Modelling, measurement and optimization of passive and active custom-made hearing protectors

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Dank

Ziehier, vandaag, voor u, mijn thesis. Natuurlijk ben ik blij vier jaar onderzoek te kunnen gieten in een boek, compleet met kaft en inhoudstafel. Maar omdat ook een kritische houding geen kwaad kan, wil ik niet nalaten enkele woorden toe te voegen naar het voorbeeld van de filosoof Bertrand Russell. Zijn ‘Geschiedenis van de westere filosofie vanuit de politieke en sociale omstandigheden van de Griekse oudheid tot in de twintigste eeuw’ begint hij immers met het statement dat elk boek(je) het resultaat is van welbepaalde keuzes die de auteur maakte. Hoe weldoordacht en verantwoordbaar deze ook zijn, het blijven keuzes waarvoor alternatieven bestaan. In de pagina’s die volgen heb ik geprobeerd mijn onderzoek coherent, degelijk en volledig te rapporteren, maar ik besef terdege dat iemand anders over hetzelfde onderwerp waarschijnlijk een minder of meer verschillend verhaal had verteld. Ik ben ervan overtuigd dat net deze gezond relativiteit der dingen een belangrijke stimulans is om de wetenschappelijke wereld draaiende te houden.

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Een handdruk,

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Samenvatting

In de industrie worden onder meer gehoorbeschermers gebruikt om lawaaislechthoornheid te voorkomen, maar ondanks deze en andere voorzorgsmaatregelen blijft werkgerelateerd gehoorverlies persisteren. Dit suggereert dat gehoorbeschermers op de werkvloer niet de geluidsreductie realiseren die wordt voorspeld op basis van laboratoriumtesten. Een mogelijke oplossing bestaat erin ook in situ metingen uit te voeren, bijvoorbeeld door middel van de MIRE-techniek (Microphone In Real Ear). Dit is een snelle en betrouwbare methode die het geluidsdrukniveau achter de gehoorbeschermer bepaalt.

Het hier beschreven onderzoek focust in eerste instantie op de implementatie van deze werkwijze voor op-maat-gemaakte gehoorbeschermers uitgerust met testkanaal. Bij normaal gebruik is dit kanaal uiteraard afgesloten, maar ter controle kan een meetmicrofoon worden ingebracht die dan via het testkanaal het geluidsdrukniveau in de gehoorgang opmeet. Dit ontwerp laat snelle en stabiele metingen toe, de hamvraag is echter hoe het geluidsdrukniveau ter hoogte van het trommelvlies kan berekend worden uit metingen ter hoogte van het testkanaal.

Deze relatie, verder transfer functie genoemd, wordt op verschillende manieren geanalyseerd. Ten eerste wordt het geluidsdrukniveau simultaan gemeten ter hoogte van het trommelvlies en de MIRE meetmicrofoon door gebruik te maken van een hoofd-en-torso-simulator. De zo gevonden transfer functies blijken stabil en reproduceerbaar: numerieke FDTD (finite-difference time-domain) modellen tonen verder dat ze bepaald worden door duidelijk identificeerbare kenmerken van gehoorbeschermer en gehoorgang.

Wanneer in een tweede fase analoge metingen bij proefpersonen worden uitgevoerd, blijken de transfer functies hetzelfde patroon te volgen maar hun exact verloop in functie van de frequentie verschilt substantieel tussen geteste oren. Het is echter wel mogelijk via geïndividualiseerde FDTD modellen de geluidsdrukfluctuaties in een specifiek oor te voorspellen. Deze simulaties zijn te rekenintensief voor praktische doeleinden en dienen daarom als basis voor het ontwikkelen van toepasbare filterbenaderingen.

Naast een correcte demping is consequent gebruik cruciaal voor succesvolle implementatie van gehoorbeschermers. In praktijk blijken communicatieproblemen een belangrijke reden om de beschermering niet continu te dragen. Om dit te voorkomen worden nieuwe technologieën toegepast in zogenoemde ‘verbeterde’ gehoorbeschermers die toelaten de attenuatie automatisch aan te passen aan het achtergrondlawaaï in tegenstelling tot de klassieke beschermers met een constante demping. In het tweede luik van dit project wordt spraakverstoring onder deze verbeterde gehoorbeschermers vergeleken met klassieke oordopjes en dat voor normaalhorende proefpersonen. Uit de analyses blijkt enerzijds dat naargelang het soort achtergrondlawaaï bepaalde innovaties wel degelijk een meerwaarde kunnen bieden. Anderzijds kan spraakverstoring alleen verbeterd worden als het ingangssignaal niet te zeer vervormd wordt en de voordelen van binaural horen gevrijwaard blijven.
Summary

In industry, hearing protectors are often implemented to prevent noise-induced hearing loss, but despite these efforts occupational hearing loss seems to persist. This can partially be explained by the fact that the use of protectors at the workfloor might substantially vary from laboratory conditions. Therefore it seems advisable to complete laboratory testing with in situ measurements. In this regard, the MIRE approach (Microphone In Real Ear) offers a quick and reliable way to assess the sound pressure level behind the hearing protector with a microphone.

The first aim of this research project is to investigate the implementation of the MIRE technique for custom-made hearing protectors designed with a test bore. In normal wearing conditions, this bore is closed, but for verification a MIRE measurement microphone can be slided inside to measure the sound pressure in the ear canal. This setup allows quick and stable measurements, but the question arises whether the sound pressure at the eardrum can be derived from levels obtained at the test bore.

This relationship, henceforth called transfer function, is studied in different ways. First, the sound pressure level is registered simultaneously at both points of interest by using a head-and-torso-simulator (HATS). The hence retrieved transfer functions appear to be stable and reproducible; furthermore numerical FDTD (finite-difference time-domain) models show that their morphology depends on clearly distinguishable characteristics of earplug and ear canal.

In the second stage, transfer functions are measured on human subjects in a similar way. They seem to follow the same global pattern, but vary substantially with respect to their exact configuration in function of the frequency. Nevertheless, individualized FDTD models are able to predict the transfer function for a specific ear. However, these simulations require too much calculation time for practical applications, conversely they can serve as a base for workable filter approaches.

Effective implementation of hearing protectors not only requires appropriate attenuation but also consistent use. By contrast, protectors are not seldom removed to facilitate communication. In this regard, new technologies are implemented in so-called augmented protectors so that the applied attenuation is automatically adapted to the background noise in contrast to the classical protectors with constant attenuation over a considerable range of input levels. The second part of this project compares speech intelligibility between those augmented and classical earplugs for normal-hearing subjects. The analyses show that certain innovations might indeed be beneficial depending on the type of background noise. Conversely, speech perception can only be ameliorated if distortion is avoided and if the positive effects of binaural listening are retained.
Résumé

Malgré une utilisation fréquente de protecteurs auditifs en milieu industriel pour prévenir la perte auditive due au bruit, on constate que cette perte auditive persiste au travail. Cela peut s’expliquer par l’écart qui existe entre les conditions d’utilisation dans les laboratoires et les conditions réelles sur le terrain. Une solution à envisager serait d’effectuer des tests complémentaires in situ par le biais de la méthode MIRE (Microphone In Real Ear - Microphone placé dans une oreille réelle). Cette méthode est une technique efficace et fiable qui permet de mesurer la pression acoustique derrière le protecteur auditif dans l’oreille-même du travailleur.

Ce travail de thèse de doctorat a pour premier objectif d’évaluer l’implémentation de la méthode MIRE pour les bouchons d’oreille sur mesure dotés d’un canal de mesure. Normalement ce canal traversant le bouchon d’oreille est fermé, mais une sonde microphonique peut être introduite, permettant ainsi la mesure objective de la pression acoustique dans le conduit auditif. Cette méthode permet une mesure rapide et stable, mais la question est de savoir comment on peut calculer la pression acoustique au tympan sur base des données au niveau du canal de mesure.

Cette relation - que l’on appelle dorénavant ‘fonction de transfert’ - est analysée de différentes façons. En première instance le niveau de pression acoustique est mesuré simultanément aux deux endroits concernés à l’aide d’un simulateur tête et torse (HATS - Head And Torso Simulator). Les fonctions de transfert ainsi trouvées se révèlent stables et reproductibles. Des modèles FDTD numériques (finite-difference time-domain) montrent en outre qu’elles sont déterminées par des caractéristiques clairement identifiables du protecteur auditif et du conduit auditif. Lorsque dans une deuxième étape on réalise des mesures analogues sur des personnes, on constate que les fonctions de transfert suivent le même schéma global, mais que le schéma exact en fonction de la fréquence varie substantiellement par rapport aux sujets testés. Toutefois il est possible de prédire les fonctions de transfert pour une oreille spécifique en utilisant des modèles FDTD individualisés. Le temps de calcul nécessaire pour ces simulations est trop élevé pour qu’on puisse les utiliser à des fins pratiques, mais elles peuvent servir comme base pour le développement d’une approche avec filtre acoustique.

Une attenuation adéquate ne garantit pas en soi le succès de l’utilisation des protecteurs auditifs. Le port continu des protecteurs est également essentiel. Cependant, des problèmes de communication semblent être une raison importante pour ne pas porter les protecteurs auditifs continuellement. Pour remédier à ce problème, de nouvelles technologies sont appliquées afin de permettre une adaptation automatique de l’atténuation du bruit de fond contraière des protecteurs classiques avec une attenuation constante sur un large plage de niveaux d’entrées. Dans une deuxième partie de ce travail de thèse de doctorat, la reconnaissance de la parole avec des bouchons d’oreille classiques est comparée à des protecteurs auditifs ‘améliorés’ et ce pour des personnes dont l’ouïe est normale. D’une part il ressort des analyses que certaines innovations peuvent bel et bien être bénéfiques selon le type de bruit de fond. D’autre part la re-
connaissance de la parole ne peut être améliorée qu’au cas où le signal d’entrée n’est pas déformé et que les avantages de l’écoute binaurale sont garantis.
Chapter 1

Introduction

The main theme of this research project is prevention of noise-induced hearing loss with hearing protectors. In this regard, the adverse effects of excessive noise exposure on health have been well-established. When sound levels exceed 74 dB(A), both temporary and permanent hearing threshold shifts might occur (Mills and Going 1982). Secondary, loss of concentration, irritation, fatigue, headache and sleep disturbances are also reported (Nandi and Dhatrak 2008; Bernon and Rogez 2007).

In this chapter, the pathophysiology of noise-induced hearing loss will be discussed first, followed by control mechanisms for occupational noise exposure and the role of hearing protectors in this matter. Finally, the research questions of the current study are introduced.

1.1 Pathophysiology of noise-induced hearing loss

Noise-induced hearing loss predominantly results from trauma to the sensory epithelium of the cochlea (Figure 1.1), consisting of one inner and three outer rows of stereocilia hair cells within the organ of Corti (Figure 1.2). In temporary threshold shifts (TTS), several potentially reversible effects include regional decrease in stiffness of the stereocilia, intracellular changes within the hair cells, edema of the auditory nerve endings and degeneration of synapses within the cochlear nucleus. In permanent threshold shifts (PTS), changes become irreversible and include breaks in the rootlet structures of the stereocilia, disruption of the cochlear duct and organ of Corti, loss of hair cells and degeneration of cochlear nerve fibers (Agrawal et al. 2007; Nordmann et al. 2000).

The extend of damage produced in a particular ear is determined by the specific combination of sound pressure level, exposure duration, spectral and temporal pattern of the noise, individual susceptibility of the auditory system (Ward 1997; Henderson et al. 1993) and the combined exposure to certain ototoxicological agents (Alberti 2003). In more extreme cases, high-intensity impulse noise might not only damage the inner ear but also the tympanic membrane.
Figure 1.1: Semi-schematic overview of the outer (Auricula and Meatus acusticus externus), middle (Membrana tympanica and Cavitas tympani) and inner ear (Labyrinthus cochlearis and Labyrinthus vestibularis). The organ of Corti is situated in the Labyrinthus cochlearis or cochlea (Putz and Pabst 2000).

Figure 1.2: Section through the spiral organ of Corti (Gray 1918).
and ossicles.
Leaving aside these injuries to the middle ear, noise-induced hearing loss is most often characterized as bilateral, symmetrical and sensorineural, first affecting the higher frequencies (between 3000 Hz and 6000 Hz) and later expanding to lower frequency regions between 500 Hz and 2000 Hz (Nandi and Dhatrak, 2008). Despite on-going research (Pallarito, 2008; Shim et al., 2009) this loss is to-date irreparable by medical or pharmacological intervention (´Sliwinska-Kowalska and Kotylo, 2007; Nandi and Dhatrak, 2008).

1.2 Controlling occupational noise exposure

Apart from recreational noise exposure, a substantial amount of industrial workers is employed in noisy environments (Dobie, 2008). Because working in such surroundings is unpleasant and even potentially harmful (Tak and Calvert, 2008), decreasing the global sound pressure level seems a logical step. For this purpose, a variety of measures are possible such as adapted machinery design and choice of operational method, vibration isolation, use of damping and sound absorbing materials, machinery enclosure, implementation of dampers (Bies and Hansen, 2003) and active noise and vibration control (Crocker, 1997).

Unfortunately, economical and practical constraints might preclude these measures from reducing the sound level below save limits (Arezes and Miguel, 2002). In this regard, the European Noise Directive on exposure limit values stipulates that personal hearing protectors are compulsory when the eight-hour time-weighted average or the peak sound pressure matches or exceeds respectively 85 dB(A) and 137 dB(C) (2003-10-EC). Although in theory personal protectors are most certainly suitable to reduce the noise at the wearer’s eardrum, occupational hearing loss still remains (Mrena et al., 2008). Naturally, the question arises why.

1.3 Implementation of hearing protectors

Personal hearing protectors basically leave the harmful sound levels unaltered. This implies that sufficient noise reduction not only relies on appropriate attenuation of the protectors, but also on correct and consistent use (Winters et al., 2005). The first condition for effective hearing conservation is a careful selection of hearing protection, meaning that the type of background noise, working conditions and use of other personal protection certainly have to be taken into account in conjunction with the anatomy of the individual outer ear and personal preferences (Nandi and Dhatrak, 2008). After the right protector is selected, thorough training, education and motivation of the individual users are equally indispensable (Witt, 2008; Nandi and Dhatrak, 2008).

All the physical and practical constraints result in very specific demands for hearing protection depending on the working environment. Hence, a wide variety of protectors has been developed which can be classified according to
their appearance or their operating mechanism. Concerning the shape, one can roughly distinguish between earmuffs (Figure 1.3a), helmets (Figure 1.3b) and earplugs (Figure 1.3c to 1.3f) (Berger and Casali, 1997a). As for the earplugs, further distinction can be made between premolded, semi-insert, formable and custom-made earplugs. Premolded earplugs (Figure 1.3c) and semi-insert earplugs (Figure 1.3f) have a well-defined, unchangeable shape. The first, formed from flexible materials, are usually available in a range of sizes to fit most ears and are pushed into place in the ear canal, whereas the latter consist of soft pods that are held in place against or slightly inside the rim of the ear canal by a lightweight band. By contrast, formable earplugs (Figure 1.3d) are formed prior to insertion into the ear canal or pressed into the canal and forced to deform to seal the canal at its entrance. To allow a satisfying fit, the formable earplugs are to be made of more or less kneadable materials such as slow-recovery foam, fiber glass and silicone putty. Finally, custom-made earplugs (Figure 1.3e) are most tailored to seal a specific ear because they are manufactured from an individual impression of the ear canal (Berger and Casali, 1997a). Therefore, there is no need to adapt the earplug’s shape when inserting and hence more solid materials like acrylic and certain silicones can be used.

As for the operating mechanism, one can make a division between ‘standard’ or classical protectors on the one hand and ‘augmented’ on the other (Robinson, 2003). The standard ones solely block the sound path to the eardrum whereas augmented protectors actually process the incoming sound with or without electronics, i.e. in an active or passive way, to improve sound reduction (Buck and Zimpfer-Jost, 2005), to make the attenuation more comfortable (Dantscher, 2007) and to diminish the masking effect of noise on signals (Casali et al., 2004; Casali and Berger, 1996).

Making a careful selection between all these styles and types should lead to a protector that is easy to use, comfortable to wear and appropriately attenuating. The latter requirement means on the one hand that the sound level at the eardrum is sufficiently low and on the other that communication and detection of warning signals is not hampered (Giguère et al., 2008). In this regard, it is important to notice that the concept ‘attenuation’ is not unambiguous (Berger, 2005).

1.4 Attenuation of hearing protectors

The European Noise Directive (2003-10-EC) on exposure limit values stipulates that a worker’s effective exposure must take into account the attenuation provided by his hearing protectors. This can for instance be accomplished by measuring the sound pressure level behind the hearing protector. On the other hand, one can describe the attenuation of the protector by means of its ‘noise reduction’ or ‘insertion loss’.

The insertion loss is defined as the difference between the sound pressure level before and after noise treatment, i.e. with and without a hearing protector in place. The noise reduction is the difference between the sound pressure outside
1.4. Attenuation of hearing protectors

Figure 1.3: Examples of the major types and styles of hearing protectors.

