, which results in estimated SNRs in each frequency band. The speech intelligibility index transforms the SNRs into a SRT.

Cognitive level

The cognitive component of an individual hearing impairment has been of increasing interest in the literature within the last decade. A large body of literature has considered the covariation among hearing loss, the loss of speech perception abilities in noise and a number of experimental outcomes that characterise different aspects of the individuals (remaining) cognitive abilities, such as, e.g. working memory, lexical access time, size of vocabulary (Rönnberg et al, 2013; Lunner et al, 2009). Moreover, a number of models have been established to relate different aspects of cognitive performance to each other, by primarily considering qualitative models and

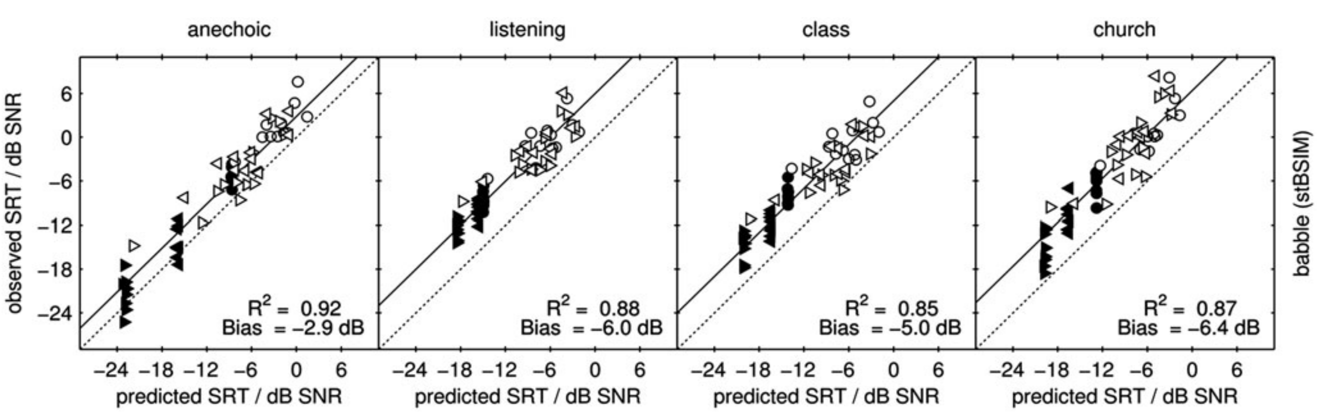

Figure 5. Scatter plots of observed SRTs against predicted SRT (from Beutelmann et al., 2010) for fluctuating babble noise and four different rooms using the short-term BSIM-model. The data from normal-hearing subjects are plotted with filled symbols and the data from hearing-impaired subjects with open symbols. The noise azimuths are indicated by symbol type (circle: 0, left-pointing triangle: −45, right-pointing triangle: 105 degree). The squared correlation coefficient and the bias (mean difference between predicted and observed SRTs) are given in the lower right corner.

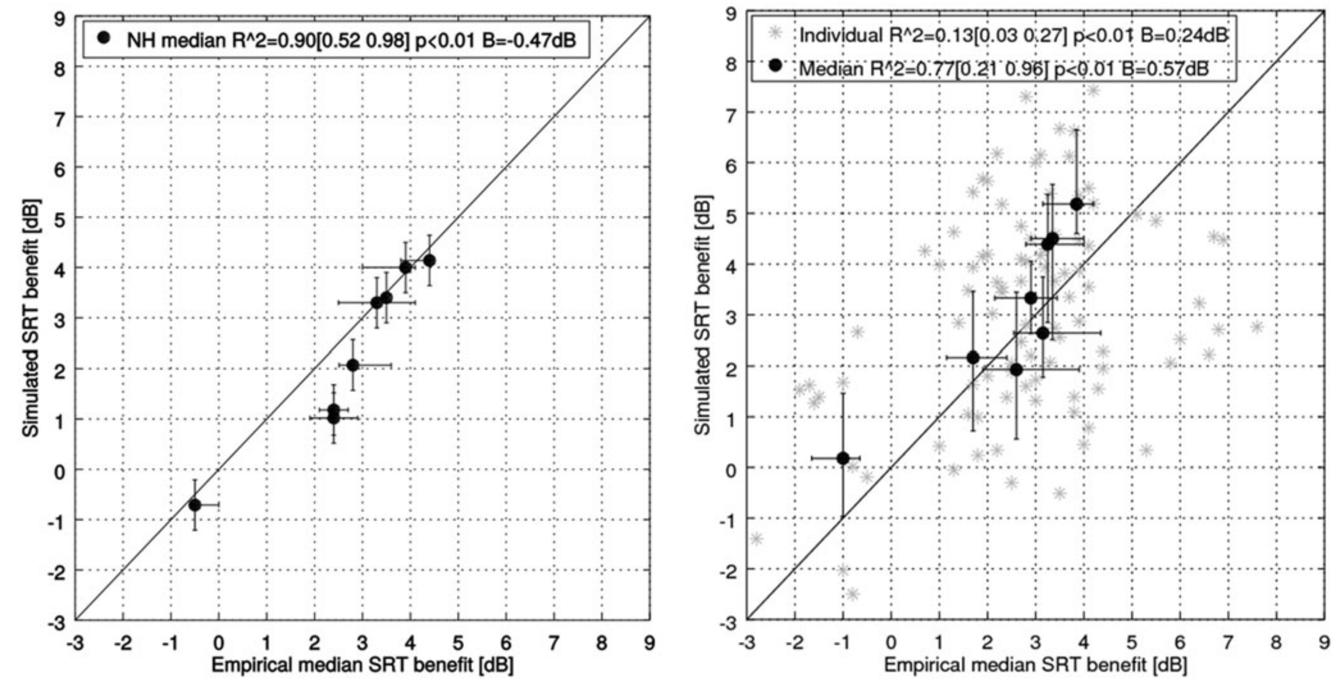

Figure 6. Predicted benefit in SRT from various binaural hearing aid processing schemes (empirical data from Völker et al., 2015) predictions from Kollmeier et al. (2016) for a group of listeners with normal (left) and impaired (right) hearing. Plotted are group medians of simulated versus empirical benefits with interquartile ranges. The individual predictions of SRT benefit for listeners with impaired hearing are also given in grey (right).

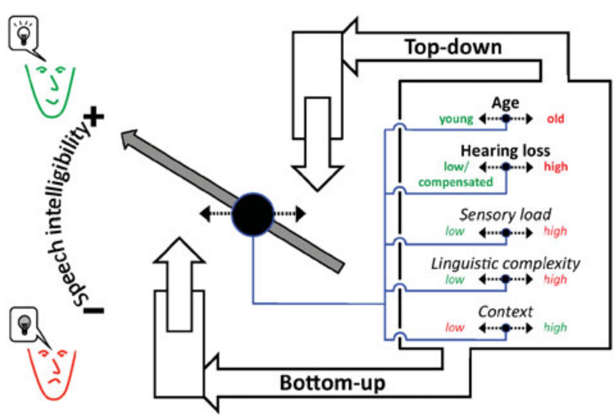

Figure 7. Mechanical model of the change in balance between bottom-up and top-down processing by Uslar et al. (2015). Words printed in green depict the side of the respective scale which usually leads to higher intelligibility, whereas words printed in red depict the side of the respective scale which usually leads to lower intelligibility. The pivot point (and hence the balance) between top-down and bottom-up processing is shifted by each of the five main factors depicted to the right.

empirical functional relations. The most prominent cognitive model is the ELU model by Rönnberg (ease of listening and understanding; Rönnberg, 2003) that separates two kinds of internal speech processing, i.e. one regular, low resources-consuming implicit processing path and one irregular, effortful explicit processing path requiring additional resources which is only activated if some processing problems occur, thus assuming two distinct processing strategies.

