Using limitations of the auditory system to optimize hearing-aid design

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Recent research has demonstrated that people with hearing impairment have limited ability to take advantage of temporal fine structure information (Strelcyk and Dau, 2009; Hopkins and Moore, 2011). This means that they will not be able to fully utilize auditory cues, such as interaural time differences and detailed pitch perception, which rely on such information. On the other hand, this reduced ability can also be used to improve on certain aspects of hearing-aid functionality. One such area is feedback suppression. Many of the latest hearing-aid introductions feature feedback suppression algorithms which apply a slight frequency shift to de-correlate the hearing-aid output from the input and thus minimize the risk of feedback. This paper will review evidence on temporal fine-structure abilities and relate this to how hearing-aid feedback systems can be designed to achieve a dual goal: to optimize the perceived sound quality of the listener with hearing impairment, whilst minimizing the occurrence of feedback.

BACKGROUND

The human ear processes sound using a number of auditory filters into a series of relatively narrow frequency bands. These bands have good or narrow frequency specificity. When a broadband sound comes in, it is band-pass filtered corresponding to the ‘correct’ position on the basilar membrane, which is tonotopically organized. An incoming signal can be considered as a slowly varying envelope superimposed on a rapid temporal fine structure. Information about the envelope is carried by changes over time in the firing rate of the auditory nerve while information about the temporal fine structure is embedded in the phase-locking pattern. When cochlear hearing loss occurs, the ability to use these fast changes, the temporal fine structure, is believed to decrease (Moore, 2007).

A study by Hopkins and Moore (2007) found that hearing-impaired individuals have trouble utilizing temporal fine structure information compared to normal-hearing individuals. In a group of individuals with moderate cochlear loss they tested complex harmonic tones compared with similar tones with all components shifted by ΔHz. For normal-hearing individuals, a shift like this would be perceived as the shifted tone having higher pitch than the un-shifted one with lower harmonics. Both tones, shifted and un-shifted, had a similar envelope repetition rate of F_0. For

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normal-hearing individuals, the smallest detectable shift in frequency was $0.05F_0$. The hearing-impaired group with moderate cochlear loss performed poorer. For most subjects and F_0s the performance was not significantly above chance level even for the maximum shift at $0.5F_0$. Above chance performance was only seen when the hearing loss at the center frequency of the band pass filter was little or none.

The inability to detect shifts in temporal fine structure in individuals with moderate cochlear loss have led to the idea of devising a Temporal Fine Structure (TFS) test (Moore and Sęk, 2009). The underlying hypothesis of the TFS test is that TFS information might be useful in deciding the most appropriate speed of compression for a hearing-impaired individual. The test should be applied quickly and reliably in a clinical situation.

An important note is that even slight or mild hearing losses experience issues with temporal fine structure. Therefore, it is expected that any hearing loss will show a reduced of sensitivity to temporal fine structure information (Ardoint *et al.*, 2010).

ERIKSHOLM RESEACH USING THE TFS1 TEST

Researchers at the Eriksholm Research Centre have explored whether the TFS1 test could be used to unveil differences in normal-hearing and hearing-impaired persons with mild to moderate hearing loss (Hietkamp *et al.*, 2010).

The TFS1 test

The TFS1 test used in the study was similar to the test described in Moore and Sęk (2009)[1]. The test paradigm is based on an A/B comparison of two sequences or intervals of stimuli: one where the F_0 harmonic is held constant in all presentations and one where every other stimulus is shifted by xF_0 (Fig. 1). The task is to select the sequence that contains the shifted stimuli. After each response, the participants get visual feedback whether the response is correct or incorrect.

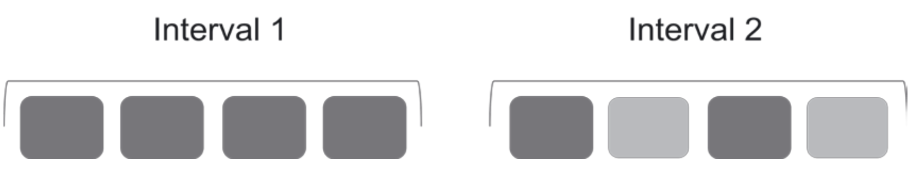

Fig. 1: The task for the participants was to choose the interval with fluctuating stimuli. The order of presentation was randomized for each trial. Test parameters for each test condition are adjustable and include the fundamental frequency (F_0), center frequency (F_c), and harmonic.

