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A model for prediction of own voice alteration with hearing aids

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For normal hearing persons, own voice perception is a complex function of airborne sound transmission and bone-conduction transmission; the dynamics of the system depend on the motion of the mouth and, consequently, the articulation. The sound transmission between mouth and the cochlea can be modelled as a time-depended, non-linear filter. For persons with hearing aids, the system becomes even more complex. First, the open-fit receiver, vented or closed earmold affect both the airborne and bone-conducted transmission due to acoustic filtering and the occlusion effect. Second, the hearing aid amplifier influences the airborne sound; this influence is even more difficult to predict due to microphone settings and hearing aid compression. A model for own voice sound transmission was devised and used to predict changes in own voice perception subsequent to hearing aid fitting. The model is a combination of ear-canal acoustics and active amplification. The predictions were verified in 30 subjects fitted with hearing aids. The hearing aid types included open fittings, classic BTE and ITE hearing aids. The model gave good prediction of the own voice ear canal sound pressure alteration caused by the hearing aid. However, this sound pressure alteration do not predict subjectively rated own voice problems.

INTRODUCTION

The alteration of one’s own voice subsequent to fitting of a hearing aid is one of the top ten factors affecting overall satisfaction with hearing aids (Kochkin, 2010). In the MarkeTrak VIII (Kochkin, 2010), 9% of the total sample reported dissatisfaction of the sound quality that relate to sound of voice and 14% that relate to chewing/swallowing sound. Even if this indicates that the majority of hearing aid wearers do not have own voice problem, in a minority of users this is a problem that eventually leads to non-use of the hearing aids. With the introduction of hearing aids using open fittings, the ear canal is in large part open and the occlusion effect is avoided (Stenfelt and Reinfeldt, 2007; Kiessling *et al.*, 2005). There is evidence that this reduction in occlusion also leads to better quality of own voice compared with occluding devices, at least for experienced hearing aid users (Taylor, 2006; Gnewikow and Moss, 2006; Kiessling *et al.*, 2005). However, due to feedback problems, open fittings can only be used with limited amplification and ear mold devices is still a common solution for hearing aid users.

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Although it is hypothesized that users with low-frequency hearing thresholds greater than 40 dB HL are unlikely to have occlusion effect problems (Dillon, 2001), still a large part of the hearing aid users may experience own voice problems due to the occlusion effect. Beside open fittings, ventilation tubes in the hearing aids are used to increase wearing comfort and reduce occlusion (Kiessling *et al.*, 2005; Kuk *et al.*, 2005). These ventilation tubes come in different designs (Kuk *et al.*, 2009; Kiessling *et al.*, 2005), but as a rule of thumb, the greater the diameter and the shorter the length of the ventilation tube, the less occlusion. However, this comes at the cost of more feedback problems. This trade-off between occlusion and feedback problems result in occlusion effect- related own voice problems in some hearing aid users, as seen in the latest MarkeTrak study (Kochkin, 2010**)**. It should be noted that own voice problems that do not originate in the occlusion effect are reported by hearing aid users. However, the occlusion effect is still important for own voice perception (Laugesen *et al.*, 2011).

The aim of the current study is to provide a model predicting alteration of the own voice when fitting a hearing aid. This model is based on three parts: (1) air and bone conduction transmission of the own voice, (2) the ear canal sound pressure change due to the occlusion effect, including effects of ventilation tubes (influencing bone conducted sound), and (3) the gain and filtering of the airborne sound due to the hearing aid gain and ear mold (influencing the air conduction sound). The three parts are described below and the model is evaluated in a group of hearing aid users.

AIR AND BONE CONDUCTION TRANSMISSION OF OWN VOICE

When stimulation is by air conduction, sound transmission is rather straightforward with the signal entering the external ear canal and transmitted to the inner ear via the middle ear ossicles. Bone conduction transmission is more complex involving sound radiation in the external ear, inertial effects of the middle ear ossicles and cochlear fluid, as well as alteration of the cochlear space and direct pressure transmission to the cochlea (Stenfelt and Goode, 2005). The relative importance of these different contributors is not entirely explained and depends on frequency, type and place of stimulation.