A model on cognitive variables and their influence on speech perception thresholds in hearing-impaired and aged listeners has been presented by Uslar (2014) that employs less detailed assumptions about the speech processing modules employed (Figure 7). It assumes a continuously shifting equilibrium among fast, input-driven bottom-up processing and more effortful and slower, experience-driven top-down processing. The speech recognition process is assumed to be a gauge where different listener-specific factors can shift the pivot point. Hence, individual factors like age and hearing loss are balanced with task-specific factors like sensory load (as implied by the signal-to-noise ratio), linguistic complexity of the sentences presented and the context information provided during the recognition task (more details see Uslar, 2014; Uslar et al, 2015). Hence, top-down processing is modelled to be more dominant with increasing age, hearing loss, sensory load, linguistic complexity and context, whereas bottom-up processing gains importance with a decrease of these factors.

Another quantitative way for estimating the amount of the cognitive component on the total effect of hearing impairment on speech recognition is to first predict the speech reception threshold with the individual model prediction utilising the models of the acoustical and the neurosensory level. The second step is to interpret the deviation between the predicted speech reception threshold and the actually observed SRT as the cognitive component. This approach hence estimates the remaining inter-individual variability after compensating for the sensorineural effects by predicting the individual neurosensory processing deficits by the appropriate model and denotes this estimate as cognitive component. Even though this arguably puts the cognitive component into the vicinity of the residual variance (or, equivalently unknown territory) in speech intelligibility modelling for hearing-impaired listeners, it is currently the only viable quantitative approach pursued. Nevertheless, a convincing relation between this remaining inter-individual variability and the inter-individual variability in any of the cognitive parameters tested so far has not yet been confirmed. Hence, there is still room for better models of the ''effective'' processing to be used in the hearing aid research community especially when it comes to a prediction of the effect of cognitive processing.

Consequences for hearing aid functionality

Based on physiological and psychoacoustical evidence, several factors of hearing impairment have been identified (Moore, 1995; Kollmeier, 1996; Akeroyd, 2006), which require different signal processing approaches for their compensation. As discussed above, we can distinguish between the following factors involved in (sensorineural) hearing loss that call for a certain rehabilitation strategy which again causes some technical challenges:

(A) Attenuation component of the hearing loss

Due to a loss of inner and outer hair cells in the inner ear (the cochlea), the hearing threshold is increased by a frequency-specific amount (loss of sensitivity). The rehabilitation strategy is a frequency-specific amplification which is limited in hearing aids by acoustic feedback and acoustic distortions due to the limited output power at a given acoustic quality in hearing aids. In addition, in those frequency regions where no active sensors/inner hair cells are still present anymore (CDR: ''cochlear dead regions''), the acoustic information might be transformed to be accessible to sensors at other frequency regions a strategy denoted as ''frequency lowering'' (or compression, transposition, or similar) which may lead to audible artefacts or distortions while the user benefit is still unclear and needs further research (McCreery et al, 2012; Bentler et al, 2014; Ellis & Munro, 2015).

(B) Distortion component of the hearing loss

Further perceptual consequences of a neurosensory hearing loss – most probably connected to a loss in outer hair cells – are the reduction of the dynamic range between the hearing threshold and the level at which the sound gets uncomfortably loud (''recruitment phenomenon''), a reduction in frequency selectivity, i.e. the ability to segregate two nearby frequency components of a sound and – partially as a consequence of these factors – an increased susceptibility to background noise (which will be treated in connection to the neural component, see below). To compensate for the attenuation and distortion factors, frequency-specific amplification and dynamic-range compression is required to make soft sounds audible and to compress the large dynamic range of everyday sounds into the residual dynamic range of the hearing impaired (''loudness compensation''). The challenge with these techniques is to establish an appropriate compression characteristic with suitable time constants and interaction across frequency bands.

To compensate for the reduced frequency selectivity, speech enhancement such as spectral sharpening is required which by itself may introduce artefacts and has not yet demonstrated a convincing benefit to the user.

(C) Neural component of the hearing loss

Due to the alterations in the nerve activity sent to the brain in cases of cochlear hearing loss, subsequent neural processing is also hampered, leading to deficits in binaural processing (i.e. impaired binaural capabilities) and to a reduced ability to communicate in difficult listening conditions characterised by noise and reverberation (''Cocktail-Party Effect'') that may both be attributed to lost binaural functions and to a blurred or distorted ''internal representation'' of the acoustical signals entering the ears.

To compensate for the increased susceptibility to noise and reverberation, noise reduction is applied to increase speech intelligibility and to lower listening effort in noise conditions that either operates on a single microphone channel (monaural noise reduction), on the output of multi- microphone arrays or other kinds of directional microphones (including beamformers) or on the input to both ears (binaural noise reduction). Typical challenges with all types of noise reduction are their inability to function well in situations where the respective assumptions of the algorithms are not fulfilled, which may lead to artefacts or other unwanted side effects. This calls for a sufficiently accurate procedure in estimating the underlying acoustical situation in order to activate only those algorithms with appropriate parameter values that are promising for this situation.

Note that all factors are individual to the hearing-impaired person and processing schemes thus need to be fitted to the individual hearing loss, e.g. the amount of frequency-specific amplification and compression. Several audiological measurement procedures and fitting rules are established to individualise the processing (see section Fitting software and Dillon, 2012).

The processing schemes mentioned above can be divided into two different categories. *Compensation schemes* aim at processing the signal so that the perception of the hearing-impaired, when listening to the processed signals (''aided perception''), matches as good as possible the perception of a normal-hearing (reference) person without processing. Compensation schemes are explicitly or implicitly based on a model of the altered perception (auditory models or: perceptual models). Examples are loudness compensation by amplification and compression and spectral sharpening. A major problem of these schemes is the fact that non-linear system compensation is hard to achieve: ''Aided'' perception is described by a succession of two non-linear systems that does not necessarily reproduce the non-linear normal-hearing (reference) system for all possible signals.

The other class of approaches consists of *substitution schemes* aiming at substituting aspects of auditory perception that are no longer functional in the impaired system anymore. Noise reduction for improving the signal-to-noise ratio (SNR) is an example of this type of processing. Substitution schemes might simulate the auditory processing in the normal-hearing auditory system or might be implemented following a purely technical approach that does not take into account how signals are processed in the auditory system.

A principal constraint of all the approaches mentioned so far is that they have to operate in real hearing aids in real-life situations which limits the choice of processing schemes, e.g. to only those algorithms that run in real time and exhibit a high robustness against ''unknown'' acoustic conditions which they are not originally designed for. Figure 8 shows the signal processing block diagram of a typical binaural hearing aid employing the different processing schemes outlined above. Due to the availability of ultra-low power wireless transmission of parameters and signals across ears, full binaural processing, i.e. comparison and joint processing of signals and signal parameters from left and right ear is possible, at least in today's high-end hearing aids.