[1] Software for the TFS test is available from University of Cambridge's homepage at http://hearing.psychol.cam.ac.uk/

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The stimuli were band-pass filtered around one of the harmonics of the tone (e.g., the 5th or 11th). The auditory system does not appear to resolve harmonics above the 8th harmonic. This means that all components within the pass band were unresolved when the filter was centred at $11F_0$. Consequently, the excitation patterns were very similar for un-shifted and shifted stimuli. The band-pass filter had a central flat region width a width of $5F_0$ and skirts that decreased by 30 dB/octave. These are relatively shallow slopes which ensure minimal changes in the excitation pattern as components move in and out of the pass band.

The TFS1 test is an adaptive test in the sense that the size of the shift, the x in xF_0, is adaptive based on the response from the test person. This means that for individuals who are able to detect the difference in the stimuli, the test is able to calculate a threshold for the amount of perceivable shift in F_0 for a given F_0 and a given harmonic. Thus, the first presentation will include stimuli with the maximum shift. If this is correctly identified, the next presentation will include a smaller shift and so forth. The threshold is then calculated by averaging a specified number of the last reversals. Furthermore, for the study at Eriksholm the reversal rate had a maximum standard deviation (SD) of 0.15 to be deemed reliable. If the SD exceeded this level, another round of testing was included. The test was performed with speech at 20 dB SL and noise at 5 dB SL; the latter was introduced to mask any differences between the stimuli that were not attributed to the difference in temporal fine structure.

Of the hearing-impaired participants many were unable to complete the adaptive procedure. Here a percent correct method was used where the number of correct response at the maximum shift (here $0.5F_0$) was calculated.

Before initiating the test, the participants were trained in the procedure. As the sequences used in the test are of fairly short duration (0.2 s) and with short inter-stimuli pause (0.3 s), the test also requires a certain level of concentration and some short term memory. The training session differed significantly from the test sessions as it only needed to provide a baseline understanding of the test paradigm. For the training session the comparison was between two intervals with the same F_c, but with different F_0s or different repetition rates. Hence, in the training session, the fluctuation was between an interval consisting of four stimuli with an F_c of 1100 Hz and an F_0 of 100 Hz, and an interval that shifted between stimuli with F_0 of 100 Hz and F_0 of 150 Hz. This change in F_0 around the center frequency of 500 Hz changes the repetition rate between the harmonics from 300, 400, 500, 600, and 700 Hz to 100, 350, 500, 650, and 800 Hz. For the test condition, the focus is on the shift of F_0, for the training the focus is on the spread (size) of F_0. In the training session, all participants were able to perceive a difference between the presented sequences, and for all participants a threshold was obtained. The stimuli used for training did not contain a filter, nor was any background noise introduced, leaving greater differences between the stimuli presented in the fluctuating interval.

The baseline for the comparison was five harmonics of F_0 around a center frequency of F_c. Table 1 shows the F_cs and F_0s as well as the maximum shift in F_0 that was used in the study. The 'cleanest' comparison is the one made using the 11th harmonic, cf. its un-resolved nature in the auditory system.

	5th harmonic			11th harmonic		
F_c (Hz)	500	1000	2000	1100	2200	4400
F_0 (Hz)	100	200	400	100	200	400
Max F_0 shift (Hz)	50	100	200	50	100	200

Table 1. The test conditions varied on harmonic, center frequency (F_c), and fundamental frequency (F_0).

All participants were tested using the adaptive procedure, but if they were unable to go below the maximum shift of $0.5F_0$, a percent-correct method was used instead.