(a) Earmuff.
(b) Helmet.
(c) Premolded earplugs.
(d) Formable earplugs.
(e) Custom-made earplugs.
(f) Semi-insert earplugs.
the ear canal and inside behind the hearing protector (Berger, 2005).

The easiest and fastest way to assess the performance of a hearing protector is using Acoustic Test Fixtures (ATF) such as artificial ears and head-and-torso-simulators (HATS’s) (Berger, 1986, 2005; Birch et al., 2003; Burkhard, 1978; Chasin and Behar, 2003; Parmentier et al., 2000). This approach is very useful in experimental settings (Schroeter, 1985; Schroeter and Poesselt, 1986) or in situations where the use of human subjects would be ethically unacceptable (for instance in impulse noise) (Buck and Dancer, 2007a, b). In addition, these devices can be used for both standard and augmented protectors (Casali and Robinson, 2003). However, even the most up-to-date simulators (Buck and Dancer, 2007b; Berger et al., 2003) are never an exact replica of the human body’s acoustical features and therefore measurement results should always be generalized with caution (Berger, 1986, 2005; Chasin and Behar, 2003).

The current standard in laboratory testing with human ears is the Real Ear At Threshold (REAT) technique laid down in ISO 4869-2 (a), ISO 4869-2 (b) and ANSI S12.6-1997. Basically, subjective audiometry measures hearing threshold shifts with and without hearing protectors to assess the insertion loss. Standardization includes two fitting protocols; an experimenter-supervised fit (method A) and a naïve subject fit (method B) (Berger et al., 1998; Murphy et al., 2004, 2009; Royster et al., 1996). Method A allows the use of subjects who have received training or are experienced in fitting hearing protectors. By contrast Method B requires subjects who have not received any training and have limited experience with protector testing and wearing hearing protectors (Murphy et al., 2009). These techniques are well-documented (Berger, 1986, 2005; Berger et al., 1998; Chasin and Behar, 2003; Murphy et al., 2004), but to-date not useful for impulse noise (Hiselius, 2005) or augmented protectors (Casali and Robinson, 2003) because the REAT technique requires stimuli at detection threshold levels.

These limitations can be overcome by applying the objective Microphone In Real Ear (MIRE) technique (ISO 11904-1) using one or two microphones (Berger, 1986, 2005; Chasin and Behar, 2003). In the single microphone technique, the insertion loss is measured by placing the receiver in the ear canal during separate, consecutive measurements with and without a hearing protector. Using the two-microphone technique, one microphone is placed inside the ear canal underneath the hearing protector whereas the other simultaneously measures the sound level outside the ear to calculate the noise reduction. This objective approach only accounts for air-conduction and thus neglects the sound that reaches the inner ear via bone-conduction. However, bone-conduction seems only critical when dual hearing protectors are worn (Berger and Casali, 1997b).

1.5 Modelling of hearing protectors

Measuring the attenuation of hearing protectors in laboratories will provide crucial information, but might be inadequate to get a full picture of their performance. This is the case when experimental procedures are time-consuming,
1.6. Performance of hearing protectors in situ

Valid laboratory tests and theoretical approaches are obviously indispensable in development, verification and comparison of hearing protectors. However, directly relying on these results for the prediction of an individual’s noise exposure seems less advisable since laboratory tests have shown to approximate the upper limits of achievable protection (Berger and Kerivan, 1983; Berger et al., 1998; Franks, 2003; Hiselius, 2005; Royster et al., 1996; Witt, 2007; Neitzel et al., 2006), even with the REAT method B approach (Berger et al., 1998). One could consider to simply derate the laboratory insertion loss with a fixed dB-value, for instance based on the standard deviation of the laboratory results, but the relationship with an individual’s result remains questionable (Witt, 2008; Franks et al., 2000) given the intersubject variation at the workplace (Franks, 2003; Lenzuni, 2007; Neitzel et al., 2006). Additionally, Casali et al. (1995) have shown that the required derating amount appears highly variable among different types of hearing protectors. In this regard, performing in situ measurements seems a better alternative (Casali et al., 1995; Witt, 2007), all the more because these tests might be included in the training process teaching workers.
1. Introduction

The correct use of their protectors (Joseph et al., 2007; Tsukada and Sakakibara, 2008). The usability of any in situ measurement is not only determined by its accuracy, but also by its rapidity and the required acoustical conditions (Casali et al., 1995). The strict requirements laid down in the ISO and ANSI REAT tests are unrealizable in practice and hence different alternatives have been suggested (Franks et al., 2003). On the other hand the MIRE approach offers a quick and objective way to evaluate the attenuation (Berger, 2005), yielding to an acceptable trade-off between speed, accuracy, repeatability and correspondence with actual practice (Berger et al., 2007). The two different microphone implementations described earlier are implemented successfully with earmuffs (Kotarbinski et al., 2007), but the application with earplugs often requires additional adaptations (Hager, 2006; Pääkkönen et al., 2006; Toivonen et al., 2002). Therefore Voix (2006) and others (Berger, 2007; Néisse et al., 2007) describe a custom-made earplug with an inner bore that allows insertion of a miniature microphone registering sound pressure levels inside the ear canal behind the hearing protector. In practice, this MIRE measurement microphone is mounted in a probe that also contains a reference microphone measuring the sound pressure outside the ear canal (see Figure 1.4).

Since this test design becomes more widespread (Berger, 2007; Burks and Michael, 2003; Neitzel et al., 2006; Voix and Laville, 2009a) a more thorough investigation of the underlying acoustical mechanisms is required, especially with regard to the spatial variation of sound pressure levels in the subject’s outer ear canal and the earplug’s inner bore. Of particular interest is the relation in function of the frequency, henceforth called transfer function, between the sound level measured by the MIRE measurement microphone at the inner bore and the sound level at the eardrum, since the latter is predicted from the former. Between these two points, an apparent difference is expected at certain frequencies, since several authors report substantial variations in the sound pressure amplitude even in an unoccluded ear canal (Gan et al., 2006; Hammershøi and Möller, 1996; Hellström and Axelsson, 1993).

1.7 Research questions

From the introduction it becomes clear that noise-induced hearing loss is an important risk factor in industry. The disorder can be prevented with hearing protectors, but only if those are used both correctly and consistently. In this regard, custom-made earplugs deserve extra attention because they tend to be more positively rated with respect to usability and comfort (Hsu et al., 2004). Given the substantial variability between protector’s performance in laboratory conditions and in situ, even these earplugs merit individual field attenuation measurements (Berger et al., 1996). Therefore, the acoustical mechanisms underlying the MIRE test design depicted in Figure 1.4 are investigated more thoroughly as summarized in Sections 1.7.1, 1.7.2 and 1.7.3. In addition, Section 1.7.4 describes the research concerning speech intelligibility with standard...
1.7. Research questions

Figure 1.4: Earplug with two inner bores; one to adjust the attenuation (b) and the other test bore (a) for insertion of the MIRE probe (c) with measurement (d) and reference (e) microphone. The measurement microphone measures the sound level in the ear canal behind the hearing protector whereas the reference microphone registers the incoming sound level.

and augmented earplugs.

1.7.1 Verifying the attenuation of earplugs in situ: method validation using an artificial head and numerical simulations

In Chapter 2 based on Bockstael et al. (2008) the nature of the transfer functions is first explored by measurements with a head-and-torso-simulator (HATS). Not only the frequency spectrum, but also the reproducibility and generality are of interest. In a second stage, it is verified whether the measured transfer functions can be approximated with numerical finite-difference time-domain (FDTD) simulations of an ear canal occluded by an earplug. A numerical approach is preferred over analytical techniques or circuit-models to take the complex three-dimensional geometry of earplug into account and to enable more easily individualized models in a later stage. Furthermore, a relatively wide frequency range is of interest – i.e. between 100 Hz and 10000 Hz – and thus simulations are carried out in the time-domain. Moreover, the FDTD approach is straightforward, there is long experience in the laboratory with this technique and it is easy to implement.

In the model, the geometrical characteristics of the ear canal-earplug complex are derived from detailed measurements. Further, the impedance of the earplug, the ear canal and the eardrum are used as boundary conditions to include the ear’s and earplug’s acoustical properties. The mechanisms governing the observed sound pressure fluctuations will then be studied starting from the com-
comparison between measurement and numerical results.

1.7.2 Verifying the attenuation of earplugs in situ: method validation on human subjects including individualized numerical simulations

Working with a HATS is advantageous in terms of a verifiable and stable test setup, but at the same time the presence of certain unrealistic phenomenas can not be ruled out due to the artificial character of the HATS’s components. Therefore, Chapter 3 based on Bockstael et al. (2009) addresses the transfer functions registered on 19 human subjects and compares them with the results from the HATS. In the subjects’ ears, the sound pressure is measured simultaneously at the earplugs’ test bore with the MIRE microphone and at the eardrum by sliding a thin silicone tube in the ear canal. Furthermore, numerical models are established to investigate whether the inclusion of individualized parameters can account for the apparent intersubject variability in measured transfer functions. In the concrete, the general properties of the numerical model for the HATS are combined with the specific geometrical dimensions of each ear canal and earplug. These individualized models yield to accurate predictions of the transfer functions, but they require substantial measurement and calculation time.

1.7.3 Filters for accurately verifying the performance of hearing protectors in situ

Once it is possible to predict an individual’s transfer function with reasonably certitude, the practical implementation becomes of interest. The main issue of Chapter 4 based on Bockstael et al. (Accepted) is establishing a workable model which includes (as much as possible) the knowledge obtained in the previous chapters. In this matter, a balance is sought between accuracy on the one hand and feasibility on the other. First, the transfer functions are approached with a simplified one-dimensional analytical model, but the obtained functions correspond too little to the earlier results. By contrast, the research conducted with the FDTD model has lead to very satisfying results and hence it seems more sensible to work further with this technique instead of extending the analytical model with a variety of available extensions and without direct perspectives on a good outcome.

Therefore, a filter approach is chosen instead based on the FDTD simulations. The filter characteristics are linked to the specific geometrical features of ear canal and earplug by multiple linear regression and multivariate orthonormal vector fitting (MOVF). Both fitting procedures predict an individual’s transfer function with satisfying accuracy, but the MOVF might be more suitable in more extreme cases.
1.7.4  Speech recognition in noise with active and passive hearing protectors: a comparative study

Apart from those efforts to verify the performance of hearing protectors in situ, one should never lose track of the fact that adequate attenuation does not imply consistent use. Even when the sound pressure level in the ear canal can be accurately determined, people will be reluctant to wear protectors if they feel that communication and detection of warning signals are compromised (Hong et al. 2008; Okpala 2007; Lwow et al. 2007; Nakashima et al. 2007). To fill this need, augmented protectors have been developed to increase environmental awareness. Chapter 5 based on Bockstael et al. (Submitted) assesses their claimed benefits by comparing speech intelligibility in different types of background noise with two types of augmented protectors and with standard custom-made earplugs. Speech intelligibility scores are obtained for 60 normal-hearing subjects and analyzed in function of the different protectors and types of background noise. Furthermore, research is carried out to link the speech intelligibility scores to the acoustical properties of signal and noise.
Chapter 2

Verifying the attenuation of earplugs in situ: method validation using an artificial head and numerical simulations

This chapter is mainly based on the article published in the Journal of the Acoustical Society of America (Bockstael et al., 2008). Further, results for measurements with silicone earplugs are added as well as data on the influence of different transducers.

2.1 Introduction

As stated in the general introduction, correct implementation of the MIRE approach benefits from insight in the transfer function between the sound pressure at the MIRE measurement microphone and the level of interest at the eardrum. This issue can be addressed by tests with a HATS consisting of a torso and artificial head equipped with pinna, ear canal and ear simulator. The latter mimics the impedance of the middle and inner ear so that the HATS’s microphone can truly represent the eardrum (Parmentier et al., 2000). In addition, the pinna and ear canal model a human’s outer ear and therefore allow the insertion of an earplug in conditions close to reality. Hence the MIRE measurement can be carried out as is meant in practice and simultaneously the sound pressure at the eardrum can be monitored. In general, the HATS indeed appears to give reproducible results (Schroeter, 1985), close to those obtained with methods working with human subjects (Parmentier et al., 2000). Nevertheless, one should always
bear in mind the impossibility to simulate all features of the human head and auditory system \cite{Berger2005}. The HATS measurements that are most important for this research project are made with a free-field loudspeaker, allowing a very stable and verifiable setup. In addition, the transfer functions are also assessed under headphone to address the influence of different transducers. These investigations are carried out to make sure that fluctuations in sound pressure amplitude between the MIRE measurement microphone and the eardrum are fundamentally independent of the sound source, which is very important if practical implementation is considered. In the remainder of this project, the measurement results always refer to the free-field setup unless it is explicitly stated otherwise. Not only the transducer, but also the earplug’s design might influence the transfer function. Therefore different materials and styles are included. Apart from the measurements, simulations may be performed to increase understanding of the results with the HATS and to allow optimizing the MIRE method in a later phase. For the numerical models the FDTD technique is chosen since this time domain method is both efficient and accurate and relies on a simple and straightforward concept. In the concrete it allows to calculate the transfer functions’spectra over the whole auditory spectrum at once \cite{VanRenterghem2004}. These advantages lead to a wide range of applications, including the modeling of sound propagation in and around the human outer ear \cite{TianQing2003,NakazawaNishikata2005}. Its major drawback, i.e. the large computational cost, will be handled in Chapter 4.

2.2 Material and methods

2.2.1 MIRE measurement: general setup

2.2.1.1 Hearing protectors

The tested hearing protectors are manufactured especially for the HATS in acrylic and silicone (40 shore). Each protector has two inner bores, one allowing the insertion of the MIRE probe (the test bore), the other containing a filter or an adjustable valve determining the attenuation (see Figure 1.4). The attenuation of the earplugs with valves may be varied, ranging from an open valve to a completely closed one. Hence, measurements are performed for different conditions; open valve, closed valve and a valve offering an estimated attenuation of respectively 20 dB, 25 dB and 30 dB as preset by the manufacturer. Conversely, the attenuation offered by the filters is fixed at respectively 20 Lohm, 35 Lohm and 65 Lohm. The unit ‘Lohm’ is used by The LEE Compagny to reflect flow resistance of gasses and is calculated by the following equation

\[
\text{Lohms} = \left[ \frac{K \cdot f_T \cdot P}{Q} \right]
\]

(2.1)

with \(Q\) representing the gas flow (in standard liters per minute), \(K\) the gas units constant (to prevent the need to convert pressure and flow parameters
2.2. Material and methods

into specific units), \( f_T \) a temperature correction factor and \( P \) the upstream absolute pressure (in Psia or pounds-force per square inch absolute). Combination of the earplug's material and attenuation style yields for this study to the following subgroups; five acrylic protectors with adjustable valve and two sets of three earplugs with different filters, respectively one set in acrylic and one in silicone.

2.2.1.2 MIRE probe

As stated previously, the MIRE measurements are performed with a probe containing two Knowles low noise FG-3652 microphones; the reference microphone measuring the incoming sound level and the measurement microphone registering the sound pressure in the ear canal behind the hearing protector (Figure 1.4). Using the MIRE probe to assess a worker's sound exposure in situ is only valid if the probe seals the test bore perfectly. Otherwise, the probe might introduce extra leakage that is not present in real wearing conditions and hence the noise exposure will be overestimated. To address the requirement, the attenuation of earplugs is assessed on a Brüel & Kjær HATS type 4128 C by measuring twice the sound pressure at the HATS's eardrum, once with a completely closed test bore and once with the test bore ended by the MIRE probe. The similar attenuation values thus obtained confirm that the condition of perfect sealing is fulfilled.

2.2.2 Free-field measurements with the HATS

2.2.2.1 Measurement system

Measurements are performed with a laptop PC connected to a four input channel data acquisition front-end of Brüel & Kjær (type 3560-C) linking all sound equipment. Recording equipment consists of two prepolarized free-field 1/2” microphones type 4189 (Brüel & Kjær) with preamplifier (type 2669C, Brüel & Kjær), two Knowles low noise FG-3652 microphones (the MIRE microphones) connected to a 9 V preamplifier and the head-and-torso-simulator (HATS) type 4128 C of Brüel & Kjær with a dual microphone preamplifier (Brüel & Kjær, type 5935). The test stimulus is low pass filtered pink noise with a cut-off frequency of 12.8 kHz generated on the PC using Brüel & Kjær’s Pulse Labshop version 7.0. The signal is then transmitted via the front-end and a Pioneer A-607 R direct energy MOS amplifier through a Renkus-Heinz (model CM 81) loudspeaker. The quality of the sound generation system is not critical since the sound signal will be calibrated out in all measurements.

2.2.2.2 Setup

Testing takes place in an anechoic room to prevent disturbances from sound reflection and background noise, therefore the PC is placed outside and the room is only entered between two successive stimuli. The aim of the measurements is the determination of transfer functions between the microphone at the HATS's
ear simulator and the MIRE measurement microphone. However, these results should not be influenced by the typical characteristics of the microphones, nor by the features of the test environment and the test signal. Therefore, the responses of the microphones are compared with the simultaneously registered responses of one free-field microphone. The HATS and this free-field microphone are symmetrically placed in front of the loudspeaker at 1.61 m, see Figure 2.1. The right test ear of the HATS is oriented toward the loudspeaker. Mounting the free-field microphone at approximately the same place as the HATS could seem more convenient but appears impossible since reflections at the HATS’s body disturb the reference signal. An extra measurement is carried out with the second free-field microphone replacing the HATS to calculate the transfer function between the two measurement points (see also Section 2.2.2.4).