It should be noted that the delay between output and input signal caused by the signal processing (algorithmic delay) and the hardware (analogue-digital and digital-analogue converters and

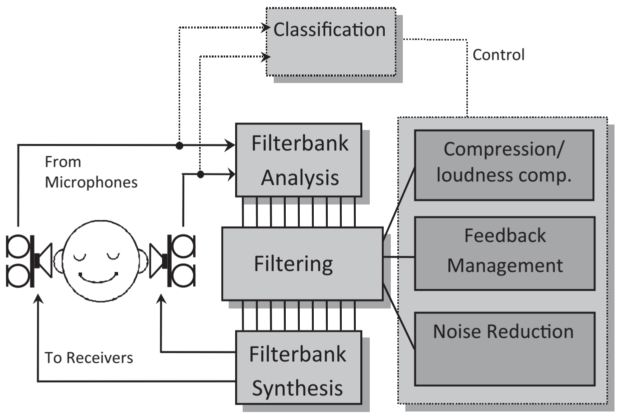

Figure 8. Signal processing block diagram of a typical binaural hearing aid. One or more (typically two, as shown here) microphones receive the acoustic signals close to the ears. Signals are processed digitally by a digital signal-processing unit (analogue–digital and digital–analogue conversion not shown here) and processed signals are sent to the speakers (''receivers'') at both ears. Typically, frequency analysis and synthesis is performed by a filterbank and filtering is then performed in several frequency bands. Filtering typically comprised compression/loudness compensation, feedback management and noise reduction. Filtering can be controlled by automatic classification of the current acoustic signal (e.g. activating or deactivating noise reduction depending on whether the presence of noise is detected).

signal buffers) needs to be considered. To keep perceived audio and visual information synchronised and to accommodate lip reading, a maximum allowable delay of about 50 ms was empirically found. In the very frequent cases of small hearing loss at low frequencies, however, both the direct and the processed signals are audible in specific frequency regions. This effect occurs particularly for the acoustically transparent ear canal in open-ear fittings and lowers the allowable delay further. The superposition of the direct and the delayed signal sounds reverberant, which is disturbing to the user and may lower speech intelligibility. Empirical results show that the delay should be smaller than 10 ms to avoid detrimental effects (Agnew & Thornton, 2000). Systems based on the Fast Fourier Transform (FFT) almost reach this limit. Filterbank-based systems employ lower delays between 6 and 2 ms, depending on frequency. A consequence of the delay limit is that processing schemes that require short-term memory (e.g. directly compensating for or substituting cognitive processes ranging from approx. 50 ms to several seconds) cannot be implemented in the signal path, because they would cause a delay beyond the limit. Instead, a post-hoc processing has to be employed which estimates any necessary changes in the signal path from the past. One possibility is to use a parallel estimation path that constantly updates its memory of the past signals to make predictions on the future. This has the obvious disadvantage that the estimated change for the signal path may come too late and might be less accurate than if the complete knowledge of the input signal (including a view into the future) would have been available. However, such a parallel estimation process integrating across time yields the only opportunity to implement sophisticated signal modifications based on, for example, comprehensive auditory models (as reviewed in sections Neurosensory level and Cognitive level) that cover time constants between several milliseconds and several seconds.

Compensation strategies considering the attenuation component

Frequency-specific amplification, feedback and acoustic distortion reduction

As already discussed above (Table 3), frequency-specific amplification with an appropriate overall gain and frequency shaping constitute the ''standard treatment'' targeted for hearing aids that has also been the subject of the prescription and fitting rules described above (section Fitting software). This target definition of hearing aids is only limited by technical factors (such as, e.g. form factors and quality of the transceivers and electronic circuits for driving them), thus requiring an adequate feedback suppression and limitation of the output dynamic range in order to avoid too much distortion. From a modelling perspective, the following conclusions for this kind of processing in hearing aids can be drawn:

- A fixed spectral shaping of the amplification (i.e. fixed gain across frequency) is only a very rough compensation for hearing loss, primarily applicable for conductive hearing losses and for those cases where a small range of input signal levels and a limited output level range can be assumed which is primarily applicable for a number of laboratory experiments. However, this assumption does not hold in real life where the large dynamic range of input signals and the (non-linear) neurosensory hearing loss component usually involved in most hearing aid users require some kind of dynamic compression.
- The frequency resolution of the frequency shaping can be comparatively low (i.e. not exceeding the spectral resolution of the auditory critical bands covering the whole auditory frequency range with approximately 32 bands). This resolution limit is a consequence of the usual spectral smearing in the basilar membrane and excitation patterns (Zwicker, 1961; Moore & Glasberg, 1987) indicating that spectral energy in one frequency cannot be changed in the internal representation of the acoustical signal independently from the neighbouring frequency band if both bands are close together in frequency (as expressed in an auditory frequency scale related to the auditory critical band concept).
- Note, however, that some specific compensation functions may require a much higher spectral resolution (e.g. feedback reduction where notches and peaks in the complex transfer function of the acoustic environment in interaction with the hearing device have to be compensated for, or noise reduction schemes for stationary noise components where the SNR in one auditory band can only be improved if the noise within this band is separated from the signal component). Therefore, hearing aids need a high frequency resolution for slowly changing processes that, for example, reflect the adaptation to the environment whereas they need a low frequency resolution for fast-acting processes that compensate for deteriorated physiological functions of the user (such as, e.g. fast-acting dynamic compression to compensate for impaired cochlear processing).

Frequency lowering

To circumvent the effect of ''cochlear dead regions'' (Moore, 2001), frequency lowering schemes have been advocated in the literature and have been applied in commercial hearing devices without a clear prove of a benefit for listeners. They employ

Table 3. Processing strategies for the rehabilitation of the different factors involved in hearing impairment including the resulting technical challenges to be solved (and references).

Factor	*Perceptual consequence*	*Rehabilitation strategy*	*Technical challenges*	*References*
Attenuation component	Loss of sensitivity; increased threshold level	Increase audibility by (a) frequency-specific amplification (b) frequency compression	Acoustical feedback cancellation Acoustical distortion	Plomp (1994), Turner and Henry (2002), Grimm, et al. (2009), Greenberg et al. (2000), Turner and Hurtig (1999), and Sakamoto et al. (2000)
Distortion component	(a) Loss of sensitivity; reduced dynamic range ('Recruitment') (b) Reduced frequency selectivity (c) Increased susceptibility	(a) Automatic Gain Control (AGC), Multiband dynamic compression (b) Spectral enhancement (c) Noise reduction (see also: 'neural component')	(a) Compression characteristics, time constants band coupling (b) Artefact removal (c) Estimation/classification of speech and noise signals	Stone et al. (1999), Verschuure et al. (1996), and Jenstad et al. (1999) Baer and Moore, (1994) and Yang et al. (2003) Nordqvist and Leijon (2004), Büchler et al. (2005) and Ostendorf et al. (1998)
Neural component	(a) Increased susceptibility to background noise ('Cocktail-Party Effect') (b) Impaired binaural capabilities	(a) Monaural noise reduction (b) Directional microphones (c) Beamformer (d) Binaural noise reduction	(a) Estimation error & artefact removal (b) Lower corner frequency (c) Beam characteristics (d) Estimation error removal & transmission across ears	Ephraim and Malah, (1985) and Tchorz and Kollmeier (2003) Elko and Pong (1995) and Maj et al. (2005) Campbell and Shields (2003), Widrow and Luo (2003), Wittkop and Hohmann (2003), Bodden (1993) and Klasen et al. (2006)

compensation schemes that aim at spectrally redistributing signal information avoiding stimulation of such assumed dead regions (Alexander et al, 2014; Kollmeier et al, 2016). In the design and evaluation of such systems, the following consequences from the modelling approaches given above can be drawn:

- From an information theoretic view, frequency lowering makes sense because of the high redundancy/covariation of adjacent frequency bands during speech transmission, i.e. a complete and nearly perfect transmission of speech information is usually possible with as low as three independent spectral components (as has been shown, e.g. by approaches within a three-band vocoder (Shannon et al, 2004), or spectral peak resynthesis (De Vos et al, 2014; Mirkovic et al, 2015). This assumes, however, that the information in these few frequency bands can be independently decoded by the impaired auditory system and be interpreted correctly (normally requiring an extended training, i.e. regular use of the hearing aids). However, if any of these assumptions are violated (i.e. if the SNR of the internal noise or the spectral resolution in the hearing-impaired ear is not sufficient), the effect can be even negative because speech quality is degraded by the coding scheme. Moreover, speech reception is not only linked to spectral analysis but relies to a very high degree also on spectro-temporal and temporal processing which requires that any frequency lowering scheme must not interfere with the temporal representation of speech.
- As a consequence, only a limited effect is expected in the majority of hearing-impaired listeners and the vast variation of natural acoustical environments. Moreover, an extensive training would be required which can hardly be supplied in laboratory studies. This may contribute to the fact that no clear advantage of this kind of processing has been demonstrated in the literature so far.