Results and conclusions from the study

The results from the study showed that hearing-impaired participants were significantly poorer at detecting the interval with shifted stimuli embedded in it. This was most clear for the test conditions using the 11th harmonic and center frequencies between 1100-4400 Hz. Here the division between hearing-impaired and normal-hearing participants was near binary. Only very few were able to reach a threshold and most scored no better than chance at the maximum shift of $0.5F_0$. The results also suggest that hearing-impaired individuals are better able to perceive differences in the repetition rate of F_0 than differences due to a shift in F_0. Thus, the hearing-impaired listeners seemed more sensitive to changes in envelope than to changes only in the temporal fine structure.

FREQUENCY SHIFTS IN HEARING INSTRUMENT SIGNAL PROCESSING

Using a frequency shift in signal processing in hearing instruments has both advantages and drawbacks. A great advantage is that the frequency shift alongside the feedback cancellation system decreases the susceptibility to entrainment, when external tonality is mistaken for internally generated feedback. A drawback lies in the artefact that may be perceivable when the difference between a shifted and a non-shifted signal is audible. This is most likely to occur in open solutions for individuals with fairly good low frequency hearing. The greater the shift, the more audible it can potentially be. However, greater shifts also provide more efficient de-correlation of hearing-aid output.

These attributes demands some consideration when implementing a frequency shift. In particular, how great a frequency shift must be used and for which inputs are the addition of a frequency shift necessary.

FEEDBACK SHIELD ON THE OTICON INIUM PLATFORM

The Oticon anti-feedback strategy builds on the principles of dynamic phase inversion or feedback cancellation (DFC) for destructive interference between a feedback signal and a cancellation signal produced by the anti-feedback system. In the anti-feedback system, the feedback path is constantly measured. Rapid changes in the feedback path, e.g., when the sound environment is very dynamic or when there are physical movement/changes close to the hearing instruments, warrant rapid and frequent measures of the feedback path. When the feedback path is more stable or only changes slowly, measurements of the feedback path are needed less often.

When the system is whistling or close to whistling, the phase inversion is put to its full use. The feedback path is measured and a mirror or phase-inverted version of the feedback signal is imposed to cancel out the feedback. This may seem simple enough, but the precision of the inverted signal is paramount. The phase inversion must not only be precise, but also fast to cancel feedback out even before it is actually perceivable as feedback or whistling. To be successful in removing the feedback loop, the signal needed to cancel out the feedback must be equally complex.

However, under some conditions or in some environments with high auto-correlation, updating the feedback path is less beneficial for the anti-feedback system because of the risk of disturbing the sound quality. When this is the case, feedback shield maintains the last good feedback path estimate and cancels out feedback based on this estimate. In other situations, it is safe and wise not only to keep updating, but sometimes also to increase the frequency of updates. When doing this, a frequency shift of 10 Hz is enabled to render the system less susceptible to tones in the environment, thus making it easier to correctly identify internally generated feedback loops and not mistakenly attempt to cancel out external tones. The frequency shift works by shifting the entirety of the input signal above a given transition frequency 10 Hz upwards. In shifting the majority of the signal, the envelope is kept intact and only the fine structure is affected.

Because the output of the hearing aid is slightly different from the input due to the frequency shift, potential acoustic leakage will not line up with the input and create a feedback. Thus, processed sound going back to the microphone is more easily distinguished from input from the environment. Using a frequency shift is a very effective method for decreasing system-sensitivity to tonal inputs, thus allowing other parts of the system to update.

Three modes are implemented in the current feedback system to accommodate for the changing/alternating need for using frequent updates, and thus the addition of the frequency shift. The shift between the modes is based on input from two detectors: a howl detector and a tonal detector (Fig. 2).

The main job of the anti-feedback system is to prevent audible feedback or whistling from happening. Thus, the primary task is the detection of potential audible feedback. Based on a calculation of the auto-correlation in the output, the howl detector will determine whether the system will need to adapt and go to a more aggressively updating mode or whether the choice of mode can be based on the input from the tonal detector.

A detection of audible feedback will enable fast updates to the DFC system supported by the 10-Hz frequency shift.