During speech production, vocal chords, oral and nose cavities, tongue, teeth and lips are used to form the signal. The speech production does not only produce an airborne speech signal but also generates bone conduction sound, depicted in Fig. 1a.This dual sound generation is the way most people experience bone conducted sound. When listening to a recording of their own voice people are often struck by the difference between the voice characteristic of the recording and the way they normally perceive their voice. The cause of the difference is that the recording only picks up the airborne portion of the sound whereas we hear our own voice through both airborne and bone conduction sound. According to Békésy (1949), the two components of our own voice, the air and bone conduction parts are of approximate equal magnitude. However, the air and bone conduction parts of our own voice are expected to be a function of both frequency and the voice production itself. From a signal transmission point of view, the airborne sound can be seen as a signal source

that is filtered by a time-dependent filter. Similarly, the bone conduction part can be viewed as a generator filtered by a time-dependent filter. Since the two outputs for the air and bone conduction signals of our own voice differ, the two time-dependent filters associated with the outputs differ.

In an attempt to describe the relative difference between the air and bone conduction parts of our own voice, Reinfeldt *et al.* (2010) measured the ear canal sound pressure during sound production of ten phonemes; four vowels (Fig 1b) and six consonants (Fig 1c). The bone conduction part was obtained by using an ear muff that gave no or minimal occlusion effect thus removing the air conduction contribution to the ear canal sound pressure. The estimate of the air conduction part was obtained by measurement of the airborne sound outside the ear and relating that to ear canal sound pressure in front of the eardrum by individually measured transfer functions. The results show that the relative contribution of bone and air conduction for the own voice depend on frequency, but also that there are substantial differences between the ten phonemes.

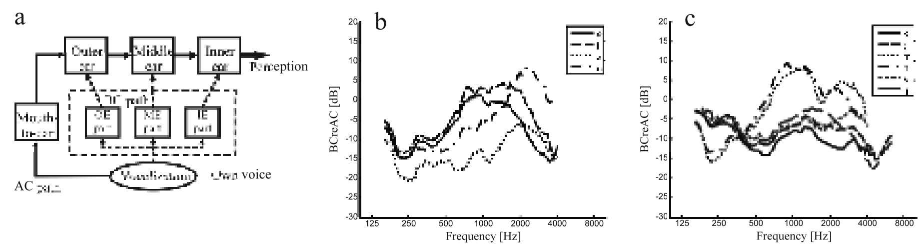

Fig. 1: a) Air and bone conduction components involved in sound transmission of the own voice. b) and c) The relative contribution of bone and air conducted sound during sound production measured as the ear canal sound pressure. Figures are modified from Reinfeldt *et al.* (2010).

It should be noted that the results in Fig 1 is not the contribution for the perceived own voice but the contribution by air and bone conduction of the sound pressure in the ear canal. To get an estimate of the relative importance from bone and air conduction of the perceived own voice, the importance of the bone conduction pathway for the ear canal should be related to other pathways, see Fig. 1a. The author is not aware of such analysis when the source is the own voice (considered as a distributed source) but estimates of the importance from external stimulation indicates that the ear canal component of bone conducted sound is 5 to 10 dB below that of other contributors (Stenfelt, 2006; Stenfelt *et al.*, 2003). Consequently, to get an estimate of the relative contribution for the perceived own voice, the results in Fig. 1 should be shifted 5 to 10 dB.

MODEL OF OCCLUSION EFFECT

A model of the occlusion effect based on ear canal impedance, terminating impedances (eardrum and middle ear and occlusion device), and canal surface area for sound generation was described in Stenfelt and Reinfeldt (2007). The structure of the model is depicted in Fig. 2. The model is divided in two parts, one that describes the bony part of the ear canal (left hand side of Fig. 2) and one that describes the soft tissue and cartilage part of the ear canal (right hand side of Fig. 2). Structurally, both parts are identical but the parameter values differ between the parts. The four impedance boxes Z_1 to Z_4 describe T-lattice impedance network of the ear canal. The parameters of the boxes, indicated in Fig. 2, are functions of ear canal length, width, and position of the occluding device. The terminating impedance, Z_{OCC}, is either the radiation impedance of the open canal or the impedance of the terminating device, that could include a ventilation tube as in a hearing aid mold. The other end is modeled by terminating impedance for the eardrum and middle ear while the bone conduction sound source is modeled by two current generators, one in the bony part and one in the cartilage part of the canal. The details and parameter values of this model is presented in Stenfelt and Reinfeldt (2007).