Compensation strategies considering the distortion component

Multiband dynamic compression

The compressive non-linearity of the inner ear can be modelled as a fast-acting physiological dynamic compression of basilar membrane movement. Its loss is most commonly assumed to be caused by the loss of outer hair cells in the inner ear which should be compensated by a fast-acting multiband dynamic compression in hearing aids. Even though this concept for recruitment compensation (Villchur, 1977; Hohmann & Kollmeier, 1995) has led to several developments and implementations of multiband dynamic compression in current hearing devices, there is still a discrepancy between theory (i.e. fast-acting dynamic compression in many independent frequency bands) and practice (i.e. slow acting AGC in few frequency bands). This discrepancy is simply because of the low acceptance of fast-acting multiband dynamic compression system by hearing aid users for several possible reasons.

On one hand, fast-acting multiband dynamic compression reduces both the spectral contrast (i.e. minima in the spectrum get amplified, maxima get attenuated) and the temporal contrast. A sensorineural hearing loss with a suprathreshold distortion component, on the other hand, is supposed to cause an additional blurring (or, distortion) of the spectro-temporal contrast which might even worsen the problem of hearing-impaired listeners instead of compensation for the hearing impairment (which would rather require an enhancement of spectro-temporal contrast). Plomp (1988) even argued that the reduction in the modulation depth by dynamic compression (as expressed by the modulation transfer function which is the input for the speech transmission index by Houtgast & Steeneken, 1985) is directly responsible for the reduction of speech intelligibility caused by hearing aids. This argument could be counteracted by Hohmann and Kollmeier (1995) who showed that speech intelligibility can be preserved under fast-acting compression as long as the momentary signal-to-noise ratio in each frequency band is not changed indicating that the reduced modulation transfer function by multiband dynamic compression is not the reason for the limited benefit of fast-acting multiband dynamic compression schemes. Instead, slow-acting dynamic compression schemes that operate on comparatively broad filter bands and hence do not produce spectral smearing have been shown to be rather successful in the past with a highly perceived audio quality, thus establishing the standard compression scheme employed in modern hearing aids (Dillon, 2012).

One reason for the obvious need of preserving the spectral contrasts in practice – which contradicts the assumed model requirement of multiband dynamic compression in independent narrow bands – might be simply the fact that the model assumption of independent spectral bands in the auditory system is not correct: Several psychoacoustic and speech perception experiments and theories assume a strong across-frequency processing in the auditory system to recognise spectral profiles and to perform, e.g. object separation by comodulation masking release (Grose & Hall, 1992; Verhey et al, 1999). The assumption is that the ''internal'' auditory spectrum after preprocessing by the cochlea and the brainstem is subject to a central pattern recogniser which would be disturbed by a multiband dynamic compression with high compression factors and high spectral resolution.

This assumption that spectral contrasts are needed on the internal level (not necessarily on the input signal level, see below) is backed up by the model by Denk et al (2016) which models the property of the auditory system to be highly compressive for on-frequency input signal components (i.e. if the input signal is centred in the auditory band belonging to the respective auditory filter), whereas the auditory system behaves more passive and linearly for off-frequency components (i.e. no amplification and further compression if the incoming stimulus component is not centred within the auditory band considered). This local spectral processing (physiologically described by the effect of two tone-suppression and tuning curves) can be described as an instantaneous-frequency-dependent, local amplification and compression scheme within each frequency band: if an incoming frequency component coincides with its instantaneous frequency with the centre frequency of the band in consideration, then this component should be amplified and the gain should be increased. However, if the same level of the input component is produced by some kind of ''leaking over'' from the adjacent auditory band, the instantaneous frequency of the component does not coincide with the centre frequency and hence should not be amplified but rather treated as a side band excitation without further amplification and compression. Such a model leads to an instantaneous-frequency-sensitive processing scheme (Kortlang, 2016) which preserves the spectral contrast and still yields a multiband dynamic compression on a larger time scale. The results from such a processing (described by Lybarger, 1963; Kortlang et al, 2016 in the current issue) suggest that multiband dynamic compression with fast time constant and preservation of spectral contrast might be successfully used in future hearing devices.

A major reason why no clear ''optimum'' algorithm and set of processing parameters exist for performing dynamic compression as a compromise between the physiological requirements (i.e. limiting the output dynamic range, performing fast compression to compensate for the loss of ''internal'' compression in the auditory system) and the requirements for a high audio quality (i.e. producing as little distortions and artefacts as possible, long time constants) is the discrepancy between well-defined laboratory conditions and the more extreme acoustic conditions in real life. While laboratory situations typically exhibit limited dynamic range of presented test sounds, limited variations and temporal succession of acoustical scenes presented, and limited amount of movements of the microphone placement within the room, real-life situations usually cover a larger dynamic range, an abrupt transitions between acoustical scenes and more or less unpredictable head and body movements with a high variation of the input signal to the hearing aid microphone. Hence, laboratory studies carry only a limited validity for the benefit of a dynamic compression scheme in real life. In addition, the difference between the perceived benefit in dynamic compression schemes is small, i.e. the change in dynamic compression parameters usually causes only a very limited change of the perceived sound quality or obtained benefit from the processing, thus yielding a very shallow optimisation ''landscape'' in the multi-parameter room of dynamic compression algorithms. Hence, by principle, it is nearly impossible to find a global optimum in dynamic compression parameters and algorithms in the lab that will prove to be an optimum solution for real life conditions as well.

In addition, the empirical approaches to dynamic compression installed in most current commercial hearing aids already provide an acceptable solution which is hard to further improve with any new dynamic compression scheme. Hence, even though dynamic compression and model-based compression schemes appear to be a challenging research topic with a potential advantage of compensation for distinct auditory deficits, the achievable benefit over empirically optimised compression approaches in commercial hearing devices is very limited (see Kortlang et al, 2016).