2.2.2.3 Measurement sequences and processing

All free-field microphones and the HATS’s microphone are calibrated before each measurement session using the pistonphone 4228 from Brüel & Kjær. Thereafter, the following steps are carried out for each hearing protector. The earplug of interest is placed in the HATS’s ear canal and the MIRE probe is inserted in the test bore at a fixed depth so that the probe does not touch the pinna. Subsequently, the position of the HATS is checked, the investigator leaves the room and the door is carefully shut. Each measurement is completely repeated in order to verify the reproducibility and to detect possible errors.

The signals from the microphones are registered by the Pulse Labshop software mentioned earlier. Linear averaging is carried out over 3000 samples and overloads are rejected. In the frequency range between 0 Hz and 10 kHz the responses are spectrally analyzed using FFT (6400 points).

2.2.2.4 Calibration and post-processing

As stated previously, the transfer functions calculated from these measurements should be absolutely independent of the test signal, the test space and the microphones’ characteristics. To fulfill these conditions, several calibration steps are carried out.

The first step takes place while the measurements are performed. The test setup allows to calculate the frequency response function between on the one hand the MIRE measurement microphone and the HATS’s microphone and on the other hand the free-field microphone; respectively $H_m^{(1)}$ and $H_h^{(1)}$ with $m$ representing the MIRE measurement microphone, $h$ the HATS and $f$ the first free-field microphone. The equation for the frequency response function can in general be written as

$$H_{xy} = \sqrt{\frac{G_{yy}(k)}{G_{xx}(k)} \cdot \frac{G_{yy}(k)}{G_{xy}(k)}}$$

(2.2)

where $H_{xy}$ is the frequency response, $G_{xx}(k)$ and $G_{yy}(k)$ are the autospectra,
2.2. Material and methods

Figure 2.1: Test setup in the anechoic room with earplug and MIRE-probe (a), HATS (b), loudspeaker (c) and free-field microphone (d). To calculate the transfer function $H_{Ff}$ between the two measurement points, the HATS is replaced by the second free-field microphone.

$G_{xy}(k)$ is the cross-spectrum and $G_{xy}^*(k)$ is its complex conjugate (Pinnington, 1998).

To compensate for a possible inhomogeneous sound distribution across the anechoic room, a second calibration step is carried out by calculating the frequency response function (Equation 2.2) between the second free-field microphone replacing the HATS and the first one, the latter placed at its original position. With this transfer function $H_{Ff}^{(2)}$, the measured transfer functions $H_{xf}^{(1)}$ can be calibrated using the following expression

$$H_{xf}^{(2)} = \frac{H_{xf}^{(1)}}{H_{Ff}}$$

(2.3)

with $x$ being respectively $m$ or $h$ and $F$ standing for the second reference microphone.

Further, the influence of the different microphone characteristics has to be accounted for. For the MIRE measurement microphone, the frequency response function is measured by closely mounting this microphone to the second free-field microphone placed straight in front of the loudspeaker. This procedure yields to a calibration function over the frequency range of interest namely $H_{mF}^c$. Calibrations over the different test days reveal that $H_{mF}^c$ is very stable.
With this function the measurements can be corrected yielding to

\[ H_{mF}^{(3)} = \frac{H_{mF}^{(2)}}{H_{mF}^{(2)}}. \]  

(2.4)

Finally, the transfer function between the MIRE measurement microphone and the HATS \((H_{mh})\) can be calculated, namely

\[ H_{mh} = \frac{H_{mF}^{(3)}}{H_{hF}^{(2)}}. \]  

(2.5)

### 2.2.3 Measurements with different transducers

Completely similar to the measurement procedure described above, tests are also carried out under headphone for the acrylic earplugs. In this, two circumaural headphones are used, one with closed cups (Philips SDC HP-890) and one with open (Sennheiser HD 280 pro). If a headphone is acoustically closed, it is intended to prevent any acoustic coupling between the external environment and the ear canal whereas open cups intentionally provide an acoustic path between the ear and the surroundings.

Measurements to compare the influence of the transducers are made on different occasions than the solely free-field measurements, but the general setup is equaled as much as possible, including among other things the variety in protector’s attenuation styles (see Section 2.2.1.1). Naturally, calibration with the free-field microphones (see Section 2.2.2.4) is impossible when working with headphones, so the responses from the MIRE measurement microphone and the HATS are directly compared, increasing the risk of artifacts. However, this does not seem critical since the aim is not the assessment of the transfer functions themselves, but conversely only possible differences between transducers are of interest. The results are covered in Section 2.3.2.

### 2.2.4 Numerical simulations

The numerical simulations are in the first place developed for the acrylic hearing protector, the rationals behind this approach are elaborated in part 2.3.1.2.

#### 2.2.4.1 Key factors of the simulations

The key factors and choices made for the numerical FDTD simulations are briefly discussed. In the absence of background flow, the equation for particle velocity \(\mathbf{u}\) and acoustic pressure \(p\) can be written as

\[ \frac{\partial \mathbf{u}}{\partial t} + \frac{1}{\rho_0} \nabla p = 0 \]  

(2.6)

\[ \frac{\partial p}{\partial t} + c^2 \rho_0 \nabla \cdot \mathbf{u} = 0 \]  

(2.7)
2.2. Material and methods

where \( t \) denotes time, \( c \) the speed of sound (in these simulations set at 340 m/s) and \( \rho_0 \) the density of air (1.21 kg/m^3) (Pierce, 1997).

For the FDTD simulations, both \( p \) and \( u \) are discretised in space in Cartesian grids (resolution: 0.00035 m) that are staggered by shifting the grid for discretising \( u_\alpha \) over half of a grid step, \( \frac{d\alpha}{2} \), in direction \( \alpha \) with respect to the grid chosen for discretising \( p \). The spatial organization for the staggered grid is illustrated by Figure 2.2. In time, staggering is obtained by calculating \( p \) at \( t = l dt \) and \( u \) at \( t = (l + \frac{1}{2}) dt \) (time step: \( 5.9 \cdot 10^{-7}s \)). The resulting equations

\[
\begin{align*}
    u^{l+\frac{1}{2}}_\alpha (\alpha + \frac{1}{2}) &= u^{l-\frac{1}{2}}_\alpha (\alpha + \frac{1}{2}) - \frac{dt}{\rho_0 d\alpha} \{ p^l (\alpha + 1) - p^l \} \quad (2.8) \\
    p^{l+1} &= p^l - \sum_{\beta=x,y,z} \frac{\rho_0 c^2 dt}{d\beta} \{ u^{l+\frac{1}{2}}_\beta (\beta + \frac{1}{2}) - u^{l+\frac{1}{2}}_\beta (\beta - \frac{1}{2}) \} \quad (2.9)
\end{align*}
\]

allow to step in time replacing old values by newly calculated ones without much memory overhead (i.e. in-place computation). The brief notation \( (\alpha + q) \) is used to indicate that the value is taken at a point shifted by \( q \) spatial steps \( d\alpha \) in the \( \alpha \)-direction with respect to the reference location referred to by indices \( (i,j,k) \).

The first equation is repeated for \( \alpha = x,y,z \) where \( x \) and \( y \) are defined along the cross-section of the ear canal and \( z \) represents the longitudinal axis.

The staggered-grid approach is advantageous because it results in higher accuracy. Moreover, the scheme is very compact, making the implementation of boundary conditions easier. In a staggered spatial grid, the acoustic pressures are typically situated in the center of each computation cell whereas the components of the particle velocity are on the faces. This implies among other things that a rigid boundary condition is obtained by simply setting the orthogonal particle velocity at that point equal to zero (Van Renterghem and Botteldooren, 2007).

In general, boundary impedance of the form

\[
Z = j\omega Z_1 + Z_0 + \frac{Z_{-1}}{j\omega} \quad (2.10)
\]

can easily be implemented in the FDTD method (Botteldooren, 1995). Such boundary conditions will be used to model the ear’s and earplugs’s acoustical properties.

Finally, the Courant number is set to 1 to ensure stability and to minimize the phase error (Botteldooren, 1995). All this allows to perform accurate simulations up to 10000 Hz, given that the spatial resolution is sufficiently high for this wave length (Botteldooren, 1995). Solely by way of illustration, Figure 2.4 depicts a snapshot of the sound pressure distribution in the FDTD model of the ear canal occluded by an earplug with two inner bores.

2.2.4.2 Modelling the impedance of the outer, middle and inner ear

Impedance of the outer ear For the propagation of sound in the outer ear canal, the impedance of bone (6.12 kg/m^2s) (Wit et al., 1987) is included as...
boundary condition. This approximates the real-life situation where the hearing protector fills the cartilaginous part of the outer ear canal and hence relevant sound propagation effects occur mainly in the ossicular part.

**Impedance of middle and inner ear** The acoustics of the middle and inner ear are represented by the impedance at the eardrum, $Z_d = \frac{P}{u_n}$, where $u_n$ is the orthogonal component of $u$, since explicitly modelling the middle ear is outside the scope of this work. The impedance at the eardrum is included because different authors have shown that, especially for intra-aural devices, the eardrum impedance needs to be simulated (Schroeter and Poesselt 1986; Hammershøi and Møller 1996).

Various eardrum impedance models are suggested in literature. Here the terminating impedance is based on one-dimensional circuit models of the middle and inner ear since the cross sectional variation is of little importance for the problem at hand. These models allow the approximation of the effective impedance by compounding networks of acoustical and mechanical compliances, masses, frictional resistances and transformers (Gan et al. 2006) representing the relevant physical parts of the auditory system (Kringlebotn 1988).

Different circuit models by Hudde and Engel (1998b), Kringlebotn (1988), Pascal et al. (1998) and Shaw and Stinson (1981) are investigated. For each approach, the overall eardrum impedance is calculated and compared with the eardrum impedances for human subjects as reported by different authors (Farmer-Fedor and Rabbitt 2002; Keefe et al. 1993; Margolis et al. 1999; Voss and Allen 1994). Furthermore, an unoccluded ear canal is modeled using numerical FDTD simulations with various eardrum impedances according to the different network models. The outcome of these simulations is compared to measurements made by Hammershøi and Møller (1996).

From all these comparisons it can be seen that pressure patterns derived from the model proposed by Kringlebotn (1988) resemble the experimental results

---

**Figure 2.2:** Spatial organization of one cell in a three-dimensional staggered grid.
most closely. The difference in peak frequency between the FDTD model and experimental results (Hammershøi and Møller 1996) may be caused by the different canal lengths in experimental subjects and the model (Gan et al. 2006). Hence, the eardrum impedance resulting from Kringlebotn’s network is included in the FDTD model as was done by Hiselius (2004) in his two-port model of an occluded ear canal.

To include the complex impedance in the FDTD model a ratio of polynomials in $j\omega$ is first fitted. A bilinear transformation is then used to transform this analogue filter to a digital equivalent:

$$p = \sum_{i=0}^{n} d_i z^{-i} \sum_{k=0}^{m} c_k z^{-k} u_n. \quad (2.11)$$

The impedance of the eardrum does not represent locally reacting material; therefore it is implemented on the average field:

$$u_n^{+\frac{1}{2}} = \sum_{k=0}^{m} c_k z^{-k} \langle p' \rangle \sum_{i=0}^{n} d_i z^{-i} \langle p \rangle \quad (2.12)$$

where $\langle \rangle$ denotes spatial averaging over the eardrum. The temporal and spatial mismatch due to the use of staggered grids turn out to have little influence. Hence the coordinate system is chosen such that a coordinate plane $x = \frac{1}{2} dx$, where $u_n$ is discretised, coincides with the eardrum. The pressure half a grid step away at $x = dx$ is assumed to be a good approximation for the pressure on the eardrum. The single digital filter representing the middle and inner ear is implemented using standard filter routines in the software MATLAB from The MathWorks\textsuperscript{TM}.

Due to the complicated frequency dependence of the impedance, $m = n = 5$ is needed. Figure 2.3 shows amplitude and phase of the complex surface impedance of the eardrum and the digital approximation that is used.

### 2.2.4.3 Modeling the hearing protector

**Dimensions of the hearing protector** The relevant dimensions of the hearing protector in total and of the two inner bores are measured with a digital caliper accurate to 0.01 mm. However, the length of the inner bores (in reality 1.74 cm) is slightly lengthened in the simulations with 0.5 cm to enhance similarity with the measurements. This modification will be discussed in Section 2.4.

**Modeling the sound field in the narrow channels in the hearing protector** The channels in the hearing protector are very narrow. Therefore the effect of viscosity and heat conduction becomes potentially important. To avoid having to compute numerically the strong spatial dependence of the field close to the boundaries, a sub grid scale approximation for the boundary is used. This approach is based on an analytical description of the vorticity and entropy layer close to a flat boundary (Pierce 1997). It is previously introduced as a
Figure 2.3: Amplitude and phase of the complex surface impedance of the eardrum with digital approximation.
sub grid scale approximation in FDTD simulation by Botteldooren (1997). The
time domain approximation of the square root is refined in comparison to this
earlier publication to be applicable over a wider frequency range. The derivation
for the viscosity effect starts from the equation for conservation of impulse for
the parallel component $u_\beta$ averaged over a grid cell orthogonal to the surface

\[
\sqrt{j\omega \rho_0 \mathfrak{n}_\beta} = -\frac{\partial p}{\partial \beta} - \frac{1}{d\alpha} \tau_\beta \tag{2.13}
\]

where $d\alpha$ is the grid cell size orthogonal to the surface, $\tau_\beta$ the shear stress, the
over stripe indicates the averaging, and

\[
-\frac{1}{d\alpha} \tau_\beta = \mu \frac{d\alpha}{d\alpha} \int_0^{d\alpha} \frac{\partial^2 u_\beta}{\partial \alpha^2} d\alpha. \tag{2.14}
\]

The latter term accounts for the influence of viscosity ($\mu$ is the dynamic viscos-
ity). Within the boundary layer approximation, the second order derivative is
dominated by the boundary layer field. It vanishes at the edge of the boundary
cell ($\alpha = d\alpha$) as long as the boundary layer thickness is small compared to the
grid cell size: $\delta = \sqrt{\frac{2\nu}{\omega}} < d\alpha$ where $\nu = \frac{\mu \rho_0}{\rho_0}$. Thus we obtain

\[
\tau_\beta = -\frac{\partial u_\beta}{\partial \alpha} \bigg|_{B'}. \tag{2.15}
\]

Introducing the exponential decay of the boundary layer results in

\[
\tau_\beta = \rho_0 (1 + j\sqrt{\frac{\nu \omega}{2}}) \mathfrak{n}_\beta. \tag{2.16}
\]

Equations 2.13 and 2.16 are now transformed back to time domain. For Equation
2.16 this is not trivial. Direct transformation of the product of $\sqrt{\omega}$ with
the field value results in a convolution involving a $\sqrt{t}$. Numerical approxima-
tion of this convolution can result in very long calculation times. Therefore the
$\sqrt{\omega}$-term is first approximated by a ratio of polynomials in $j\omega$. This methodo-
logy is inspired by classical electronic filter design. For digital filter design, a
bilinear transformation is traditionally used to transform continuous to discrete
time since it unconditionally results in a stable digital filter that is also a ratio
of polynomials. The digital filter approximation that is obtained using stan-
dard approximation techniques focusing on a frequency range [0 Hz, 5000 Hz] is
written as

\[
\tau_\beta = \sqrt{\mu \rho_0} \sum_{i=0}^{n} b_i z^{-i} \mathfrak{n}_\beta, \tag{2.17}
\]

where $z^{-1}$ is the equivalent in Z-domain of a single time step delay. This expan-
sion is used to discretise Equation 2.13 in the staggered grid both in space and
in time. As usual this equation is evaluated at $t = l dt$ (where $p$ is known) while
the components of $u$ are discretised at $t = (l + \frac{1}{2}) dt$. Thus an additional $\frac{1}{l + \frac{1}{2}}$
is introduced to resolve this mismatch. This eventually leads to the adapted FDTD update equation:

\[
\left( \frac{\rho_0 u_0}{dt} + \frac{\sqrt{\mu \rho_0}}{2d\alpha} \right) u_{\beta}^{l+\frac{1}{2}} = - \sum_{k=0}^{m} a_k \frac{\partial p}{\partial \beta} |_{l-k} - \frac{\rho_0}{dt} \sum_{k=1}^{m+1} (a_k - a_{k-1}) u_{\beta}^{l-k+\frac{1}{2}} - \sqrt{\mu \rho_0} \sum_{i=1}^{n+1} (b_i + b_{i-1}) u_{\beta}^{l-i+\frac{1}{2}}
\]

(2.18)

where \(a_{m+1} = b_{n+1} = 0\) are introduced to simplify notations. Spatial discretisation is not explicitly denoted.