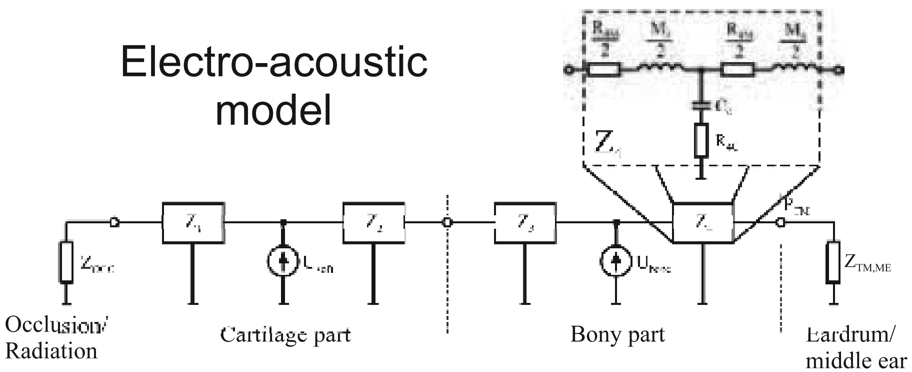

Fig. 2: An electro-acoustic model for the occlusion effect. The figure is modified from Stenfelt and Reinfeldt (2007).

MODEL OF OWN VOICE PERCEPTION WITH AMPLIFICATION

As stated previously, beside the occlusion effect, the hearing aid gain influences the sound of own voice. The occlusion effect influences the bone conduction pathway while the hearing aid gain influences the air conduction pathway. In addition, the ear mold, open, closed, or with ventilation tubes act as an acoustic filter for the incoming airborne sound. Consequently, the own voice is changed according to these factors.

The model for own voice perception is based on the change in ear canal sound pressure when a hearing aid is positioned. This is not equivalent to the alteration of the perception of the own voice, and the result of the model needs to be adjusted in a

second step to give an estimate of perceived change. First, the own voice is divided into its air and bone conduction components. The bone conduction component is affected by the occlusion effect, according to the model in Fig. 2. The result of such calculation for a hearing aid positioned 7 mm down the ear canal (approximately just beyond the second bend of the ear canal) incorporating a 5 mm ventilation tube of different diameters (from 0 [closed] to 4 mm) is shown in Fig. 3. It is clear that the greatest low-frequency occlusion effect is obtained in a closed fit and the occlusion effect decreases with increase in ventilation tube diameter. With ø4 mm ventilation tube the occlusion effect has almost vanished. The effects seen at frequencies above 2 kHz is due to changes in canal resonances: these depend on residual ear canal length and terminating impedance.

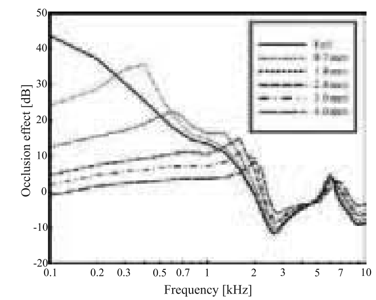

Fig. 3: Occlusion effect for bone conducted sound according to the model in Fig. 2. The earmold is positioned 7 mm down the ear canal and the ventilation tube is 5 mm long.

The air conduction component is altered primarily by the gain of the hearing aid but there is also an acoustic filtering effect due to the earmold. Moreover, the directivity of the microphone is of importance since the sound source of the own voice differ from that of a sound source when listening. This amplified air conduction sound is added to the modified bone conduction sound in the ear canal producing the altered own voice sound pressure.