Spectral enhancement

Since the discovery of the limited dynamic range in sensorineural hearing-impaired listeners and the loss of spectral contrast produced by multiband dynamic compression schemes, spectral enhancement has been proposed and tested in a number of studies that either showed a very small or even a negative effect (Baer et al, 1993; Gerkmann et al, 2015). From the perspective of the models described above, there are several reasons that may explain why:

- The main assumption that several perceptual consequences of hearing impairment are due to an increased auditory filter bandwidth in sensorineural hearing-impaired listeners which should be compensated for by some spectral enhancement may not be valid: the largest difference in estimated auditory bandwidth between normal and hearing-impaired listeners can be found if the same level above individual threshold is considered, particularly in tone-in-noised detection experiments (Patterson et al, 1982; Moore & Glasberg, 1983; Moore, 1996). This difference vanishes at least partially if a comparison is made on the basis of the same absolute level or even when considering not the assumed constant input, but a constant output of the auditory filter (Hafed et al, 2016). This is a consequence of the non-linearity of the auditory system with an ''effective'' filter characteristic that varies with level and is altered as a consequence of sensorineural hearing impairment: Any filter bandwidth estimation procedure that simultaneously takes the altered non-linearity of the impaired auditory system into account, therefore, yields smaller differences in estimated filter bandwidth. In addition, the effect of increased filter bandwidth with increasing level in normal listeners seems to have much less consequence on the performance (e.g. tone-in-noise detection thresholds or speech recognition thresholds in noise) as would be predicted if the filter bandwidth would have such an dominant influence on these performance measures (Plomp, 1986). This limited effect of bandwidth widening is also supported by modelling results of speech recognition thresholds using a psychoacoustically motivated auditory frontend where the variation in filter bandwidth produces nearly no effect on the predicted speech recognition thresholds (Schädler et al, 2016).

- The second argument why spectral enhancement does not appear to be an appropriate solution was already discussed above: clearly, a signal component classified as ''side excitation'' (i.e. energy which is leaking from adjacent frequency bands without being normally processed within the respective frequency band in the normal auditory system) should not be amplified, but rather suppressed in a model-based dynamic compression scheme (see Kortlang et al, 2016; Denk et al, 2016). However, a pure spectral peak and valley detector subsequent to an auditory filter bank is not able to differentiate if a certain minimum in the spectral shape is due to a leakage from adjacent filter bands or if an independent, soft sound component is present which should be amplified rather than attenuated. Hence, an effective spectral enhancement scheme will have to utilise more information than just the spectrum (such as, e.g. instantaneous frequency information or better knowledge of the target spectrum to be expected, see above).
- A third caveat for a limited utility of multiband dynamic compression is the spectral smearing property of the auditory system by itself: even if a highly peaky, spectrally optimally sharpened input signal spectrum is available, the filtering of the auditory system with comparatively broad ''effective'' auditory analysis filters will inevitably produce a comparatively smooth spectral ''internal representation'' of the incoming signal. A more pronounced or ''peaky'' internal spectral presentation might only be achieved, if some ''negative'' spectral energy would be inserted which is, however, not possible. Hence, from a modeller's view, it appears to be reasonable to not invest too much further research efforts in developing hearing aid systems with spectral enhancement properties.

Substitutional strategies considering the neural component

Since the main problem of any deterioration in the internal, neural representation of the acoustical scenario in the auditory system is the loss in being able to separate across auditory objects (such as e.g. target speech and background noise) which is equivalent to losing the ''internal contrast'' in a mixture of signal and noise, we will concentrate in the following on approaches to perform a noise reduction. As pointed out before, such schemes aim at substituting lost auditory functions by an appropriate processing of the signals before being presented to the patient. Hence, they are classified as substitutional strategies. They can be classified according to the underlying processing principles into monaural or single-channel noise reduction, directional microphones and/or beamformers and binaural noise reduction.

(Monaural) noise reduction

In most current hearing aids, one or another type of single channel noise reduction is employed, since only one microphone signal (or the combined output of several microphones to a directional microphone array output signal) is utilised for further processing. Using concurrent sophisticated noise reduction techniques (usually employing offline methods like blind sound source separation), impressive noise reduction can theoretically be achieved on these signals as long as one of the common assumptions about the interfering noise is fulfilled (i.e. stationarity, deviations from the target signal in terms of spectral shape, modulation spectrum, amplitude statistics or pitch). However, when implemented for real-time usage with low latency and to be used in real-life situations, the performance of such techniques is very limited and the possible benefit may vanish completely due to the following reasons:

(a) A significant increase in signal-to-noise ratio on a single-channel input can only be achieved if noise removal is performed with a sufficiently high spectral and temporal resolution. It must be higher than the spectral or temporal differences between target signal and interfering noise especially under the restriction that a spectral and temporal smearing occurs in the auditory system (as described by the models discussed above). However, if this ''ideal'' processing is no longer based on a priori knowledge of the target signal and interfering noise statistics, but instead a ''non-ideal'' case having insufficient a priori knowledge assumed, some processing artefacts will inevitably occur due to the suboptimal filter action in the time and frequency domain. One way to circumvent such artefacts is to use an auditory filterbank approach with a time and frequency resolution fitted to the auditory systems which results in the least audible artefacts. However, such a time–frequency manipulation on the scale of auditory temporal and spectral resolution is not sufficient for an efficient separation between signal and noise at each pixel in the time–frequency domain (see above) and hence cannot improve the signal-to-noise ratio inherent in each auditory frequency band. This mutual excluding requirement (i.e. high spectral and temporal resolution to effectively remove noise components versus low or auditory-model-based spectro-temporal resolution toward avoiding audible processing artefacts) clearly limits the achievable benefit from single channel noise reduction algorithms.

(b) Validity of the statics considered: a slow, but steady progress in improving the performance of single-channel noise reduction algorithms has been achieved over the past years mainly because the inherent statistics of target speech signals and interfering noise or environmental sound sources have been taken into account in much more sophisticated ways than before (Martin, 2001; Gerkmann & Krawczyk, 2013). If the underlying assumptions that are pursued with their respective statistics are fulfilled, still some improvement in ''effective'' SNR'' can be achieved if filter characteristics with a better time- or frequency-resolution than the internal auditory filter process can be derived (see discussion above). However, a linkage between the auditory time constants normally required to fulfil these restrictions and the statistical parameters for background noise and target speech does not exist. It may be assumed that future work with such a linkage between the statistics and auditory characteristics might offer more, but still limited progress in this area.

(c) A number of *instrumental estimation techniques* are available for assessing the effect of an algorithmic intervention (such as, e.g. noise reduction) with an instrumental measure that has implemented auditory-model-based signal processing to a certain extent. Even though most of these methods implement some kind of auditory model to evaluate the effect of any signal-processing scheme not on the acoustical, but on some kind of assumed perceptual, internal level, their validity in predicting the effect of a given algorithm on human performance in normal or hearing-impaired listeners is still very limited. However, the most recent approaches like HASQI (Arehart et al, 2011) and PEMO_Q (Huber & Kollmeier, 2006) or even the binaural version by (Fleßner et al, 2014) already achieve an impressive prediction accuracy. Since these models are usually optimised and adjusted to a certain set of training materials and algorithms, and a number of model assumptions have been made that are not necessarily fulfilled in the individual's auditory system, the generalisation ability of these methods to yet unknown processing schemes and their potential insight into ways of better processing acoustical information in hearing aids is very limited. Hence, caution is necessary if the evaluation of the effect of a hearing aid algorithm is only judged by such instrumental measures without a clear validation of this model prediction with real human test persons.