Note that the approach used to include boundary layer effects is rather memory extensive. It requires storage of \(m\) old values of the spatial derivative of \(p\) and \(\max(m,n) - 1\) additional old values of \(u_\beta\). Fortunately, this additional storage is required at the boundaries only. For the simulations reported in this article, \(m\) and \(n\) are chosen equal to 2. This results in the coefficients \(a_0 = 1, a_1 = -1.871, a_2 = 0.87213, b_0 = 391.02, b_1 = -769.2, b_2 = 378.2\), and a reasonably accurate approximation over the frequency range of interest.

The influence of heat conduction on sound propagation can also be introduced using a sub grid scale approximation based on boundary layer theory (Botteldooren, 1997; Howe, 1998). The equation that describes the evolution of the small acoustic pressure fluctuation is usually derived from the conservation of energy and the conservation of mass. By keeping terms that describe heat flux near the flat surface and by assuming that this flux is largest within a small boundary layer close to that surface, the grid cell averaged equation can be written as

\[
\omega \bar{p} = -\rho_0 c^2 \nabla \cdot \bar{u} - \frac{\gamma - 1}{d\alpha} \bar{q}_s
\]

(2.19)

where \(d\alpha\) is again the grid cell size orthogonal to the surface, \(\gamma\) the ratio of specific heats, \(q_s\) the heat flux, the over stripe indicates the averaging, and

\[
-\frac{1}{d\alpha} \bar{q}_s = \frac{\kappa}{d\alpha} \int_0^{d\alpha} \frac{\partial^2 T}{\partial \alpha^2} d\alpha
\]

(2.20)

where \(\kappa\) is the heat conductivity and \(T\) is the acoustic temperature fluctuation. Within the boundary layer approximation, the second order derivative is again dominated by the boundary layer field. It vanishes at the edge of the boundary cell (\(\alpha = d\alpha\)) as long as the thermal boundary layer thickness is small compared to the grid cell size: \(\sqrt{\frac{2(\gamma-1)\kappa}{\omega_0 \rho_0 R}} < d\alpha\) (\(R\): gas constant). Thus we obtain

\[
\bar{q}_s = -\kappa \frac{\partial T}{\partial \alpha} |_{\alpha = B}. \quad (2.21)
\]
Introducing the exponential decay of the boundary layer results in

$$q_s = (1 + j) \sqrt{\frac{\kappa \rho \gamma R \omega}{2(\gamma - 1)}} T. (2.22)$$

The cell averaged acoustical temperature fluctuation $T$ is dominated by isentropic acoustic propagation. Thus the known relationship that relates acoustic pressure to acoustic temperature $p = T$ can be used to obtain

$$q_s = (1 + j) \sqrt{\frac{\kappa(\gamma - 1) \omega}{2 \rho \gamma R}} p = (1 + j) \sqrt{\frac{\kappa \omega}{2 \rho c_p}} p \quad (2.23)$$

where $c_p$ is the specific heat at constant pressure for the gas (air). Exactly the same way as for the viscous boundary layer effect, equations 2.19 and 2.23 are now transformed back to time domain and discretised following the FDTD scheme as described above.

Note that the approach used to include boundary layer effects is rather memory extensive. It requires storage of $m$ old values of the divergence of $u$ and max($m, n$) − 1 additional old values of $p$ for implementing the thermal conduction effects. Fortunately, this additional storage is required at the boundaries only. For the simulations reported in this article, $m$ and $n$ are chosen equal to 2. This results in the coefficients $a_0 = 1, a_1 = -1.871, a_2 = 0.87213, b_0 = 391.02, b_1 = -769.2, b_2 = 378.2,$ and a reasonably accurate approximation over the frequency range of interest.

**Modeling the impedance of the hearing protector’s material**

The impedance of acrylic is calculated based on measurements of the complex Young modulus $E$ (Hillström et al., 2000, 2003) and Poisson’s ratio $\sigma$ (Hillström et al., 2003). From these data, the complex longitudinal sound speed $v$ may be calculated using the following expression (Jarzynski et al., 2003; Pierce, 1997)

$$v = \sqrt{\frac{E}{3(1-2\sigma)} + \frac{4}{5} \frac{E}{2(1+\sigma)}} \rho \quad (2.24)$$

with the density of acrylic $\rho = 1183 \frac{kg}{m^3}$ (Hillström et al., 2003).

The characteristic impedance of acrylic can easily be calculated based on $\rho$ and the complex wave number $\kappa_\delta = \frac{\pi}{\lambda}$ with $\omega$ the angular frequency. Considering the reflection at an infinitely thick and long layer, this yields to a merely constant and real impedance in the frequency range of interest, $Z \approx 3.1 \cdot 10^6 \frac{kg}{s \cdot m^2}.$

**Impedance of the entity earplug - ear canal**

The surface impedance of the earplug terminating the residual part of ear canal between hearing protector and eardrum does not simply equal its material impedance. By contrast, the combination of earplug and resilient ear canal is believed to influence the resulting surface impedance (Hiselius, 2005; Schroeter, 1985).
This surface impedance is measured by tightly mounting the HATS’s pinna with hearing protector on an impedance tube. The reflection coefficient and normal surface impedance are determined by measuring the two-microphone transfer-function according to [ISO 10534-2]. Because the impedance of the earplug’s material is so high, viscous damping in the measurement tube has to be calibrated out. Within the frequency range of interest, the surface impedance can be approximated by an expression of the form written in Equation 2.10 with constants $Z_0 = 0.50 \cdot 10^4 \frac{kg}{s \cdot m^2}$, $Z_1 = 0.41 \frac{kg}{m^2}$, and $Z_{-1} = 1.69 \cdot 10^7 \frac{kg}{s^2 \cdot m^2}$.

Modeling the impedance of the miniature microphone The acoustic impedance of the microphone’s diagram system is modeled by a very high impedance. This is based on the fact that a small diameter increases the diagram’s impedance substantially [Bruel & Kjaer], yielding to a very high impedance for $\frac{1}{10}$” microphones.

2.3 Results

2.3.1 Transfer function between MIRE and HATS microphone

Examples of the measured transfer functions ($H_{mh}$) between the MIRE microphone and the HATS microphone are depicted in Figure 2.5. The $H_{mh}$ for 25 dB attenuation is omitted from Figure 2.5a to enhance clarity of the graph and because this $H_{mh}$ is almost identical to the $H_{mh}$ for 20 dB attenuation. In general, little difference is seen between the different transfer functions for acrylic earplugs on the one hand and silicone ones on the other.

Figure 2.4: Example of momentaneous sound pressure distribution in an ear canal occluded by an earplug with two inner bores as seen on the cross-section of the FDTD model.
Figure 2.5: Amplitude of the measured transfer functions between the MIRE measurement microphone and the HATS’s microphone. Figure (a) depicts the transfer functions for the acrylic earplugs with a completely closed valve (‘closed’), a completely open valve (‘open’) and a valve offering an attenuation of 20 dB and 30 dB. Figure (b) shows the transfer functions for acrylic earplugs with respectively a 20 Lohm, 35 Lohm and 65 Lohm filter whereas Figure (c) contains the transfer functions for the silicone earplugs with the same filter values.
2.3.1.1 Transfer function for acrylic earplugs

An extra figure is made for the acrylic earplugs to allow the comparison between the measured and the simulated transfer functions (Figure 2.6). For most earplugs and for the numerical simulations, the transfer function between the MIRE microphone and the microphone of the HATS is found to be flat in the lower and middle frequency region (up to 2500 Hz). However, the spectrum shows a distinct peak around 1000 Hz for some earplugs in all sets of hearing protection devices. This peak is never seen in the numerical approach with parameters described in Section 2.2.4. The hypothesis of accidental measurement errors is somewhat inconsistent with the fact that repeated tests show the same aberration. An alternative line of thinking is the assumption that this peak corresponds to an extra mode caused by the artificial character of the HATS. The transfer functions’ amplitude increases above 2500 Hz with a clearly distinguishable peak around 3600 Hz. The FDTD simulations show that merely adaptations in the characteristics of the test bore and its boundaries affect this maximum. Hence, this structure is held responsible for the observed resonance. Measurements with the MIRE microphone attached to the hearing protector in free-field conditions (i.e. outside the HATS’ ear canal) confirm this assumption by showing a resonance peak at the same frequency.

In the frequency range between 4000 Hz and 7000 Hz, the spectral behavior of the transfer function is dominated by the resonance of the residual volume of the ear canal. The simulation with closed test bore shown by the dashed line in Figure 2.6 already indicates that the residual volume is responsible for this broad dip in the transfer function. From data reported by Tomdorf (1988) the length of the residual volume between the hearing protector and the eardrum is estimated at 16 mm, yielding to a resonance frequency consistent with the currently measured and simulated transfer functions. The resonance of the test bore thus compensates for the dip produced by the resonance of the residual volume. Unfortunately the compensation does not result in an overall flat transfer function.

At still higher frequencies, the measurements show an increased complex and peaky behavior. This could be due to different higher order modes involving amongst others the vibration of the eardrum. It can not be taken for granted that the HATS measurements are representative for measurements in real humans in this upper frequency range since the middle ear model of the HATS might fail at these frequencies.

The phase of the transfer functions reveals that the MIRE microphone and the HATS microphone are in phase. Fluctuations in phase are observed at the resonance frequencies described earlier, these findings are consistent with the Kramer-Kronig relations.

2.3.1.2 Transfer function for silicone earplugs

The transfer functions of the silicone earplug clearly differ from their acrylic counterparts (see Figure 2.5). The configuration appears very similar up to
2.3. Results

Figure 2.6: Amplitude of the simulated transfer function (thick red line) between the sound pressure at the MIRE measurement microphone’s position, i.e. at end of the test bore, and at the eardrum. In the background, the measured transfer functions for all acrylic earplugs are plotted in gray. The dashed line represents the simulated transfer function with closed test bore.

2500 Hz but for higher frequencies a plateau is seen instead of multiple extrema. The silicone earplugs are based on the same impression as the acrylic and similar filters are used, hence there are no apparent differences in configuration that might explain the observed dissimilarities. This suggests that either variation in attenuation or in material might be held responsible. However, additional measurements with the acrylic earplug point out that the transfer functions are independent of the applied attenuation (see Section 2.3.3), hence differences in the transfer functions are most likely linked to the material of the hearing protectors.

First, possible influences of the material on the resonances in the test bore are investigated by attaching the MIRE probe to the earplug in free-field conditions outside the HATS’s ear simulator. In these conditions, a well-defined peak is found around 3500 Hz, indicating that the material has no influence on the resonance-frequency of the test bore.

An alternative line of thinking is that different boundary impedances at the residual part of the ear canal might influence the sound pressure distribution. This could be easily verified by varying the earplug’s impedance (see Equation 2.10) in the numerical simulations. Since none of those models yield to transfer functions similar to the silicone measurements, differences in boundary impedance seem a little valid explanation for the observed variations.

These observations yield to the idea that the silicone results might be attributed to problems with the experimental setup, meaning that the MIRE measurement
microphone somehow captures sound that does not reach the eardrum. The extra sound path is possibly caused by a somewhat looser fit between the MIRE probe and the test bore or direct sound propagation through the hearing protector. In this regard it is not surprising that these aberrations are merely seen in the frequency region where the attenuation is the highest, implying that increases in the sound level at one point are more influential.

### 2.3.2 Reproducibility

The reproducibility is verified by comparing the results from repeated free-field measurements and in this regard the responses of the MIRE measurement microphone appear to be stable if the microphone is carefully placed. In the concrete, the earplugs made of acrylic yield to reproducibility within a 1 dB-range in the frequency range from 300 Hz to 10000 Hz and for silicone this is the case between 100 Hz and 2500 Hz. Outside this range, no variations greater than 2 dB are found.

Additionally, the uncalibrated transfer functions obtained with the free-field loudspeaker and under headphones are compared for the acrylic earplugs. Overall, different transducers do not seem to introduce systematic deviations since they all yield to very similar results, see Figure 2.7 as an illustration. Nevertheless, the transfer functions from the headphone with closed cups appear a bit more sensitive to sporadic sharp resonances.

### 2.3.3 Differences between transfer function

Although only little differences are observed between transfer functions measured within the groups of hearing protection devices, it is investigated whether this variance is caused by possible (small) variations in the design of the different earplugs or by their attenuation. This is done by performing multiple measurements using the acrylic hearing protectors with their adjustable valve put in its different positions. The variation between transfer functions obtained from the same earplug with different attenuation levels nor the differences between hearing protectors with the same attenuation do exceed the outer limits of the reproducibility found earlier (2 dB). Thus these results do not provide evidence for substantial influence of the attenuation level or the earplug’s design on the variances between transfer functions. However, the latter finding is somewhat obvious since all protectors are manufactured based on the same impression of the HATS’s ear canal.

### 2.4 Discussion

The main aim of this chapter is to gain insight in the basic principles governing the transfer function between the MIRE measurement microphone and the eardrum. This requires in the first place a very stable and controlled test design so that possible effects of confounding factors are minimized. Therefore
the HATS 4128 C is preferred over human subjects to perform the MIRE measurements. The free-field responses of the HATS in this study’s setup are very similar to those reported by the manufacturer for similar sound incident directions. This indicates that the HATS's measurements are free from influences by the test signal and the test environment.

All measurements are performed with fixed sound incident directions because Hammershøi and Møller (1996) have shown that the transmission to the eardrum from any point between the eardrum and the point 6 mm outside the ear canal can be considered directionally independent. Moreover they have proved that this directional independence is also valid for a blocked entrance. Closely related to this directional independence are the findings that the transfer functions are not influenced by the actual sound source. Naturally, the stable and verifiable setup makes the free-field approach clearly preferable in research conditions, but in practice headphones might be more convenient (Berger 1986). In this, the circum-aural headphone with closed cups seems a little bit more prone to sporadic sharp resonance peaks, therefore open cups might be preferred. In the FDTD simulations, the sound pressure level at the eardrum is based on the sound pressure level at one central point in the ear canal because the sound pressure is constant across the ear canal below 10 kHz and thus only the longitudinal mode is present (Hammershøi and Møller 1996), given that any nonplanar modes in this frequency range are strongly attenuated (Voss and Allen 1994).

The back radiation impedance, i.e. the radiation impedance seen outward from
the ear entrance [Hudde and Engel (1998b)], is not included in this model because Hiselius (2004) has shown that inclusion or exclusion of this parameter has little impact on the results for occluded ears.

The measurement results reveal stable and reproducible transfer functions which are independent of the earplug’s attenuation. Moreover, FDTD simulations of the hearing protector and outer ear canal come close to the results of most acrylic earplugs, except for the maximum around 1000 Hz visible in the spectra of some transfer functions. This maximum does never occur in the numerical approach, even when the nominal values of the different parameters are modified. Hence, the origin of the peak is most likely to be located elsewhere, for instance in the artificial character of the HATS ear simulator. In the simulations the impedance of the middle and inner ear have been taken from literature data on biological ears and might therefore not show the resonance around 1000 Hz as strongly as the HATS middle and inner ear model.

Furthermore, the test bore was slightly lengthened in the simulations to enhance similarity with its measured resonance frequency. This adaptation accounts for loss mechanisms for which exact quantification is outside the scope of this work. For instance, the viscothermal damping effects at the (sharp) edges at the end of the test bore are not accounted for. Also, the test bore’s termination toward the eardrum consists in reality of a very small cavity with rounded, inclined walls. The simulation includes a cavity with comparable dimensions but with straight walls. The more gradual transition between the air in the ear canal and the test bore might in reality enhance the radiation of sound toward the residual part of the ear canal. Hence, the resonance frequency of the test bore might be lowered.

The MIRE measurement microphone might also be (partly) responsible for the necessary lengthening of the test bore. Although the probe is carefully inserted, there might be a difference between the actual and simulated insertion depth. However, it seems somewhat unlikely that the true insertion depth deviates exactly with the same amount in all measurements situations.

Furthermore, the microphone’s acoustic impedance is based on theoretical values of a high-quality electret microphone since no data are available concerning the actual impedance of the microphone under study. This implies that only the impedance of the diaphragm is taken into account and not possible air volumes inside the probe influencing the diaphragm’s acoustical characteristics.

Enlarging the test bore’s length in the numerical model means that an extra air volume is taken into account, comparable to a tube filled with air in front of the MIRE measurement microphone. The input impedance \( Z_i \) of such a system can be written as

\[
Z_i = Z_0 \frac{Z_m + j k L}{1 + j Z_m k L}
\]

with \( Z_0 \) the characteristic impedance of air, \( Z_m \) the impedance of the MIRE measurement microphone, \( k \) the wave number and \( L \) the length of the tube (Kinsler et al., 2000), i.e. 0.5 cm if the necessary lengthening is entirely attributable to the approximation of the microphone impedance. Solving the
equations shows that $Z_m$ and $Z_i$ are dominated by the stiffness-component, but as expected the latter is smaller than the former, respectively order of magnitude $10^{10}$ and $10^4$.

These calculations show that the two impedances can be considered acoustically hard, therefore measurements would not allow to differentiate between the impedance included in the numerical simulations and the impedance including the air volume. Because both values are large compared to the characteristic impedance of air, one could state that they fall within the uncertainty interval of the microphone’s actual impedance. Hence it is indeed possible that air volumes in the MIRE probe influence the terminating impedance at the test bore, thus altering its resonance frequency.