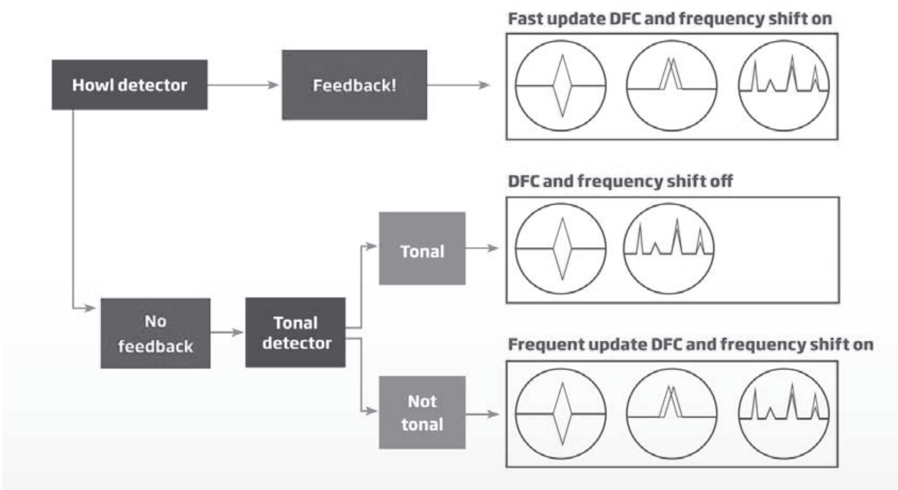

Fig. 2: Diagram of how the different detectors and modes are configured in the anti-feedback system.

If audible feedback is not present, the tonal detector determines which mode should be used based on the presence of tonality in the environment. Tonality is characterized by repeatable, harmonic content such as acoustic stimuli like speech and music. This type of content can be mistaken for feedback by the anti-feedback system, so that extra care needs to be asserted to avoid suppressing any valuable information or creating a loop in phase with the input causing additional feedback.

When the content of the input signal is tonal, the system will enable a less aggressive mode. Here the DFC behaves in a stable manner in the sense that the last updated feedback path filter/estimate is applied to the input signal. The assumption is that the acoustic properties around the hearing aid will not have changed and thus this filter will still be applicable. When the DFC is not being updated, the frequency shift is not relevant and will be disabled. When the content of the input signal is *not* tonal, the Inium feedback shield will be in dynamic mode. The risk of degrading either speech or music is no longer present, and therefore, the DFC will allow more frequent updates of the filter according to the subtle changes in the feedback path. The frequency shift is enabled in dynamic mode to ensure robust feedback path estimation.

An internal test compared the previous RISE2 platform to the Inium platform on a feedback 'stress' test of hearing-instrument performance. The test setup consisted of a head-and-torso simulator (HATS) with hearing instruments mounted on the ears and a mechanical arm moving to and from the ear. A test sequence featuring rising pure tones, clicks, classical flute play, and other very tonal sound content was played, processed by the hearing instrument, and recorded in the ear of the HATS.

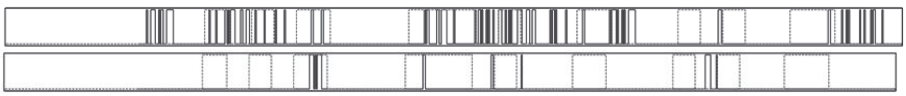

Fig. 3: Results from feedback stress test, RISE2 (top) versus Inium (bottom). The dotted lines indicate movement of a mechanical arm to and from the ear of the HATS and blue lines indicate identification of audible feedback.

A 'golden-ear' listener evaluated the occurrences of feedback in the recordings with the two instruments. The test showed that the anti-feedback system on the Inium platform reduced the number of audible instances of feedback by 80%.

CONCLUSION

The introduction of a small frequency shift can help improve an anti-feedback system by de-correlating the input without (too) many sacrifices to the requirements/demands for sound quality in hearing-impaired listeners. The implementation of such a shift plays a great role, as research indicates that a shift in the frequency content of the hearing-instrument output may be audible in the form of acoustic artefact and disturbing to the listening experience. The trade-off between the advantages and downsides to the use of the technology warrants diligence when choosing what input can or cannot be added to the signal path. The improvement of the Inium-based feedback shield has yielded great improvements to the frequency of occurrence of feedback in Oticon hearing instruments; a reduction in audible feedback by 80% was seen in an instrument stress test performed by a 'golden-ear' listener.

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