Directional microphones and beamformers

Fixed directional microphones as well as fixed or adaptive beamformers rely on the assumption that the target source is in the range of directions where no attenuation is performed by the beamformer, while the interference is supposed to be attenuated due to originating from other directions. If this assumption is fulfilled, a certain increase in ''effective'' signal-to-noise ratio results which leads to a corresponding hearing aid benefit which can be predicted adequately from models operating on the acoustical level (such as, e.g. speech intelligibility index, see above). However, the following restrictions apply:

- The assumptions of the directional microphone or fixed beamformer may no longer hold as soon as head movements and acoustically reflecting objects come close to the microphone. Even though this can partially be compensated by an adaptive beamformer (designed under the assumption that either the target or a noise comes from a prescribed range of directions) this does not add too much benefit in real situations in a comparison to fixed beamformers (Maj et al, 2006; Doclo et al, 2007). An ''ideal'' solution would foresee a closed-loop system where the user can direct the main lobe of the directional microphone according to his or her current preference. This would require a sophisticated human–machine interface (or even a brain–computer interface, e.g. De Vos et al, 2014; Mirkovic et al, 2015) which is still not feasible as a commercial product.
- The directional gain or microphone array gain critically depends on the physical dimensions of the maximum distance across the microphone in units of the sound wavelength. This indicates that – given the limited dimensions of directional microphones or separation across microphones in hearing aids – a substantial benefit can only be achieved at high frequencies while at low frequencies the microphone noise limits the processing efficiency. An ''effective'' increase of microphone array dimension can be achieved if arrays exterior to the hearing aid are used, e.g. if the microphones from both hearing aids at the left and right side are combined (binaural array processing, see below) or if microphone arrays are built into spectacles (Soede et al, 1993) or clothing (Widrow, 1998).

Binaural noise reduction

In the case of a ''true'' binaural processing in hearing aids with an exchange of full-bandwidth audio signals between the left and right side of the head, several processing options are available that aim at either supporting or even replacing the ''natural'' binaural signal processing in the normal auditory system. Hence, their potential benefit can be assessed in view of the binaural processing models outlined above (section Neurosensory level) as follows:

- In the case of a binaural microphone array ($N>=2$ microphone inputs, one array output), a high directional filtering effect (i.e. high array gain) can be achieved even at low frequencies which has been shown to produce large hearing aid benefits for a given arrangement of signal and sound sources (peripherally in anechoic conditions (Wittkop & Hohmann, 2003; Rohdenburg et al, 2008; Van den Bogaert et al, 2009, see Figure 9). However, as stated above, the effect vanishes in practice as soon as natural movements of the head and the surrounding objects are encountered and the assumptions of the respective directional filtering algorithms are no longer fulfilled. Moreover, the users report a kind of ''locked in'' impression in the sense that they can primarily follow that part of the acoustical environment which is in the (narrow) beam of the beamformer whereas they subjectively feel excluded from perceiving audible sound emanating from other directions. Hence, what appears to be a useful aspect when concentrating on a single sound source in a noisy environment and directing the head exactly to the sound source turns out to be detrimental for other hearing aid users who rather would preserve an impression of the different environmental sounds surrounding them. This effect obviously limits the utility of too effective/ too narrow beamformers. It calls for closed loop-systems to steer the beamformer or ''true'' binaural support systems that relay on some basic remaining binaural functionality in the hearing-impaired listener to keep track of different objects from different directions.
- Such a solution that aims at preserving some of the binaural cues to the binaural system to be used for localisation and some internal noise suppression is at least partially achieved by multiple input-multiple output (MIMO-) binaural hearing aid algorithms that operate on at least two binaural input channels and yield at least two output channels to be routed into both ears of the hearing aid user. The idea behind the underlying processing algorithms is either to attenuate certain ''unwanted'' sound components in both ears simultaneously using the same time- and frequency-dependent gain at both ears and hence preserve all binaural differences across ears (Kollmeier et al, 1993; Wittkop & Hohmann, 2003) or to parametrically select the degree to which interaural level differences are preserved during processing (Van den Bogaert et al, 2007; Marquardt et al, 2014). From a modelling view point which takes into account the comparatively simple binaural processing to be performed (EC-processing within the BSIM-model or other models, see above), it is clear that any extracorporal binaural noise reduction (using ''true binaural hearing aids'' with noise cancelling techniques similar to the

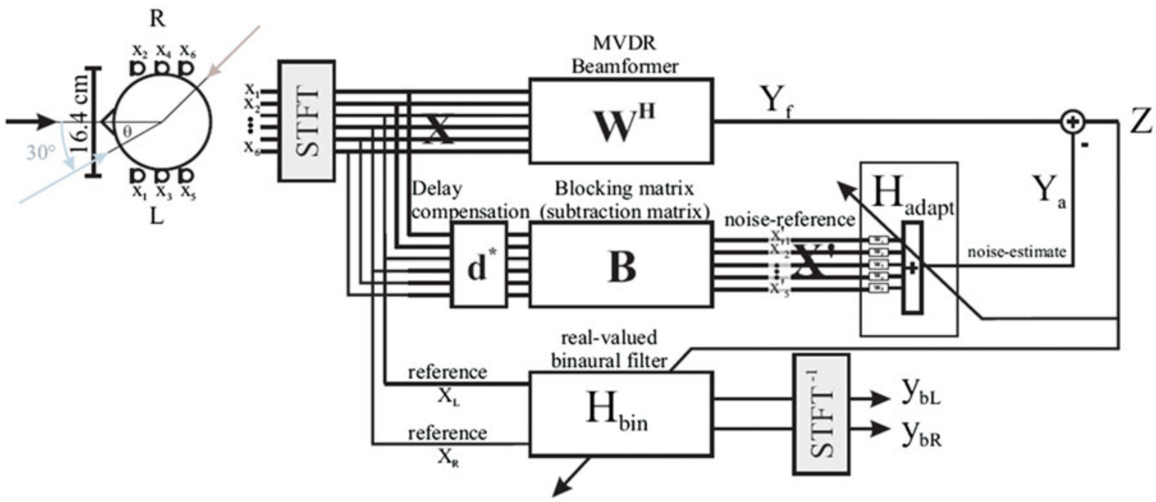

Figure 9. Binaural signal processing scheme according to Rohdenburg et al. (2008).

ones employed in auditory models to describe the internal noise reduction in the auditory system) cannot be expected to be independent from the remaining binaural cues and their maximum usability for further noise reduction in the users own binaural system. In other words, the sum out of hearing aid binaural noise reduction and internal binaural noise reduction can be assumed to be roughly constant. However, is not clear yet how the remaining binaural noise reduction capability of the individual hearing aid user should be assessed and how the (maximum) noise reduction in binaural MIMO hearing aids can be tailored to these binaural remaining capabilities in order to yield the best overall performance.

- Besides the pure noise reduction performed by binaural algorithms (assuming different interaural relations for the target signal and any interfering noise), there are also other binaural features indirectly coupled to noise suppression that can be performed by a ‘‘true binaural hearing device’’ and had been shown to yield some advantage for the subjective evaluation of hearing with both ears. One such example is interaural coherence filtering (Wittkop & Hohmann, 2003; Grimm et al, 2009) which assumes that the reverberant and distant-sounding part of an incoming sound which shows interaural correlation should be suppressed whereas the direct sound of sources close to the listeners head show a high interaural coherence and should be enhanced). Another example is interaural magnification, i.e. approaches to even increase the differences across both ears (see Durlach & Pang, 1986; Peissig et al, 1996; de Taillez et al, 2016, this issue). However, it is unclear, how such algorithms should be fitted to the individual binaural remaining capability (which is obviously coupled to the limited knowledge of the ‘‘binaural component’’ of the hearing loss, see above) and how the perceived quality after the processing with such algorithms that aim at compensating for some (lost) binaural processing capabilities can be objectively assessed. Some additional attempts to build up a binaural quality model for objectively assessing the effect of such binaural processing schemes are proposed by Fleßner et al (2014). This is obtained based on a physiological motivated model of binaural signal processing in the auditory system.

To summarise, several examples have been discussed where the available knowledge about human auditory signal processing (usually represented by one ‘‘effective’’ model of signal processing in the auditory signal) can be utilised to assess the expectation on certain hearing aid features. On the other hand, this chapter also shows that our understanding both of human signal processing and the optimum hearing aid processing as well as its fitting and situation-depending steering of parameters is still very limited. This calls for much more research, parts of which are reported in this supplement.