Anyway, the lower measured resonance frequency is thought to be due to characteristics of the test bore-probe complex only. As measurements with the hearing protector in free-field show a comparable resonance frequency, the outer ear nor the middle may influence this maximum (see Section 2.3.1). Moreover, hearing protectors differing in the design of the second inner bore all have a similar maximum (see Figure 2.5). These findings strongly support the hypothesis of the test bore determining the resonance frequency. Hence, it seems justified to include the most likely effects at both test bore’s terminations in the simulations by slightly lengthening the test bore itself.

As for the silicone earplugs, the characteristics of the measured transfer functions are most likely related to the plug’s material. However, adapting these boundary conditions in the current FDTD simulations never yields to similar behavior of the transfer functions in the higher frequency range. These results suggest that the current MIRE setup might have some shortcomings for this material. Nevertheless, one should bear in mind that a certain interaction with the artificial character of the HATS can not be ruled out at this point.

2.5 Conclusion

The currently conducted measurements yield to stable and reproducible transfer functions between the MIRE measurement microphone and the eardrum. Moreover, for the acrylic earplugs the combination of measurements and FDTD simulations makes it obvious that the most striking spectral features may be traced down to morphological aspects of the test bore and the ear canal’s residual cavity. Conversely, for silicone the measurement results do not correspond to the numerical approach. The aberrations are most likely related to the earplug’s material, but it is thus far unclear whether the discrepancy is caused by the test design itself or by the artificial character of the HATS.

The next chapter will focus on the correspondence between the transfer functions for the HATS and human subjects. Measurement results are compared for both acrylic and silicone earplugs and additionally the aptness of the FDTD simulations is assessed for the acrylic plugs.
2. Verification in situ: method validation on HATS
Chapter 3

Verifying the attenuation of earplugs in situ: method validation on human subjects including individualized numerical simulations

The content of this chapter is based on the article published in the Journal of the Acoustical Society of America \cite{Bockstael2009}. Further, results for measurements with silicone earplugs are added.

3.1 Introduction

The previous chapter reveals that measurements on a HATS yield to stable and reproducible transfer functions between the MIRE measurement microphone and the eardrum. Moreover, the results for acrylic earplugs can be quite accurately approached with numerical simulations and their most striking spectral characteristics can be traced down to morphological features of ear canal and earplug. By contrast, the origin of certain aspects like maxima around 1000 Hz and multiple minima in the higher frequency range is less clear. Further, the transfer functions of the silicone earplugs seem to be controlled by other mechanisms than their acrylic counterparts.

Thus far combining the MIRE approach with transfer functions appears a valid option to assess sound exposure when wearing acrylic earplugs. However, it is unwise to generalize the results obtained on a HATS to human subjects with-
out additional research. Despite the technological progress in the design of HATS’s (Buck and Dancer [2007b]), these devices are never an exact replica of the human body’s acoustical features (Berger [1986], 2005). In this way, transfer functions measured with the HATS might display certain aspects that are not purely attributable to fluctuations in sound pressure amplitude but by contrast are caused by the artificial character of the simulator. Apart from possible artifacts, one HATS can never account for differences between human subjects. This might be critical when assessing sound pressure distribution within the ear canal since Hammershøi and Møller (1996) have shown that the intersubject variability is quite high in this. Therefore, the transfer function obtained for the HATS is farther elaborated for humans by simultaneously measuring the sound pressure at the eardrum and at the MIRE measurement microphone for a sample of volunteers; both silicone and acrylic earplugs are included.

Furthermore, the relationship between the morphology of an individual’s acrylic hearing protector and ear canal on the one hand and the transfer function’s appearance on the other will be assessed with FDTD simulations based on the earlier developed numerical approach for the HATS. It will be verified whether FDTD models representing a specific individual’s occluded ear canal provide satisfying correspondence with measured transfer functions. Furthermore, the possible gain in likeness will be assessed when using personalized numerical models versus the numerical model of the HATS. Finally, the influence of different dimensional parameters on the accuracy of the simulations will be addressed.

3. Material and methods

This study has been approved by the Ethical Committee of the University Hospital of Ghent (Belgium).

3.1 Subjects

Nineteen subjects, eleven female and eight male, between 18 and 48 years old are selected from students and employees at Ghent University. Fifteen of them were inexperienced with respect to hearing protectors, none of them has a history of otological problems. All participated voluntarily and signed an informed consent.

Before the actual measurements are executed, otoscopy is carried out and the hearing of the volunteers is tested with pure-tone audiometry performed by a qualified audiologist in accordance with the modified Hughson-Westlake technique [ISO 8253-1]. This takes place in a sound-proof audiometric cabin at the Audiology Center of the Department of Oto-rhino-laryngology of Ghent University Hospital, using a regularly calibrated Orbiter 922 audiometer. All subjects have normal pure-tone hearing thresholds, i.e. better than 20 dB HL for octave frequencies between 125 Hz and 8000 Hz [Harrel 2002].

Furthermore, the status of the middle ear is checked by carrying out tympanometry at 226 Hz with a ZODIAC 901 tympanometer of Madsen Electronics.
All ears show normal patterns, suggesting that the reflection of sound at the eardrum is not perceptibly influenced by abnormalities of the eardrum or middle ear (Fowler and Shanks, 2002).

### 3.2.2 Hearing protector devices

The custom-made earplugs this study focuses on, are made of hypo-allergic acrylic and have two inner bores (see Figure 1.4): the test bore and a second bore with an adjustable valve determining the attenuation. The test bore is closed in normal wearing conditions, but it allows the insertion of a MIRE probe to assess the hearing protector’s performance.

Measurements by the German Institute for Occupational Safety and Health (BGIA) following ISO 4869-2 (b) reveal that these particular earplugs can offer 25 dB attenuation with an almost open valve (Single Number Rating, SNR) whereas a completely closed valve reaches an SNR of 31 dB. For illustration, the maximal assumed protection values (APV) measured by BGIA are tabulated in Table 3.1.

The earplugs are manufactured for each participant on the basis of an impression of the ear canal taken by a well-trained audiologist. The fitting of each protector is tested using an Attenuation Control Unit (ACU). This device builds up a pressure of 10 mBar in the residual part of the ear canal behind the hearing protector via the test bore. If the pressure holds stable for 2 seconds, the fitting of the hearing protector is considered satisfactory. If not, a new impression of the ear canal has to be made for a new hearing protector. This procedure is repeated until each participant has a pair of perfectly fitting hearing protectors.

Apart from the acrylic protectors, half of the subjects, selected by random allocation, receive an extra pair of custom-made silicone protectors to assess whether the somehow surprising results found with the HATS are also present for human subjects.

### 3.2.3 Measurement setup

#### 3.2.3.1 Material and signal processing

The measurements are carried out in an anechoic room to prevent disturbances from background noise and reflected sound. Except for the HATS itself, all the equipment described in part 2.2.2.1 is used again. Further, a GN ReSound Aurical microphone (see Figure 3.1) is introduced to measure the sound pressure at the eardrum. This device consists of a flexible silicone tube (outer diameter

<table>
<thead>
<tr>
<th>Freq (Hz)</th>
<th>125</th>
<th>250</th>
<th>500</th>
<th>1000</th>
<th>2000</th>
<th>4000</th>
<th>8000</th>
</tr>
</thead>
<tbody>
<tr>
<td>APV (dB)</td>
<td>25.2</td>
<td>26.0</td>
<td>26.4</td>
<td>29.0</td>
<td>29.0</td>
<td>33.7</td>
<td>38.7</td>
</tr>
</tbody>
</table>

Table 3.1: Maximal assumed protection value (APV) for the acrylic earplugs measured by the German Institute for Occupational Safety and Health (BGIA).
0.85 mm) to be inserted in the ear canal, connected to an ear piece with microphone. According to the Aurical’s manual, the tube is inserted 31 mm in the ear canal for male subjects and 28 mm for female participants.

To perform the measurements, the test subject is seated diagonally in front of the loudspeaker at a distance of 1.61 m, his test ear pointing toward the speaker. One free-field microphone is placed symmetrically at the other side with its height set equal to the height of the test ear, all analogous to the setup with the HATS depicted in Figure 2.1. As stated previously in part 2.4, the sound incident direction is less important because Hammershøi and Møller (1996) have shown that the transmission of sound to the eardrum from any point between the eardrum and 6 mm outside the ear canal can be considered directional independent.

Similar to the tests with the HATS, low-pass filtered pink noise is used (cut-off frequency 12.8 kHz) and linear averaging is carried out over 3000 samples with overload rejection. In the frequency range between 0 Hz and 10 kHz the responses are spectrally analyzed using FFT (6400 points).

3.2.3.2 Calibration and post-processing

Once more, the transfer functions calculated from the measurements should be absolutely independent of the test signal, the test space and the microphones’ characteristics. To fulfill these conditions, several calibration steps are carried out similar to the procedure described in part 2.2.2.4.

For the human subjects, Equation 2.2 is not only used to calculate the frequency response $H_{mf}^{(1)}$ between the MIRE measurement microphone ($m$) and the free-field microphone ($f$), but also between the MIRE reference microphone ($r$), the Aurical ($a$) and the free-field microphone; resulting in $H_{rf}^{(1)}$ and $H_{af}^{(1)}$.

Equation 2.3 compensates for possible inhomogeneous sound distribution across the anechoic room, yielding to the frequency responses $H_{mF}^{(2)}$, $H_{rF}^{(2)}$ and $H_{aF}^{(2)}$ with $F$ representing the second free-field microphone. In addition, the influence of the different microphone characteristics is accounted for by subsequently mounting the MIRE measurement microphone, the MIRE reference microphone and the Aurical closely to the second free-field microphone. The microphone’s calibration function $H_{cF}^{x}$ is then calculated over the frequency range of interest $x$ being respectively $m$, $r$ or $a$. By applying these functions in Equation 2.4 the measurements can be corrected and $H_{mF}^{(3)}$, $H_{rF}^{(3)}$ and $H_{aF}^{(3)}$ can be computed. The transfer function $H_{zF}^{x}$ is measured frequently over the test period and its constant character confirms the stability of the microphones’ response.

Finally, the transfer function between the MIRE measurement microphone and the Aurical ($H_{ma}$) on the one hand and between the MIRE measurement microphone and the MIRE reference microphone ($H_{mr}$) on the other can be derived, namely

$$H_{ma} = \frac{H_{mF}^{(3)}}{H_{aF}^{(3)}}$$ (3.1)
3.2. Material and methods

Figure 3.1: GN ReSound Aurical with the silicone tube (a) to be slided into the ear canal, the marker (b) to determine the insertion depth and the microphone unit (c).

and

\[ H_{mr} = \frac{H_{mE}^{(3)}}{H_{rF}^{(3)}}. \] (3.2)

3.2.3.3 Measurement procedure

At the beginning of each test day, the microphones are calibrated using a 4228 pistonphone from Brüel & Kjaer. Otoscopy and 226 Hz tympanometry are carried out for each test subject to ensure that no indication for outer or middle ear abnormalities is present. Next, the ability of the test subjects to fit his hearing protector correctly is checked with the ACU. Finally, the hearing protector is removed from the ear and the subject is seated with his test ear toward the loudspeaker as described previously.

The Aurical is inserted at the appropriate depth, 28 mm for female and 31 mm for male participants, to measure the sound pressure at the eardrum for the unoccluded ear canal. Then, the acrylic earplug (once with open, once with closed valve) and – if applicable– the silicone earplug are placed by the test subject. The MIRE probe is slided at a fixed depth into the test bore by the investigator. The position of the probe is visually inspected and contact with the pinna is avoided as much as possible. In addition, the responses of both MIRE microphones are also registered with the Aurical removed from the ear canal. The order of the different measurements is randomized across subjects and the earplugs are fitted carefully for each test. The investigator always leaves the anechoic room between the different measurement steps. After all tests are completed for one ear, the Aurical’s tube is disinfected.

It is worthwhile realizing that the flexible tube of the Aurical breaks the seal between hearing protector and ear canal and hence might lower the earplug’s
attenuation. This is not problematic since the focus of this study lies on the transfer function between the sound pressure at the MIRE measurement microphone and the sound pressure at the eardrum, which appears to be independent of the earplug’s attenuation (see part 2.3.3).

3.2.4 Individualized numerical simulation

The sound pressure distribution in an ear canal occluded by an acrylic earplug with two inner bores is numerically simulated using the FDTD technique. The approach used for the individualized simulations is based on the model designed for the HATS, the general features of the simulations can hence be found in Section 2.2.4.

In addition, some characteristics of the hearing protector and the outer ear clearly depend on the individual under study. It will be verified whether this mostly geometrical variation can explain and predict the interindividual variation of the transfer functions. Therefore, the most striking features of the hearing protector that are thought to influence sound propagation are accurately measured for each earplug and included in the simulations with a 0.35 mm gridcell size. Figure 3.2 and 3.3 depict the individual geometrical parameters described below.

First, the length of the test bore and second bore determining the attenuation is measured with a caliper accurate up to 0.01 mm. For the subjects under study, the lengths vary between minimal 1.6 cm and 2.3 cm. In the simulations, an extra 0.5 cm is added similar to the simulations for the HATS (see Section 2.2.4.3).

Secondly, the end of the earplug toward the eardrum has a particular shape for each individual. This tip is not flat, but forms a very small pit in which the two inner bores end. The cross-section of this pit is in general more or less elliptic, but the length of the major and minor axis varies, as does the pit’s depth. Moreover, the distance between the two bores’ terminus differs between subjects, just like their distance to the pit’s edge. Finally, the width of the acrylic rim around the pit appears to be typical for each hearing protector. To determine all relevant dimensions of the earplug’s tip, measurements are carried out with the Coordinate-Measurement Machine (CMM) VM-250 Nexiv, manufactured by Nikon and accurate up to 0.1 µm.

The residual part of the ear canal between hearing protector and eardrum is for all participants modeled as a straight tube with uniform cross-section. The length of this tube is the difference between the length of the hearing protector and the length of the unoccluded ear canal. The former is measured with the caliper, the latter is estimated from the first maximum of the frequency response of the unoccluded ear (Pickles 1988). Finally, the diameter of the ear canal is based on the earplug’s diameter.
3.2. Material and methods

<table>
<thead>
<tr>
<th>Number</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>length of residual part of the ear canal ( l_1 )</td>
</tr>
<tr>
<td>2</td>
<td>length of the test bore ( l_2 )</td>
</tr>
<tr>
<td>3</td>
<td>width of the hearing protector</td>
</tr>
<tr>
<td>4</td>
<td>width of the lower rim of the hearing protector</td>
</tr>
<tr>
<td>5</td>
<td>distance between the test bore and the upper rim of the hearing protector</td>
</tr>
<tr>
<td>6</td>
<td>depth of the small pit at the end of the hearing protector</td>
</tr>
<tr>
<td>7</td>
<td>width of the upper rim of the hearing protector</td>
</tr>
<tr>
<td>8</td>
<td>distance between the test bore and the lower rim of the hearing protector</td>
</tr>
</tbody>
</table>

**Figure 3.2:** Individualized geometrical parameters for the FDTD simulations depicted on the vertical cross-section of the earplug in the ear canal taken through the test bore. The parameters in green serve as input variables for multiple linear regression carried out in Chapter 4.
b hearing protector
c test bore
d second bore
9 width of the left rim of the hearing protector
10 distance between the test bore and the rim of the hearing protector
11 distance between the second bore and the rim of the hearing protector
12 width of the right rim of the hearing protector
13 width of the hearing protector
14 distance between the test bore and the second bore

**Figure 3.3:** Individualized geometrical parameters for the FDTD simulations depicted on the horizontal cross-section of the earplug taken through the two inner bores at the part of the hearing protector nearest to the eardrum. The parameters in green serve as input variables for multiple linear regression carried out in Chapter 4.
3.2.5 Statistical analysis

The results from measurements and simulations are explored with different statistical techniques all carried out with the statistical software R. They are discussed in detail in the result-sections they apply to.

3.3 Results

3.3.1 Measured transfer functions

The transfer function for the silicone earplugs confirms the results from the HATS, meaning that an enlarged plateau is seen in the higher frequencies. Combining this knowledge with the considerations in part 2.3.1.2 it seems clear that the difference between silicone and acrylic earplugs is due to the earplug’s material and not to problems with the HATS. This suggests that the implementation of this MIRE approach with silicone earplugs needs reconsideration before further analysis are meaningful and therefore the remainder of this chapter will focus on the acrylic earplugs.

Figure 3.4 shows for all ears the transfer function with open valve between the MIRE measurement microphone and the Aurical \( H_{ma} \), which is very similar to the measurements with closed valve. A positive function value implies that the MIRE measurement microphone registers a higher sound pressure level than the Aurical, obviously the opposite is true for a negative value.

It becomes clear that the majority of all transfer functions has the same global form with constant values around 0 dB for the lower frequencies, a distinct maximum between 2500 Hz and 3500 Hz and multiple minima above 4500 Hz, with often the most pronounced minimum between 4500 Hz and 6500 Hz. This morphology is in accordance with the transfer function measured on the HATS (see part 2.3.1.1), deviant spectra will be discussed in part 3.3.2.