EVALUATION OF THE MODEL

The model of the own voice, i.e. the alteration of the air and bone conduction sound in the ear canal as described above, was evaluated in 30 hearing aid users. The types of hearing aids used include open fitting as well as BTE with earmold and ITE hearing aids. All measurements were conducted in an anechoic room with air conduction stimulation presented through a loudspeaker 1 m in front of the subject at the level of the head. All ear canal sound pressure measurements were done with an Etymotic Research ER-7C probe microphone system with the probe tube opening approximately 5 mm in front of the eardrum. With that setup, the hearing aid characteristic was obtained with three ear canal sound pressure measurements while stimulation was from the loudspeaker: (1) open ear canal, (2) passive hearing aid in ear canal (hearing aid off), and (3) active hearing aid in ear canal (hearing aid on). From these measurements the filtering function and gain function of the hearing aid was obtained. The source signal was the international speech test signal (Holube *et al.*, 2010) and it was presented at a level of 70 dB SPL.

The influence of the occlusion effect was obtained by bone conduction measurements. A Radioear B-71 transducer was attached at the forehead providing bone conduction stimulation while measuring the ear canal sound pressure with open ear canal and hearing aid positioned (amplification off). This measurement gave an estimate of the occlusion effect caused by the ear mold with or without ventilation.

Figure 4 illustrates the results from the estimates provided by the model. No individual data are provided, only the average of the hearing aid gain for the 30 subjects measured (dotted line in Fig. 4a). Using this amplification function as the hearing aid gain, the other curves in Fig. 4a shows the alteration of the own voice caused by a hearing aid as measured in the ear canal. For the modeling, the speech production is based on an average of the three utterances /k/, /a/, and /s/. Moreover, the hearing aid earmold is modeled as in Fig. 3 with a position 7 mm into the ear canal and with a ventilation tube length of 5 mm.

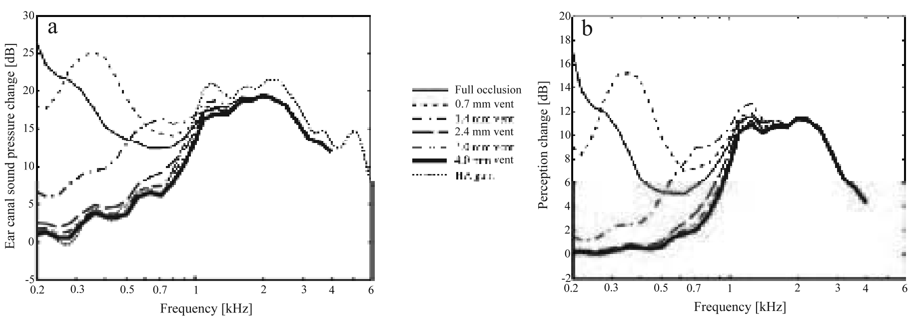

Fig. 4: a) The estimated alteration of the own voice ear canal sound pressure according to the model and hearing aid gain as shown in the dotted line. b) The data from a) recalculated to perception change of own voice.

The data in Fig. 4a is the sound pressure change in the ear canal. This is different from the perceptual change of the own voice, primarily due to bone conduction transmission directly to the inner ear as illustrated in Fig. 1a (Stenfelt and Goode, 2005). As indicated previously, the ear canal transmission pathway for bone conducted sound is 5 to 10 dB lower than other pathways (Stenfelt, 2006; Stenfelt *et al.*, 2003). As a consequence, the perception change of the own voice is different from that seen as a sound pressure change in the ear canal. Using an estimate of 10 dB between the ear canal component and other components for bone conduction perception, the data in Fig. 4a are recalculated to an estimate of the perception

change of the own voice caused by a hearing aid (Fig. 4b). It should be noted that the estimates in Fig. 4 is based on a hearing aid gain function also shown in Fig. 4a (dotted line). Any other hearing aid gain and also position of the aid results in different estimates of the own voice change than those shown in Fig. 4.

DISCUSSION

The data from the evaluation of the model are presented as average gain for the hearing aids tested. For the individual, the model provided good estimates of the own voice changes when the hearing aid characteristics were considered (gain, position in the ear canal, and ventilation tube characteristics). Accordingly, the prediction of own voice change with hearing aid on the 30 subjects participating in the evaluation was generally within 5 dB from the measured ear canal sound pressure within the frequency range 0.2 to 4 kHz. More details on the individual evaluation will be presented elsewhere.