Outlook into new solutions

Advanced amplification, dynamic compression and noise reduction for monaural hearing aids

Even though the technological and audiological progress in hearing aids has been impressive during the last decade, there are still unsolved problems in current hearing aid technology calling for improvement. Some of them are due to physical and technical constraints (such as, e.g. the limited acoustic output power and frequency range for receivers with a limited physical size, the limited energy storage capacity of batteries per volume, the power consumption of a custom-designed integrated circuit for a given complexity of hearing aid operation, etc.) and other properties are due to limitations in audiological concepts and knowledge (see above, section Consequences for hearing aid functionality). Hence, current research focuses on overcoming these limitations as has been reviewed in this paper and can also be seen from the papers in the current issue of this journal. Within the restrictions of a single, monaural hearing aid (which may as well be coupled with another hearing aid on the other side of the head), the following new, emerging solutions should be mentioned:

- ‘‘HiFi’’ *Hearing aid*: Similar to the HiFi-requirements in consumer electronics, a large transmission frequency range and a large dynamic range with a minimum distortion would be desirable for a hearing device. Note, however, that HiFi for the individual listener relates to a different concept (i.e. maximum fidelity of the transmitted signal at the output of the individual, impaired auditory system), than for the normal listener where such a specification can be made on the physical sound signal alone rather than on any assumed perceived ‘‘internal representation’’ of the hearing aid user. However, with increasing knowledge about the relevant properties to be accounted for (and eventually being included in an auditory model for the individual hearing-impaired listener), a better match will be achieved between the produced output signal of the hearing aid and the maximum achievable transmission quality for the individual user. This may involve omitting certain parts of the physical signal (such as, e.g. very high-frequency components or some signal components that are anyhow masked by other signal components during auditory signal processing). Insofar, such an ‘‘individual HiFi’’-specification might as well be termed a kind of ‘‘MP 3 for hearing devices’’, i.e. the coding and transmission of only those signal components that are relevant for the perception while suppressing (and not transmitting) those signal components that are irrelevant or being masked. This is in line with audio coding approaches such as, e.g. MP3 (Brandenburg & Popp, 2000). Besides such sophisticated model-driven signal coding algorithms (see below), a prerequisite for such a ‘‘HiFi Hearing aid’’ would, therefore, be high-quality transceivers (with a high usable dynamic range and frequency range), a close coupling between the receiver and the eardrum (usually achieved by placing the receiver output as close to the tympanic membrane as possible), and some kind of ‘‘acoustic transparency’’, i.e. the concept that inserting the amplified sound by the hearing device in the open or (partially) occluded ear canal will be equalised in such a way that the subjective impression is not changed compared to the unaided, open ear canal impression. Note that the target definition of such an acoustic transparency is not an unambiguous choice, since it might consider an equalised spectral response either at the input of the ear canal, at a certain insertion depth within the ear canal, or even at the (modelled or measured) ear drum (Blau et al, 2013; Denk et al, 2016). Hence, ongoing work considers the perceptual consequences of different reference concepts and the necessary acoustical measurements that have to be performed to compensate for any of the necessary insertion gains.

- *Spectral processing*: Even though the meanwhile commercially available algorithms for spectral enhancement and for frequency shifting appear to have only a very limited benefit (see above, discussion Compensation strategies considering the distortion component), still some aspects of processing in the frequency domain in order to achieve a better approximation to a ‘‘HiFi hearing device’’ might be feasible: this relates to more sophisticated noise reduction methods for single channel noise reduction utilising advanced statistical and machine learning knowledge (e.g. Gerkmann et al, 2015; Lybarger, 1963) and have a chance for a certain noise reduction as long as the underlying model assumptions are met. This also may hold for model-based dynamic compression schemes such as outlined by (Hohmann & Kollmeier, 2007) or (Kortlang et al, 2016) that utilise innovative audiological concepts and perception models. Ultimately, a valid and robust, but still fast working real-time auditory model might be incorporated in hearing aids (‘‘model-in-the-loop’’-concept) which performs an online assessment of the achieved perceptual sound quality for the individual listener. Such low-latency, limited complexity real-time auditory models for perceived quality prediction in hearing-impaired listeners are currently being developed (Huber et al, 2014) that might be suitable to work as ‘‘model in the loop’’. This is, however, a research challenge worth being pursued in the future.

Advanced binaural hearing aid functions

The apparent technical progress in realising binaural links across hearing devices at full bandwidth with low power consumption allows for a better microphone array processing using microphones on both sides of the head to achieve a better directionality at lower frequencies and thus a higher signal-to-noise ratio (see above). It also creates the prerequisites for compensating some of the lost binaural processing functions of the individual auditory system. Even if a clear ‘‘binaural impairment factor’’ describing the deterioration in binaural functions has not been identified yet, nevertheless some binaural enhancement which already helps normal listeners in acoustically ‘‘difficult’’ listening situations might be suitable as well for hearing-impaired listeners even with an unclear binaural processing capability. Possible emerging solutions that are current topics of research are as follows:

- Binaural enhancement

  Even though binaural noise reduction systems have been shown to achieve substantial benefits in lab situations due to suppressing unwanted, lateral sound sources and reverberation, their evaluated benefit in field tests and in real life conditions appears to be small (Völker et al, 2015; Grimm et al, 2016; de Taillez et al, 2016). Future solutions for binaural noise reduction in hearing aids will, therefore, only be successful if the reasons for this limited benefit of binaural noise suppression are clear. One such possible reason is the fact that most binaural beamformers operate on a fixed spatial configuration assuming the target signal to be in front of the subject’s head. This, however, is not realistic in real environments and does not take any head movements or attention switching of the user into account.

  (Footnote: If one compares this situation to vision, a fixed binaural processing scheme acts like a telescope which is fixed to the subjects head and does not support the magnification/enhancement of objects if eye movements position the fovea of the subject to other directions. Hence, solutions that take the dynamics of eye gazing into account are desirable and have been shown to be successful, e.g. in retina implants, Hafed et al, 2016).

  Future solutions for a ‘‘space-aware hearing aid’’ should, therefore, provide the maximum binaural gain and target orientation of a binaural beamformer towards a direction provided by the users’ intention rather than her or his momentary head orientation. Such developments are currently under investigation in several laboratories. They try to estimate the users focus of attention by eye and head movements and even use EEG signals to estimate the subjects preferred target direction (Mirkovic et al, 2015; Brimijoin & Akeroyd, 2016; de Cheveigné, 2016; Grimm et al, 2016). It is still unclear if such a closed-loop system with an attentional steering of the binaural beam might improve the performance of binaural beam formers eventually in real life situations. Since they attempt to replace the neural, binaural signal processing usually taking place in the head (intra-corporally) by an artificial, extra-corporal processing, it is unclear if these systems will perform better than classical binaural enhancement schemes aiming at supporting intra-corporal binaural signal processing.