Combining the results from both studies leads to the conclusion that the first maximum in Figure 3.4 is most probably caused by resonance in the test bore, picked up by the MIRE measurement microphone, but not by the Aurical. Additionally, the most distinct minimum is most likely due to resonances in the residual part of the ear canal behind the hearing protector, registered by the Aurical but not by the MIRE measurement microphone. However, despite this common global shape of most transfer functions, the intersubject variability appears substantial with respect to the exact frequency and amplitude of the extrema.

Within one ear, the measurements with open and closed valve resemble each other closely for most subjects and only minor shifts are seen with respect to the frequencies at which the extrema occur, see Figure 3.5 as example. Minimal deviations in the first maximum’s frequency might be due to the position of the MIRE probe in the test bore. Although the probe is as accurately as possible inserted at a fixed depth, minor differences in the exact position can not be avoided. In a similar way, the position of the earplug in the ear canal might also
Figure 3.4: Magnitude of all measured transfer functions between the MIRE measurement microphone and the Aurical microphone for acrylic earplugs with open valve.

Figure 3.5: Example of similarity between simulation and measurements for one particular ear: magnitude of the measured transfer functions with closed (‘closed’) and open (‘open’) valve and magnitude of the simulated transfer function (‘model’).
slightly differ between measurements, causing differences in the exact frequency of the minima above 4500 Hz.

For some participants, the amplitude of the transfer functions with closed valve tends to be more negative than with open valve in the higher frequency region (Figure 3.5). Since analyses for the HATS suggest that the transfer function is independent of the applied attenuation (see part 2.3.3), amplitude variation with different positions of the valve are most likely introduced by the measurement setup. Part 2.3.1.1 also suggest that the MIRE measurement microphone can not be held responsible for the observed deviation since similar aberrations have never systematically occurred when measuring with the HATS.

Hence, the Aurical microphone seems to register in proportion a higher sound pressure level than the MIRE measurement microphone in the condition with closed valve compared to the measurements with open valve. This might be at least partially caused by sound reaching the Aurical microphone unwantedly through the walls of the flexible tube. This becomes potentially important in the conditions where the sound pressure behind the hearing protector is very low, i.e. for earplugs with closed valve. The hypothesis is supported by the fact that the effect is merely seen in the higher frequency range where the attenuation of a passive hearing protector reaches a maximal value. Therefore, the transfer functions measured with open valve might be more reliable in the higher frequencies.

To sum up, the close resemblance between the measurements with open and closed valve suggests that the transfer functions for a particular ear can be registered in consistent and hence reliable way. Nevertheless, the results also suggest that errors up to 10 dB can be expected when an individual’s transfer function is predicted from the mean transfer function measured on a sample of test subjects. Hence, more detailed prediction models like numerical FDTD simulations are clearly needed.

### 3.3.2 Numerical FDTD simulations

For the ears which measured transfer functions follow the global trend, the simulated transfer functions are very similar to those measured, an example is given in Figure 3.5. At first glance, the frequency dependence at the lower frequencies seems very well predicted. As for the amplitude, all the numerical simulations have a constant magnitude of 0 dB whereas most measurements reach constant values between 0 dB and 5 dB. It is experimentally verified that bending the probe tube can indeed lower the response of the Aurical microphone up to 5 dB in the lower frequency range. Flexures of the tube could not be avoided due to the relatively difficult positioning of the tube in an ear canal occluded by an earplug, but the influence of the bends can be clearly identified and hence the difference between the simulated and measured transfer function below 1500 Hz is not considered critical.

Further, the frequency and amplitude of the first maximum are very well approached by the model, as is the frequency of the most distinct minimum between 4500 Hz en 6500 Hz. However, model and measurements tend to differ
in the exact amplitude of this minimum. For frequencies above 6500 Hz, the numerical model and the measurements still show resembling frequency dependence but the resemblance is decreased compared to the frequency region below 6500 Hz. These findings clearly confirm the results obtained with the HATS (see part 2.3.1.1).

By contrast, the numerical simulations are unable to predict the measured frequency dependence for the ears of which the measured transfer functions clearly differ from the average in shape. Hence, the question arises whether the obtained transfer functions are reliable and the numerical simulations produce a wrong forecast, or, on the contrary, whether the measured transfer functions are incorrect and therefore can not be approached by the numerical model.

In this regard, Figure 3.4 shows that all anomalous transfer functions have a more positive difference between the MIRE measurement microphone and the Aurical, meaning either that the MIRE measurement microphone registers a higher sound pressure level or that the Aurical receives a lower level.

Following the first hypothesis, the attenuation derived from the transfer function between the two MIRE microphones ($H_{mr}$) with the Aurical tube in place is expected to be lower for the ears with abnormal results. However, this is not the case, the ears with normal transfer functions appear to have a systematically lower attenuation when the valve is closed, especially for the frequencies below 1500 Hz (see Figure 3.6).

To verify whether the observed difference in attenuation is only due to random error in this sample or whether it can be generalized, a two-way ANOVA ($\alpha = 0.05$) is carried out with ear (left or right) and difference from the average transfer function in shape (low or high) as fixed factors and the perceived attenuation below 1500 Hz as dependent variable. The classification between high and low deviation could be easily made by visual inspection of the transfer function’s spectra. Before the results are interpreted, the aptness of the model is checked by looking at the standardized and studentized residuals (Kutner et al. 2004). No interaction effect between the factors ‘ear’ and ‘difference’ ($p = 0.14$) is found, hence the main effects may be interpreted. No significant difference is found for the attenuation of left and right ears ($p = 0.41$), but the mean attenuation is clearly higher in the ears with a distinctly different transfer function ($p < 0.001$).

From this analysis, it becomes clear that the irregular transfer functions are not caused by an excessive registration of the sound pressure by the MIRE measurement microphone but by an underestimation of the sound level by the Aurical. In the frequency range below 1500 Hz, the attenuation is the most sensitive to leakage because lower frequencies are more easily transmitted by the air column between hearing protector and ear canal (Apfel 1997). Hence, the difference in attenuation suggests that the flexible tube of the Aurical causes less leakage for ears with deviant transfer function than for ears with regular ones. This leads to the assumption that the Aurical’s tube must be more or less squeezed in the former ears, hindering the transfer of sound to the Aurical’s microphone, decreasing the finally registered sound pressure level.

The above reasoning yields to the conclusion that the aberrant transfer func-
3.3. Results

Figure 3.6: Closed-valve attenuation below 1500 Hz with Aurical tube in place (mean value and standard deviation) for left (blue) and right (red) ears with a high (‘High’) and low difference in shape (‘Low’) from the average transfer function.
tions do not reflect the true differences in sound pressure between the MIRE measurement microphone and the eardrum, but are caused by measurement error. Hence, these results will not be taken into account for further analysis.

### 3.3.3 Aptness of an individualized model

As described previously, the numerical simulations are on the one hand based on general characteristics of an ear canal occluded by an earplug (part 2.2.4) and on the other on a set of parameters unique for each ear (Section 3.2.4). In the next section, the aptness of an individualized model will be studied.

For this analysis, the quadratic error is calculated by squaring the difference between the model and the corresponding measurements. Further, this difference is summed for the four frequency regions that can be clearly distinguished; the lower frequency region between 100 Hz and 1500 Hz where the function has a merely constant value, the frequency region between 2500 Hz and 4500 Hz which encloses the first maximum, the region between 4500 Hz and 6500 Hz where the most distinct minimum is seen and finally the frequencies between 6500 Hz and 8000 Hz.

Since the acoustical features of the HATS mimic the characteristics of an average human body, it might be possible to approach the transfer function of an individual with numerical simulations based on the acoustical and geometrical characteristics of a HATS. To find out whether the individualization of these geometrical features does enhance the resemblance between measured and predicted transfer functions, the quadratic error between the measurements and respectively the individualized model and the model of the HATS is compared. This is done by separate pairwise tests for the different ears and the different positions of the valves. Because of the relatively small sample sizes (less than 20), normality tests might be less reliable and hence it is not always possible to state with sufficient accuracy that the conditions for the paired student t-test are fulfilled. Therefore, the non-parametric Wilcoxon signed-rank test is also performed (Moore and McCabe, 1999). For both statistical test, α is set at 0.05. These analyses are not carried out for the lowest frequency range because the individualized models and the model of the HATS have exactly the same constant amplitude around 0 dB.

For the four combinations of right and left test ear with open and closed valve, the quadratic error for the HATS model seems to exceed the quadratic error calculated from the individualized model. This is illustrated by Figure 3.7, depicting the quadratic error for the right ear with open valve. From Table 3.2 can be seen that both the parametric and non-parametric analysis consequently yield to a statistically significant higher quadratic error for the majority of the test situations when the model of the HATS is used (p < 0.05). Only the left ears with closed valve fail to give statistically significant results in the two higher frequency regions (p > 0.05), possibly due to the smaller amount of data in this group (Moore and McCabe, 1999). This analysis yields to the conclusion that individualized numerical simulations will in general lead to a more accurate prediction of the transfer function between the MIRE measurement microphone.
Figure 3.7: Quadratic error for the right ears between the measured transfer function with open valve and respectively the individualized model (gray) and the model based on the HATS (white), calculated for the frequency region from 2.5 kHz to 4.5 kHz, from 4.5 kHz to 6.5 kHz and from 6.5 kHz to 8 kHz. The boxes span the middle half of the ordered observations and the thick black lines inside represent the median. The whiskers extend to the most extreme data point which is no more than 1.5 times the interquartile range from the boxes. The circles represent data points that fall outside these limits.
Table 3.2: For the different test situations (left ear with closed valve (L.C.), left ear with open valve (L.O.), right ear with closed valve (R.C.) and right ear with open valve (R.O.)), the quadratic error between the measurements and the corresponding individualized model is statistically compared to the quadratic error between measurements and the HATS model. The p-values from the paired student t-test (upper number) and from the Wilcoxon signed-rank test (lower number) are tabulated, as is the number of data points (between brackets).

<table>
<thead>
<tr>
<th></th>
<th>2.5 kHz - 4.5 kHz</th>
<th>4.5 kHz - 6.5 kHz</th>
<th>6.5 kHz - 8 kHz</th>
</tr>
</thead>
<tbody>
<tr>
<td>L.C. (13)</td>
<td>0.0003</td>
<td>0.09</td>
<td>0.7</td>
</tr>
<tr>
<td></td>
<td>0.0004</td>
<td></td>
<td></td>
</tr>
<tr>
<td>L.O. (15)</td>
<td>0.002</td>
<td>0.0003</td>
<td>0.04</td>
</tr>
<tr>
<td></td>
<td>6.0e-05</td>
<td>0.0001</td>
<td>0.04</td>
</tr>
<tr>
<td>R.C. (15)</td>
<td>0.0001</td>
<td>0.01</td>
<td>0.01</td>
</tr>
<tr>
<td></td>
<td>0.0002</td>
<td>0.008</td>
<td>0.02</td>
</tr>
<tr>
<td>R.O. (16)</td>
<td>0.001</td>
<td>0.002</td>
<td>0.0004</td>
</tr>
<tr>
<td></td>
<td>3.0e-05</td>
<td>9.0e-05</td>
<td>0.001</td>
</tr>
</tbody>
</table>

and the sound pressure of interest at the eardrum.
Apart from this conclusion, the question also arises whether the general characteristics of the model are suitable for both men and women. In particular, different studies state different conclusions upon the gender dependency of energy reflectance at the eardrum. Some declare that no statistically significant differences are found (Johansson and Arlinger, 2003), while others state the opposite (Feeney and Sanford, 2004). Possible influence of gender in this study is assessed by testing the difference in quadratic error for men and women for different ears and different positions of the valve. The comparison is made for the frequency region between 100 Hz and 1.5 kHz, between 2.5 kHz and 4.5 kHz, between 4.5 kHz and 6.5 kHz and finally between 6.5 kHz and 8 kHz. For each separate category, a non-parametric Mann-Whitney test ($\alpha = 0.05$) is carried out because of the limited dataset (Moore and McCabe, 1999). None of the analysis yields to a statistical significant difference between the quadratic error for men and women ($p > 0.05$), hence from this dataset it seems unnecessary to provide different male and female general characteristics for the numerical simulations.

3.3.4 Influence of geometrical characteristics
A final issue is the possible relationship between the quadratic error and the geometrical parameters. The sound propagation in ears with more extreme values for certain geometrical parameters might be more difficult to model correctly, hence the quadratic error between the simulated and the measured transfer functions might increase.
This question is addressed by drawing scatter plots between each geometrical characteristic and the quadratic error for the different frequency regions de-
scribed earlier. No distinction is made between left and right ears because each ear has its individual set of parameters. Further, the quadratic error is calculated for both the measurements with open and with closed valve to include as much information as possible. On these scatter plots, a polygon that best fits the distribution of the points is drawn automatically to aid the eye in seeing important patterns (Wickham, 2009). The vast majority of the scatter plots shows randomly distributed points and hence no relationship between these particular geometrical variables and the quadratic error can be deduced from this dataset, making further statistical analysis pointless. However, some plots do reveal a certain correlation between the variable under study and the quadratic error for one or more frequency regions; this relationship is assessed in more detail.

Figure 3.8 reveals that the quadratic error from 2.5 kHz to 4.5 kHz tends to be higher for hearing protectors with a limited number of 0.35 mm gridcells in the x-direction between the ear canal’s wall and the starting point of air in the pit at the end of the hearing protector. This quadratic error also seems to increase slightly with increasing cell numbers. Comparing the difference in quadratic error for ears with a more extreme rim (2 cells or 5 and more cells) on the one hand and ears with three and four cells on the other yields to statistically significant differences for both the two sample t-test \( (df = 37.004; \ p = 0.002 < 0.05) \) and the Mann-Whitney test \( (p = 0.001 < 0.05) \).

Although these data have to be interpreted with some caution because of the possible correlation between the different points, they do seem to confirm the tendencies visible in Figure 3.8. This might be explained by the fact that a very small rim or very large rim implies a more abrupt transition between the earplug’s inner bore and the ear canal. The boundary layer theory used in this approach is less suitable to model the viscothermal effects at more curved edges (see part 2.4), possibly increasing the difference between simulations and measurements.

The quadratic error for the frequency region between 6.5 kHz and 8 kHz seems less randomly spread for eleven geometrical parameters, all related to dimensions in the x- or y-direction. However, the observed trends are in essence introduced by the left and right ear of one participant apparently having small cross-sectional characteristics and an enlarged quadratic error in this frequency region. The fact that the aberrant data belong to one participant somewhat undermines the hypothesis of a causal relationship between the ear canal’s dimensions and the associated quadratic error. This idea is further refuted by the finding that for most parameters more extreme geometrical values are also found for ears with only limited quadratic error.

### 3.4 Discussion

As stated previously, one important issue in the MIRE approach under study is the difference between the sound pressure measured by the MIRE measurement microphone and the sound pressure of interest at the eardrum. In this regard, the results obtained with a HATS (see Chapter 2) indicate that the transfer
Figure 3.8: Relationship between the quadratic error for the frequency region from 2.5 kHz to 4.5 kHz and the number of cells in the x-direction between the ear canal and the starting point of air in the pit at the end of the hearing protector.
functions between the sound level at the MIRE measurement microphone and at the eardrum are stable, reproducible and clearly related to the morphology of earplug and test bore. The current chapter elaborates the question whether tests with human subjects yield to similar conclusions.

For the new measurements, the equipment and setup described in part 2.2 are merely copied since they have proved themselves worthy (see part 2.4). In addition, a GN ReSound Aurical microphone is introduced to register the sound pressure at the volunteers' eardrum, a procedure supported by different studies. For instance, Hellstrom and Axelsson (1993) have demonstrated that a miniature microphone with a probe tube attached is useful for sound measurements in the outer ear canal. Their results indicate that a probe tube with a larger outer diameter than the tube in this study, respectively 1.5 mm and 0.85 mm, has negligible influence on the sound pressure in front of the eardrum. Moreover, this study’s registration point close to the tympanic membrane induces least errors in the case of small movements of the probe tube (Hellstrom and Axelsson, 1993). In closing, the generalization of sound pressure in front of the entire eardrum from the sound transmitted by the probe tube seems also justified. Koike et al. (2002) have demonstrated that the distribution of pressure in front of the eardrum is almost uniform at 0.1 kHz and at 7 kHz differences are within a 2 dB-range.

The thus measured transfer functions vary substantially between themselves, but most of them follow the same global frequency dependence as the HATS. This is in accordance with results from Hammershøi and Møller (1996) who have measured the sound transmission to the eardrum from the entrance of the ear canal for different subjects. Furthermore, analyzing the transfer functions with deviant spectra yields to the conclusion that those suffer from measurement errors and do not reflect true fluctuations in sound pressure amplitude between the registration points (see part 3.3.2). The resemblance between transfer functions for a HATS and human subjects also confirms that simulator approaches quite well the acoustics of a human head and torso, only the maximum around 1000 Hz is probably due to a certain mismatch between the ear simulator and a human auditory system.