As shown in Fig. 4, a closed ear mold increases the low frequency content of the own voice. This low-frequency increase, often referred to as the occlusion effect, is reduced with the introduction of a ventilation tube. According to the model with a 5 mm long straight tube, the occlusion effect is almost non-existing with tube diameters more than 2 mm. However, it can also be seen that, due to resonances in the tube itself, a small tube diameter may cause worse occlusion effect problems with own voice than with full occlusion (no ventilation). This may be more of an academic problem since there is usually some leakage between earmold and ear canal that reduces the resonance effect of a small ventilation tube.

In a study investigating the size of ventilation tube required to achieve acceptable own voice perception Carle *et al.* (2002) found a relation between ventilation tube size and the impedance of the eardrum. The impedance of the eardrum is modeled by $Z_{TM,ME}$ in Fig. 2. The finding that a greater compliance requires greater ventilation tube size could not be corroborated by the model. On the contrary, the model predicts less occlusion effect with lower impedance (greater compliance) of the eardrum, opposite to that reported by Carle *et al.* (2002).

During the evaluation process the 30 participants were instructed to read a text while measuring ear canal sound pressure produced by the own voice. This was done for two conditions, without hearing aid and with hearing aid on. No specific instructions for the level of vocalization were given. In order to compensate for variations in vocalization effort between conditions, a reference microphone positioned 20 cm in front of the mouth recorded the voice. When analyzing the sound production it was found that the vocalization effort reduced by 2 to 4 dB when the hearing aids were on. Consequently, the hearing aid influences the own voice production by reducing the vocalization effort.

The occlusion effect can be an origin for own voice problems, but own voice problems can be present in hearing aid users with no occlusion effect (Laugesen *et al.*, 2011). The alteration of the own voice in a hearing aid user with high frequency gain of 20 dB is shown in Fig. 4. That gain result in an estimated perceptual change of about 10 dB between 1 and 3 kHz. Accordingly, the change of the own voice due to the hearing aid is less than the change of other voices, e.g. a voice by a significant other. Therefore, own voice problem cannot solely be explained by great alteration of voice characteristics due to change of spectral content but small changes can, for some subjects, result in severe disturbance of own voice perception.

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Comparative evaluation of cochlear implant coding strategies via a model of the human auditory speech processing

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Traditional cochlear implant (CI) coding strategies present some information about the waveform or spectral features of the speech signal to the electrodes. However, neither of these approaches takes the cochlear traveling wave or the auditory nerve cell response into account, though these are given in acoustic hearing. Therefore, a new CI coding strategy based on an auditory model including the above mentioned properties of the healthy cochlea was evaluated and compared with an n-of-m-coding strategy, in which n electrodes out of m possible electrodes are stimulated in each stimulation cycle. The selection of the n electrodes is based on the n highest spectral maxima of the momentary signal. Simulated electrical output of both CI coding strategies served as input to a model of the electrically stimulated auditory system, which consisted of an auditory nerve cell population. The nerve cells generated delta pulses as action potentials in dependence on the spatial and temporal properties of the electric field produced by the electric stimuli. This model is used to predict CI user performance in terms of speech intelligibility and pitch discrimination for both coding strategies. Furthermore, an additional model of normal hearing is presented, the output of which is compared to the neural representation resulting from the modeled CI stimulation. We will show under which circumstances and to what extent an auditory model based coding strategy may outperform a traditional CI speech coding algorithm.

INTRODUCTION

Speech recognition performance in noise as well as the ability to discriminate pitch exhibits a high variation in cochlear implant (CI) users. The most probable origins of these differences are degenerative functional changes of the auditory nerve and dissimilarities between the used speech coding strategies.

To describe the quantitative relation between parameters of the auditory processing and speech perception with CIs, a model of the electrically stimulated auditory nerve (Hamacher, 2004), has been modified, and speech intelligibility is simulated for a

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