- Test environments
  The difficulty of predicting the real-life benefit of fixed monaural or even binaural beam formers from lab situations on to real-life conditions uncovers one of the other still unsolved problems in hearing aid fitting and evaluation: What are the most appropriate test situations for lab assessments (including those simple test scenarios accessible to practical hearing aid fitting) in order to best predict real-life performance? Ongoing research in different labs worldwide currently try to answer this question by providing lab test environments with an increasingly high level of preserving environmental and spatial acoustic cues (Seeber & Hafter, 2013; Best et al, 2016; Grimm et al, 2016). They employ, e.g. virtual acoustics (through headphone presentation or free field crosstalk-cancelling presentation with few loudspeakers), simple loudspeaker arrays (using commercially available surround sound formats) or even solutions with very high numbers of loudspeakers surrounding the subjects, i.e. Higher Order Ambiguous or wave field synthesis presentation (Nowak et al, 2013; Best et al, 2015; Grimm et al, 2015). In the ideal case, a perfect reproduction of the real sound field scenario can be achieved (including, e.g. appropriate visual stimulation or a complete virtual reality environment) for a restricted spatial area and for a restricted frequency range with the most sophisticated multiple loudspeaker environments using Wave Field Synthesis (WFS). However, for hearing aid evaluation (and for certain degree also hearing aid fitting), much less effort might be sufficient: an authentic sound reproduction (as opposed to an acoustically correct reproduction) with only a limited sweet spot or sweet range within the reproduction room (i.e. spatial area where the subject maybe located to receive the full simulation effect) might be sufficient. This clearly reduces the technical and financial efforts to be undertaken to reproduce such a virtual acoustic environment. Grimm et al (2016) even could show that the maximum achievable hearing aid performance benefit in lab situations is substantially higher than for real life situations even if a very high amount of experimental effort is invested. This clearly limits the transfer ability even from sophisticated lab results to real life and calls for well-designed, easy-to-realize and standardised lab situations which can be used to show a similar benefit in the lab as for real life with in affordable amount of experimental effort.
- Model-based binaural fitting
  As was the case with binaural noise suppression strategies, binaural signal presentation and amplification strategies may also suffer from a limited transferability from lab results (where audiological diagnostics and hearing device fitting is primarily performed monaurally and with narrow band signals) to real-life situations (where primarily broadband signals are presented to the subjects at both ears concurrently). Recent studies on binaural and spectral loudness summation have shown that a high variability exists across hearing-impaired listeners with respect to the combined loudness impression of both ears especially for broadband signals and especially at high levels. These studies offer a potential explanation why hearing aid users often complain about their hearing aids being too loud in real-life situations even though not such a problem exists when tested monaurally in the laboratory or fitting booth (Oetting et al, 2016). Unfortunately, no current loudness model is able to describe these phenomena in a consistent way. This clearly limits the usage of loudness models for hearing aid fitting and calls for other model concepts than loudness alone in order to better explain the relation between the observed interindividual variability in binaural loudness summation and other auditory functions relevant for hearing aid fitting in hearing-impaired listeners. For this purpose, ''effective'' psychoacoustic processing models (as described in section Neurosensory level) describing the effect of hearing impairment on speech recognition, binaural hearing and other psychoacoustical functions should be extended to describe loudness as well. Currently, loudness is not well modelled by such an approach even though recent ''effective'' processing models at the cochlear level appear promising (Pieper et al, 2016). Such a model concept would be highly desirable to enable advanced fitting of future, binaural, model-supported hearing devices. In the nearest future, at least some approximation for an amplification correction may be individually applied that should depend on the bandwidth and the level at both binaural inputs to a hearing device as suggested by Oetting et al (2016).

Fusion with other communication techniques

The new emerging techniques in consumer electronics with an increased connectivity between and integration of traditionally separate devices, such as, e.g. telephone, stereo equipment, television, fitness trackers will also progressively convert the use case of hearing aids. Thus, it is foreseeable that a hearing aid will be just one part of a multi-functional, multi connectivity-driven assisted communication device of which the limits between a (wireless) headset and a hearing device are more and more diminishing. This provides a number of challenges and changes for hearing aid technology as an integral part of future consumer electronic devices.

- Self-fitting
  Emerging solutions in consumer devices (such as, e.g. smartphones) for persons with a mild-to-moderate hearing loss that require some hearing assistance will be user-centred and will not necessarily be supervised by a professional audiologist. Hence, appropriate self-fitting interfaces will become increasingly popular. The challenge is to provide the appropriate mix between subject-driven comfort requirements (usually low gain at high frequencies and high listening comfort) and prescriptive gain formulas to provide maximum intelligibility in noise (usually requiring a higher gain at high frequency which is only tolerated by the user after some training). Even though such assistive listening concepts may considerably alter the business model for hearing aid provision, the chance is that an early adaptation to listening assistance in consumer electronics will alleviate the problem of providing a sufficient hearing aid much too late in the patients' medical career or lifetime. One possible future development for a mix between objective, model-based suggestions of parameter settings for multi-parameter-controlled hearing aid algorithms and a user-centred assessment of such proposed settings along a ''path of probation'' is described by Pieper et al (2016, this issue): a combination of available models for speech recognition and auditory quality prediction will be used to

objectively generate a set of optimised combinations of parameter settings that represent different compromises across the selected measures. The user should judge along this preselected path in the multidimensional hearing aid parameter space the respective best parameter combination using the so-called MUSHRA drag & drop procedure (Völker et al, 2016). Nevertheless, more research is needed to provide the appropriate fitting approaches and user interfaces that will eventually lead to a high compatibility between signal presentation techniques in consumer electronic devices and in hearing aids.

Conclusions

From the review of current commercial solutions, underlying models of hearing loss for usage in future hearing devices, laboratory studies and emerging future technical solutions for hearing aid functionality it becomes clear that the ultimate goal of hearing aid function (i.e. to provide a normal hearing ability and speech perception function to all users in all environments) is not in reach in the near future. However, substantial and sustainable progress has been achieved in the past as a result of small, but consistent improvements. Such a chain of techniques has provided incremental, but steady increases in user benefit, e.g. in the fields of hearing aid amplification, feedback suppression, dynamic compression, noise reduction and situation adaptation summarised in Table 3.

– Of special importance is the advent of ''true'' binaural hearing aids that promise to provide in the nearest future those comparatively large benefits already observable under restricted laboratory situations to be achieved even in real life – under the assumption that appropriate user-centred control techniques for these algorithms will be available.
– A further major factor is the requirement of a strong individualisation of hearing aid solutions which demands less prescription based solely on thresholds of hearing sensitivity and other ''one size fits all'' philosophy, but calls for an appropriate assessment of the different components of a sensorineural hearing loss (i.e. attenuation loss, compression loss, binaural loss, central resolution loss).
– While the review provided in this article has focussed on technical and perceptual factors to be accounted for in modern hearing aid technology, the implicit assumption is that the user can take full advantage of the features provided at least following some basic counselling. This assumes an appropriate positive attitude and readiness on the user side to utilise the available and emerging solutions which is not automatically fulfilled given the variety in cognitive abilities and lifestyle of the potential users. Hence, the influence of the individual personality of the user (including her or his acclimatisation to using hearing aid technology and his or her cognitive capacity) as well as an appropriate guidance during the rehabilitation process with hearing aids has to be accounted for to provide realistic expectations.

Even though large parts of current hearing aid system technology have not been covered in this review in detail, the authors hope that the current review of audiological, functional and technical aspects of current and future hearing aids may be inspiring for professionals and hearing aid users in order to appreciate all the fascinating developments in the world of hearing aids yet to come.

Notes

1. In a physical or communication theory sense, signal processing in the ear can be described by a chain of several signal processing stages that reflect the ''effective'' processing in the auditory system (which is primarily considered here) while not necessarily taking the actual biological implementation into account. Additionally, physiologic models (Sumner et al, 2002; Zilany et al, 2009, 2014) have gained interest, as they appear suited for a detailed simulation of peripheral hearing loss consequences including effects of a reduced number of available auditory-nerve fibres (Kujawa & Liberman, 2009; Furman et al, 2013) in connection with normal audiograms (concealed or ''hidden'' hearing loss). Since a direct application of such physiologic models to hearing aid processing and fitting is difficult, this class of models will not be considered here.

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