As for the HATS, measurements are completed with numerical simulations to assess the possibility of predicting an individual’s transfer function. In the concrete, FDTD simulations are developed by combining general acoustical characteristics of an ear canal occluded by an earplug with individual geometry. In general, the numerical approach seems to predict the measured frequency dependence quite accurately in the frequency region below 5500 Hz. Above this frequency, the numerical simulations still follow the trend of the measurements, but differences in amplitude increase.

It is indeed possible that the numerical model is less accurate for higher frequencies. In the current approach, the outer ear canal is for instance modeled as a straight tube with uniform cross-section whereas in reality the ear canal is bended with variable diameter (Koike et al., 2002). The bended character is less important in this project because the two bends are located in the cartilaginous part which is merely filled with the hearing protector (Dillon, 2001). Still,
variation in cross-section might influence the transfer function in the higher frequencies (Koike et al., 2002). In addition, the vibration of the tympanic membrane becomes more complex with increasing frequency and clearly differs from the simple low-frequency pattern included in this model for frequencies higher than 5000 Hz (Tomndorf and Khanna, 1972).

Besides these considerations, the question arises whether it is actually possible to approach the true transfer function correctly in this higher frequency region. Leaving aside the small intrasubject variability (Hellstrom and Axelsson, 1993) and measurement difficulties for high attenuation values (Section 3.3.1), even very small variations in the position of the earplug may distinctly alter the frequency response of the ear canal’s residual part, changing the appearance of the transfer function above 4500 Hz. In this regard Berger et al. (2007) and Voix and Zeidan (2009) describe the issue of increasing variability with increasing frequency when measuring a hearing protector’s attenuation.

Apart from the straight ear canal, other characteristics of the numerical simulations are common for all subjects and based on the model for a HATS (see part 2.2.4), for instance the impedance defining the ossicular part of the ear canal. Since none of the participants has a history of otological problems and none of the audiological tests reveal any abnormalities, it seems correct to apply the impedance of bone in all cases.

Further, the same model for the impedance at the eardrum has been used for the different simulations because all subjects have normal tympanometric results. As discussed in Section 3.3.3, some studies have found small gender differences in middle ear impedance. Taken into account the limited number of participants in this study, the quadratic error between measurements and simulations does not differ significantly between men and women, suggesting that for this application the model of the eardrum impedance is adequate for both genders. A similar discussion exists upon the influence of age on the characteristics of the middle ear (Feeney and Sanford, 2004). Since the possible aging effect is mostly suspected for elderly subjects, it does not seem critical for this model designed for the working population.

In contrast with the models’ common properties, the individualized features are based on specific geometrical dimensions of each ear and hearing protector. The comparisons of the quadratic error between the measurements and both the HATS model and the individualized model reveal that the individualized approach matches the measurements better and therefore appears advisable. Moreover, for most geometrical characteristics the models seem suitable for a considerable range of possible values.

### 3.5 Conclusion

The satisfying concordance between the numerical approach and the registered transfer functions for acrylic earplugs clearly shows that an individualized transfer function between MIRE microphone and eardrum might be accurately predicted using a partially individualized FDTD model. The next chapter will
assess the possibility to implement the individualized transfer function in measurements at the workfloor. This will be done for the acrylic earplugs only since the implementation of this MIRE approach for silicone earplugs clearly needs further research (see part 3.3.1).
3. Verification in situ: method validation with human subjects
Chapter 4

Filters for accurately verifying the performance of hearing protectors in situ

The content of this chapter is based on the technical report submitted to Acta Acustica united with Acustica (Bockstael et al., Accepted).

4.1 Introduction

The previous chapters have clearly demonstrated that the transfer function between sound pressure at the MIRE measurement microphone and the level of interest at the eardrum are stable and reproducible within one subject, but they also indicate that the intersubject variation is substantial (see part 3.3.1). Fortunately, the transfer function of an individual ear can be approached using FDTD models that combine general acoustical features of the ear canal and earplug with individualized geometrical aspects (see Section 3.3.2).

In theory, one could determine all geometrical characteristics listed in Section 3.2.4 for each worker’s hearing protector, include them in the individualized numerical simulations and calculate the applicable transfer functions. These transfer functions can then be used to predict the sound pressure at the eardrum from in situ MIRE measurements. However, major practical constraints make this approach less feasible. First, the detailed measurements of the earplugs require specialized equipment and are time-consuming. Secondly, the computational cost for the FDTD model is also quite large, making the incorporation in field measurement equipment nearly impossible.

The above reasoning clearly illustrates that the current approach - well-suited in experimental and research settings - needs simplification for practical implementation. Based on the relatively simple shape of the transfer function and the striking relationship with the geometry of the test bore and ear canal, one
might assume that a one-dimensional analytical model is sufficiently accurate to predict the sound pressure at the eardrum in practice. This approximation is elaborated in Section 4.2.1, but the results distinctly show that the sound propagation under study is much more complicated than captured in the analytical model.

To overcome this problem, a filter-based approach is chosen instead because digital filters can easily be included in computers and measurement equipment. In the concrete, the FDTD transfer functions serve as a starting point to propose a set of filters suitable to predict the sound pressure at the eardrum from measurements with the MIRE probe. In this matter, the apparent relationship (see Chapter 3) between the exact spectrum of the transfer functions and the morphology of a particular ear should be included. Therefore multiple linear regression and multivariate orthonormal vector fitting (MOVF) are applied to predict for one specific ear the applicable filter, linking the filter characteristics to the most influential geometrical parameters of the ear canal and earplug. Naturally, the aim is to achieve the right balance between the number of needed morphological variables – as few as possible – and the accuracy of the filter – as high as possible.

4.2 Material and methods

4.2.1 One-dimensional analytical model

4.2.1.1 Key factors of the one-dimensional model

In Figure 4.1, a schematic overview is drawn of an occluded ear canal; pressure \( p \) and particle velocity \( v \) are calculated along the \( x \)-axis in the test bore and residual part of the ear canal. For the calculations, a plane wave front is assumed, meaning that each acoustic quantity is uniform over any plane surface normal to the direction of propagation. Further, the same acoustical properties hold everywhere in the fluid (homogeneity) and they are directionally independent (isotropy) (Fahy, 2001). Under these assumptions the expressions for \( p \) and \( v \) are written in the frequency domain as a function of space and this for harmonic plane waves. For the sound propagation in the test bore, the following equations are then applicable

\[
p = Ce^{-jkx} + De^{jkx} \tag{4.1}
\]

\[
v = \frac{C}{\rho c}e^{-jkx} - \frac{D}{\rho c}e^{jkx} \tag{4.2}
\]

with \( C \) and \( D \) the amplitude of the respectively positive- and negative-going wave, \( k \) the wave number, \( x \) the location in space, \( c \) the speed of sound and \( \rho \) the density of the medium. In the ear canal analogously holds that

\[
p = Ae^{-jkx} + Be^{jkx} \tag{4.3}
\]
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Figure 4.1: Schematic overview of the ear canal and test bore as included in the one-dimensional analytical approximation; pressure and particle velocity are calculated along the $x$-axis in the test bore and the residual part of the ear canal.
4. Verification in situ: filter approaches

\[ v = \frac{A}{\rho c} e^{-jkx} - \frac{B}{\rho c} e^{jkx} \]  \hspace{1cm} (4.4)

with \( A \) and \( B \) the amplitude of the respectively positive- and negative-going wave. The transfer function between the sound pressure at the MIRE measurement point (at \( x = l_1 + l_2 \)) and the eardrum (at \( x = 0 \)) can then be written as

\[ \frac{p_{(l_1+l_2)}}{p_0} = \frac{Ce^{-jk(l_1+l_2)} + De^{jk(l_1+l_2)}}{A + B}. \]  \hspace{1cm} (4.5)

To solve this equation, the unknown variables \( C, D, A \) and \( B \) have to be written in function of known quantities as will be elaborated in this section.

**Amplitude \( A \) in function of \( B \) at \( x = 0 \)**

At the eardrum, the sound pressure can be written as

\[ p_0 = A + B \]  \hspace{1cm} (4.6)

and the particle velocity as

\[ v_0 = \frac{A - B}{\rho c}. \]  \hspace{1cm} (4.7)

With the specific acoustic impedance of the eardrum \( Z_t \), the relationship between \( p \) and \( v \) can be written as (Fahy, 2001)

\[ p_0 = Z_t v_0; \]  \hspace{1cm} (4.8)

filling in the expressions from Equations 4.6 and 4.7 then yields to

\[ A + B = Z_t \left( \frac{A - B}{\rho c} \right). \]  \hspace{1cm} (4.9)

From Equation 4.9 \( A \) can be easily expressed in function of \( B \)

\[ A = -\frac{1}{1 - \frac{Z_t}{\rho c}} B. \]  \hspace{1cm} (4.10)

**Amplitude \( C \) in function of \( D \) at \( x = l_1 + l_2 \)**

Analogous to the previous derivation, \( p \) and \( v \) can be written at the MIRE measurement point as

\[ p_{(l_1+l_2)} = Ce^{-jk(l_1+l_2)} + De^{jk(l_1+l_2)} \]  \hspace{1cm} (4.11)

and

\[ v_{(l_1+l_2)} = \frac{C}{\rho c} e^{-jk(l_1+l_2)} - \frac{D}{\rho c} e^{jk(l_1+l_2)} \]  \hspace{1cm} (4.12)

which leads in combination with the specific acoustic impedance of the MIRE measurement microphone \( Z_m \) to

\[ Ce^{-jk(l_1+l_2)} + De^{jk(l_1+l_2)} = Z_m \left( \frac{C}{\rho c} e^{-jk(l_1+l_2)} - \frac{D}{\rho c} e^{jk(l_1+l_2)} \right), \]  \hspace{1cm} (4.13)
in its turn giving an expression for \( C \) in function of \( D \)

\[
C e^{-jkl_1} = -\frac{1 + \frac{Z_m}{\rho c}}{1 - \frac{Z_m}{\rho c}} e^{jk(l_1+2l_2)} D.
\] (4.14)

**Amplitude \( D \) in function of \( B \) at \( x = l_1 \)** At the transition between the test bore and the ear canal (\( x = l_1 \)) the law of conservation of mass (Fahy, 2001) states that the net mass flow into a finite volume around the interface must be zero, which implies that the mass flow rate from the test bore must be equal to the mass flow rate into the residual volume or

\[
\rho S_2 v_{l_1,R} = \rho S_1 v_{l_1,L};
\] (4.15)

assuming that \( \rho \) remains constant and applying Equation 4.4 and 4.7 gives then

\[
S_2 \frac{C}{\rho c} e^{-jkl_1} - D \frac{e^{jkl_1}}{\rho c} = S_1 \left( \frac{A}{\rho c} e^{-jkl_1} - B \frac{e^{jkl_1}}{\rho c} \right).
\] (4.16)

With Equation 4.10, 4.14 and 4.17, \( A \) and \( C \) are removed from the previous equation and thus \( D \) can be written solely in function of \( B \) as

\[
D = B \frac{S_1}{S_2} \frac{\frac{1}{1 + \frac{Z_m}{\rho c}} e^{-2jkl_1} - 1}{\frac{1 + \frac{Z_m}{\rho c}}{1 - \frac{Z_m}{\rho c}} e^{2jkl_2} - 1}.
\] (4.17)

**One-dimensional transfer function** The transfer function between the MIRE measurement point and the eardrum can be derived if \( A \), \( B \) and \( C \) are substituted in Equation 4.5 using respectively Equations 4.10, 4.14 and 4.17. This yields to a one-dimensional expression for the transfer function under study

\[
\frac{p_{l_1+l_2}}{p_0} = \frac{S_1}{S_2} \frac{-\frac{1 + \frac{Z_m}{\rho c}}{1 - \frac{Z_m}{\rho c}} + 1}{\frac{1 + \frac{Z_m}{\rho c}}{1 - \frac{Z_m}{\rho c}} + 1} \frac{e^{jk(l_1+l_2)}}{-\frac{1 + \frac{Z_m}{\rho c}}{1 - \frac{Z_m}{\rho c}} e^{-2jkl_1} - 1}{\frac{1 + \frac{Z_m}{\rho c}}{1 - \frac{Z_m}{\rho c}} e^{2jkl_2} - 1}.
\] (4.18)

**4.2.1.2 Results** To implement the one-dimensional model the same expressions are used for the impedance at the eardrum and at the MIRE measurement microphone as in the more detailed FDTD model (see Section 2.2.4.3). Further, the length of the test bore and the residual part of the ear canal derived in Section 3.2.4 are included to establish individualized one-dimensional models analog to the FDTD models. However, Figure 4.2 shows that the here included one-dimensional approach is over-simplified. The first maximum is elevated up to unrealistically high values and in addition shifted to higher frequencies, thus almost totally obscuring the minimum present in the measurements and FDTD models. This result is not
entirely unexpected, because certain assumptions might not be completely fulfilled in this setup. First, the mass conservation equation or continuity equation might not be accurate at the transition between the test bore and the ear canal (at $x = l_1$) because the difference in diameter is large. In Helmholtz resonators, an end correction is introduced to take into account the near field of the vibration near the neck-end. This yields to replacing the length of the resonator neck by a (larger) effective length which would indeed shift the first maximum to lower frequencies. Secondly, the continuity equation is not completely correct at small distances (compared to the wave length) from the transition. This means that the model is not adequate for small residual volumes because the wave front at the eardrum is no longer plane. Thirdly, the viscothermal losses can not be neglected in ducts with small cross-section. The exponential damping can be introduced in the relation between the wave number and the frequency by including the absorption coefficient or by the approach proposed by Egolf (1977).

The above reasoning clearly shows that a relatively simple one-dimensional model has some obvious shortcomings. Viscothermal losses in the hearing protector’s bores can be introduced quite easily, but it is much more complicated to include correctly the transition between earplug and ear canal with the narrow bores discharging into the much wider ear canal via a particularly shaped pit. To handle this issue, it seems more efficient basing a simplified model on the accurate and well-studied FDTD simulations than conducting new and extensive
4.2. Material and methods

4.2.2 Multiple linear regression

4.2.2.1 Filter design

In Chapter 3 FDTD simulations are carried out for nineteen subjects. Each simulation combines the general acoustical properties of an acrylic earplug and ear canal with individualized geometrical features. This approach yields to 37 different numerical models, one model per ear. In this matter, one ear is left out because the results obtained with the VM-250 Nexiv (see Chapter 3.2.4) appeared to be erroneous. In this chapter, the FDTD transfer functions are approached by IIR filters because digital filters, and especially FIR and IIR filters, can easily be included in computers and measurement equipment. One of the standard methodologies for the design of such a filter is used; the complex frequency response of each simulated transfer function is approximated with a continuous-time transfer function being the quotient of two polynomials. If so desired, the fitted transfer function can be transformed into a discrete transfer function by applying a bilinear transformation.

To accomplish all this, invfreqs from MATLAB (The MathWorks™) is used because this algorithm guarantees stability of the resulting linear system. The corresponding complex frequency response $H(s)$ can be written as

$$H(s) = \frac{b_1 s^n + b_2 s^{n-1} + \ldots + b_{n+1}}{a_1 s^n + a_2 s^{n-1} + \ldots + a_{n+1}}$$  \hspace{1cm} (4.19)$$

with $s = j 2 \pi f$, $f$ representing the frequency. The frequency range of interest is set between 0 Hz and 8000 Hz analogous to the frequencies tested with pure-tone audiometry (ISO 8253-1). The filter coefficients are deliberately determined in the $s$-domain instead of the $z$-domain. In that way, the resulting filter can be digitalized afterward with a sampling frequency adapted to the sampling frequency of the measurement system used in practice. The order of $A(s)$ and $B(s)$ is chosen as low as possible, provided that the frequency response of the analogue filter $H(s)$ is almost identical to the FDTD simulated transfer function. It appears that this requirement is fulfilled if the order of both $A(s)$ and $B(s)$ is set at 6.

4.2.2.2 Linear regression

The aim of the multiple linear regression is to find a formal relationship between the coefficients of $H(s)$ and the geometrical variables of the ear canal and earplug. This way, an individual transfer function, resulting from an individualized filter, can be used in the measurement equipment to predict the sound pressure at the eardrum from the response of the MIRE measurement microphone without the need to perform a detailed FDTD simulation first. For stability reasons, linear regression will not be carried out with the coefficients...
of $A(s)$ and $B(s)$ but instead the poles and zeros, i.e. the roots of respectively $A(s)$ and $B(s)$, serve as dependent variables.

**Dependent variables** For most ears, only the first and second zero appear to be real, all other poles and zeros are complex. For the latter, the real and the imaginary part are fitted separately using linear regression. Because all coefficients in Equation 4.19 are real, the complex conjugate of each complex pole or zero is also a pole or zero of the filter under study. In that case, only the real and imaginary part of the pole/zero with a positive imaginary part are considered for the regression analysis. The corresponding complex conjugate can then be easily deduced from the resulting formulas.

**Independent variables** The possible independent variables of the linear regression are the geometrical parameters seen in both cross-sections of the earplug and ear canal, depicted in Figure 3.2 and 3.3. Linearly dependent variables are omitted.

**Building the regression model** All statistical analyses are carried out with the statistical software R. Based on scatter plots and Pearson correlation coefficients between dependent and independent variables, a manual step-forward regression procedure is followed. The procedure is repeated until the adjusted $R^2$ equals or exceeds 0.80. When extra independent variables are added, care is taken that they are not correlated to the variables that are already included in the model and that there are always 6 to 10 observations per variable included in the regression model (Kutner et al., 2004). This procedure results in an expression to estimate the expected value of the dependent variable $\hat{y}$ based on one or more independent variables $x_i$.

**Checking the underlying assumptions** Before the obtained models are actually used to predict the poles and zeros for a particular transfer function, the aptness of the assumptions of linear regression are checked first (Kutner et al., 2004). The assumptions that the residuals come from a normal distribution is verified with the Kolmogorov-Smirnov test and the Shapiro-Wilk normality test. If the distribution of the residuals is far from normal ($p<0.01$), the regression model is reconsidered. Further, the random distribution of studentized residuals around zero is visually assessed by drawing a scatter plot. Because left and right ears of the same subject are included, possible autocorrelation between the residuals is examined by calculating the Durban Watson statistic. Outliers and influential observations are detected by computing the Mahalanobis distance, the df-betas, the Cook’s distance and the leverage. The Mahalanobis distance measures the influence of a case $i$ on the fitted value $\hat{Y}_i$ whereas the Cook’s distance considers the influence of the $i$th case on all $n$ fitted values. Subsequently, the df-beta measures the influence of the $i$th case on each regression coefficient. Finally, the leverage measures the difference between the vector
of the \(i\)th observations of the independent variable and the vector of means of all independent variables (Kutner et al., 2004). The values of influential observations are carefully inspected and action is undertaken whenever necessary.

4.2.2.3 Numerical results

For the real part of the first pole (\(\Re(p_{1})\)), the real (\(\Re(p_{3})\)) and imaginary (\(\Im(p_{3})\)) part of the third pole and the imaginary part of the fifth pole \(\Im(p_{5})\), the regression models could be established without any problems, including all observations of the dataset. The resulting models are summarized in Table 4.1, revealing that only the length of the test bore \(l_{2}\) and the length of the residual part of the ear canal \(l_{1}\) are needed as independent variables. For all the linear models, both normality tests yield to insignificant p-values (\(p > 0.01\)). Furthermore, none of the values of influential observations are unlikely nor is there reason to believe that they have been measured incorrectly.

For the real part of the fifth pole (\(\Re(p_{5})\)) and zero (\(\Re(z_{5})\)), the adjusted \(R^2\) does not reach 0.80. Since adding more variables apart from \(l_{1}\) and \(l_{2}\) does not substantially increases this value and since the residuals of the model are normally distributed, the most simple model with the highest adjusted \(R^2\) is chosen at least if the adjusted \(R^2\) approximates 0.80. This reasoning is also followed in the subsequent analysis.

For the imaginary part of the first pole (\(\Im(p_{1})\)), the real part of the fifth pole (\(\Re(p_{5})\)), the real (\(\Re(z_{3})\)) and the imaginary (\(\Im(z_{3})\)) part of the third zero and the imaginary part of the fifth zero (\(\Im(z_{5})\)), the Shapiro-Wilk test yields to a significant p-value (\(p<0.01\)) due to one – two in case of \(\Re(z_{3})\) – more extreme error term. In these cases, the following rule of thumb is applied. Since none of the simulated transfer functions or corresponding filters shows any manifest errors, the influence of the observation in question on the resulting regression is calculated. If the inclusion of the observation only influences the corresponding fitted value of this particular observation, this observation is kept. However, if the Cook’s distance of this observation exceeds 1 or if the df-beta exceeds \(\frac{2}{\sqrt{n}}\) with \(n\) the number of observations (37) the regression analysis is carried out without the observation in question (Kutner et al., 2004). The resulting models are also summarized in Table 4.1.

For the first and second zero, the regression analysis becomes much more complicated because most ears have real first and second zeros, but a distinct minority has complex first zeros. The complex zeros tend to be associated with longer ear canals although this relationship is not absolute. Additional FDTD simulations with higher values for \(l_{1}\) and the total range of values for the other parameters reveal that the combination of a longer ear canal with a shorter test bore and a deeper pit at the end of the earplug tends to influence the spectrum of the transfer function. Actually, those simulations show more often a (very) slight minimum in the lower frequency region whereas most transfer functions rise monotonically in these frequency range or remain constant. The extra minimum explains why the zeros of the corresponding filter differ from the majority. Unfortunately, the emergence of this minimum can not be absolutely predicted
Table 4.1: Summarized statistics for linear regression tabulating for each dependent variable ('dep') the predictive independent variables ('idep') with $l_1$ (in metre) the length of the residual part of the ear canal and $l_2$ (in metre) the length of the test bore; ‘inter’ stands for the intercept of the function. Further, the corresponding coefficients ('coefficients') are listed for each variable together with their p-value ‘p’. Finally, the adjusted $R^2$ is given (a$R^2$) and the case numbers of the observations that are not included in the model (‘rejected’).
based on the geometrical parameters, hence it is not feasible to make different regression models for both groups. Because of the nature of the problem, it is also very unlikely that filters with higher orders for the numerator will solve this problem.

Given the limited number of ears resulting in complex first zeros, only the real zeros are used for linear regression. Naturally, this implies that the regression model is less accurate for higher values of $l_1$. The results for the real part of the first ($\Re(z_1)$) and second ($\Re(z_2)$) zero are also reported in Table 4.1. The Durbin Watson statistic does not suggest autocorrelation for any of the tabulated models ($\alpha = 0.05$).

To sum up, inclusion of only the length of the residual part of the ear canal ($l_1$) and/or the length of the test bore ($l_2$) in the regression models appears sufficient to make reasonably accurate predictions. This clearly enhances the suitability of this approach in practice because these parameters can be measured easily, quickly and accurately for each individual.

### 4.2.3 Multivariate Orthonormal Vector Fitting

The theory and application described in this part are elaborated by Dr. Ir. D. Deschrijver.

Although the linear regression provides good results for most poles and zeros, there are several shortcomings. Firstly, the resulting equations are not entirely valid for the longest ear canals. Secondly, not all observations could be included in all models and thirdly, estimating the real and imaginary part of each pole and zero separately might increase the overall error. Therefore, an alternative approach is also used, namely the Multivariate Orthonormal Vector Fitting (MOVF) algorithm. Whereas multiple linear regression aims to fit the trajectories of each pole and zero separately, this approach computes an accurate multivariate model that describes the configuration of the poles and zeros as a whole. The overall goal of the MOVF algorithm is the same as for the linear regression; establishing a parameterized rational model that simplifies to an individualized frequency-dependent transfer function for certain values of the independent variables. The details of this approach are described by Deschrijver et al. (2008), a short outline of the modeling procedure is given here for convenience of the reader.

#### 4.2.3.1 Model Representation

Because the regression analysis has clearly shown that the length of the ear canal ($l_1$) and the length of the test bore ($l_2$) are the most influencing variables, the MOVF algorithm computes a rational trivariate model $R(s, l_1, l_2)$ that has the frequency variable $s$ (recall $s = j\omega$), but also $l_1$ and $l_2$ as parameters. It is defined as the ratio of a parameterized numerator $N(s, l_1, l_2)$ and denominator
4. Verification in situ: filter approaches

\[ D(s, l_1, l_2) = \frac{N(s, l_1, l_2)}{D(s, l_1, l_2)} = \sum_{p=0}^{P} \sum_{v_1=0}^{V_1} \sum_{v_2=0}^{V_2} c_{p,v_1,v_2}(\phi_p(s)\varphi_{v_1}(l_1)\varphi_{v_2}(l_2)) \]

The frequency-dependent basis functions \( \phi_p(s) \) are orthonormal rational functions that are based on a prescribed set of poles \( \vec{a} \). These poles \( \vec{a} \) are chosen as stable complex conjugate pairs with small negative real parts and imaginary parts linearly spaced over the frequency range of interest. The parameter-dependent basis functions \( \varphi_{v_1}(l_1) \) and \( \varphi_{v_2}(l_2) \) are also rational functions that are chosen in partial fraction form as a function of \( jl_1 \) and \( jl_2 \). They are based on a prescribed set of poles \( \vec{b}_1 \) and \( \vec{b}_2 \) respectively, which are chosen as complex pairs with small real parts of opposite sign and imaginary parts linearly spaced over the parameter ranges. A linear combination of two partial fractions is formed to ensure that they constitute a real function. The variables \( P, V_1 \) and \( V_2 \) denote the number of basis functions, and are chosen according to the dynamic behavior (i.e. the order) of each variable independently.

4.2.3.2 Calculation of model coefficients

Based on the FDTD simulations that are performed in Section 3.2.4, a dense set of data samples \( \{(s, l_1, l_2)_k, H(s, l_1, l_2)_k\}^K_{k=1} \) is obtained, taking into account the length of a particular test bore and ear canal. For the other geometrical parameters, the average values from the original dataset are chosen since the regression analysis has shown that in general their influence on the resulting transfer function is small. The goal of the MOVF algorithm is then to estimate the optimal values of the coefficients \( c_{p,v_1,v_2} \) and \( \tilde{c}_{p,v_1,v_2} \) of the trivariate transfer function in such a way that it approximates the data samples in a least-square sense. A linear approximation to this nonlinear optimization problem is obtained by using an iterative procedure called the Sanathanan-Koerner iteration (Sanathanan and Koerner, 1963). In the first iteration step \( (t = 0) \), Levi’s cost function is minimized to obtain an initial guess of the model coefficients. In successive iteration steps \( (t = 1, ..., T) \), updated values of the model coefficients are found by using the previously estimated denominator as an inverse weighting to the least-squares equations \( (D(0) = 1) \).

\[
\min_{c^{(t)}, \tilde{c}^{(t)}} \sum_{k=0}^{K} \left| \frac{N^{(t)}(s, l_1, l_2)_k}{D^{(t-1)}(s, l_1, l_2)_k} - \frac{D^{(t)}(s, l_1, l_2)_k}{D^{(t-1)}(s, l_1, l_2)_k} H(s, l_1, l_2)_k \right|^2 (4.21)
\]

This process is repeated in an iterative way until all the model coefficients have converged. To improve the convergence properties of the iteration, a relaxed non-triviality condition is applied. It is also noted that the trivariate model reduces to a regular, univariate transfer function for fixed values of \( l_1 \) and \( l_2 \).
4.3 Multiple linear regression and MOVF models

To compare the linear regression and MOVF directly, the expected transfer functions are calculated in MATLAB for each of the 37 ears in the original dataset. Worthwhile to mention is the fact that the actual computer time is acceptable for both approaches. The fitted transfer functions are then compared to the original simulations, because – although relying on two different fitting principles – they all aim to approach the FDTD transfer functions as accurately
In general, both models do well for ear canals of moderate length, an example is seen in Figure 4.5a. However, the MOVF approach clearly improves the prediction made by the linear regression model for more extreme lengths of the ear canal (see Figure 4.5b). This graphic also reveals that for this particular case, the prediction of the MOVF model is somewhat less accurate for the transfer function's minimum in the higher frequencies. The small deviation is not caused by an inaccurate fitting of the data, but it is connected with the fact that the FDTD simulations (which are used to compute the MOVF model) are based on average values for all geometrical parameters, except for the length of the test bore and ear canal. Although the influence of the other geometrical parameters is small, it is clear that they still might affect the exact amplitude of the transfer function. But since they are not included in the FDTD simulations for the MOVF, their influence cannot be included by this model. It is however noted that the resulting inaccuracy is not critical because previous research has shown that anyhow the exact transfer function's minimum is not exactly specified since it strongly depends on the position of the earplug in the ear canal (Bockstael et al., 2009).

To quantify the possible improvements between the fitted models, the difference is taken between the magnitude of the FDTD simulated transfer functions of the original data set (see Chapter 3) and the magnitude of respectively the transfer function fitted with linear regression and with MOVF, using the corresponding values for $l_1$ and $l_2$ as input. Then, this difference is squared and summed over four clearly distinguishable frequency ranges, i.e. between 100 Hz and 1500 Hz where most functions are either constant or monotonically rising; between 2500 Hz and 4000 Hz where a distinct maximum is seen; between 4500 Hz and 6500 Hz including a clear minimum and finally between 6500 Hz and
4.3. Multiple linear regression and MOVF models

Figure 4.5: Comparison between the simulated transfer function (‘FDTD’) and the transfer function fitted with linear regression (‘linreg’) or MOVF (‘movf’).

(a) Residual ear canal length: 0.0120 m.

(b) Residual ear canal length: 0.0203 m.
Verification in situ: filter approaches

8000 Hz where most function are again rising. Finally, the sums are divided by the number of frequency points to make the quadratic errors comparable across the different frequency ranges. The quadratic errors are compared between the models with a Wilcoxon signed-rank test instead of a paired student t-test because of present outliers and because the data tend to be right-skewed without enough observations to apply the central limit theorem.

For the lower frequency region, MOVF performs clearly better ($p < 0.001$), but the error is already quite small in this range. For the other frequency regions, no unambiguous statistical differences are found (see Table 4.2). Nevertheless, it is important to notice that the quadratic error always tends to be lower for the MOVF approach (see Figure 4.6) and that cases of total mismatch of the linear regression fit (see Figure 4.5b) are never seen for MOVF.

Finally, the quadratic error clearly increases with increasing frequency. This is not surprising given the difference in amplitude between the measured and the simulated transfer functions for frequencies above 5500 Hz, as reported in Chapter 3. Because both multiple linear regression and MOVF are based on the FDTD approach, it is natural that the discrepancy persists in the simplified models. As discussed in Section 3.4, this might be due to the boundary conditions included in the FDTD technique, but also to intrasubject variability, measurement difficulties and variation in the earplug’s exact position.

4.4 Discussion

Previous research has demonstrated that the MIRE method is a suitable way to measure the performance of hearing protectors in situ (Berger et al. 2007). Further, the sound pressure at the eardrum can be accurately predicted from the response of the MIRE measurement microphone provided that the transfer function is known. Especially in the frequency region below 6000 Hz, the measured transfer functions and the transfer functions predicted from the FDTD model are in good agreement (Bockstael et al. 2008).

Overall, the global shape of the obtained transfer functions is relatively simple and can be clearly traced down to the main features of test bore and ear canal. These findings might question the need for the quite complicated numerical

<table>
<thead>
<tr>
<th>Frequency range</th>
<th>Wilcoxon test</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.1 kHz - 1.5 kHz</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>2.5 kHz - 4.0 kHz</td>
<td>0.80</td>
</tr>
<tr>
<td>4.5 kHz - 6.5 kHz</td>
<td>0.19</td>
</tr>
<tr>
<td>6.5 kHz - 8.0 kHz</td>
<td>0.38</td>
</tr>
</tbody>
</table>

Table 4.2: The quadratic error between corresponding transfer functions estimated from multiple linear regression and from the FDTD models is statistically compared to the quadratic error between the estimates from the MOVF models and from FDTD models. The p-values are tabulated per frequency range for the Wilcoxon signed-rank test (Wilcoxon test).
Figure 4.6: Quadratic error for corresponding transfer functions estimated from multiple linear regression compared to the FDTD models (gray) and estimates from the MOVF models compared to FDTD models (white), calculated for the frequency region from 0.1 kHz to 1.5 kHz, from 2.5 kHz to 4.0 kHz, from 4.5 kHz to 6.5 kHz and from 6.5 kHz to 8.0 kHz. The boxes span the middle half of the ordered observations and the thick black lines inside represent the median. The whiskers extend to the most extreme data point which is no more than 1.5 times the interquartile range from the boxes. The circles represent data points that fall outside these limits.
FDTD simulations and the extensive regression and MOVF models, especially when it comes to practical implementation. Conversely, the sound pressure distribution in the occluded ear canal can be approximated with a one-dimensional analytical model. This approach remains relatively simple and in addition it allows to take into account the interaction between the resonances and the effects of non-rigid boundary conditions like the impedance at the eardrum. Despite these features, Section 4.2.1.2 clearly reveals that the basic analytical approach here presented would need further refinement to be useful in predicting the sound pressure at the eardrum from measurements by the MIRE microphone. Inclusion of viscothermal losses – for instance in the test bore – seem indispensable as is the extension to at least a two-dimensional model so that acoustically important features like the earplug’s pit can be included. In theory, these problems could to a certain extend be solved by more complex analytical calculations, but in this matter numerical techniques seem more efficient and straightforward.

Hence, the simulated transfer function are approximated with a filter approach. The advantage of this procedure is that the acoustical mechanisms included in the FDTD simulations still play their part because the filter characteristics are directly related to the simulated transfer functions. In addition, these characteristics can be linked to specific geometrical features of ear canal and earplug with multiple linear regression and MOVF. The thus found expressions only need the length of the ear canal and the test bore to predicted new transfer functions. The total length of the ear canal can easily be measured by sliding a silicone tube into the ear canal, for example at the time that the ear impression for the custom-made earplug is taken. The length of the earplug itself and of its inner bore can be determined during the manufacturing process.

The filters based on linear regression and MOVF perform both well with rather marginal statistical differences. However, it must be noted that the linear regression model is actually based on the original FDTD data set that is also used to calculate the quadratic error, whereas the MOVF starts from a dense set of new simulations (see Section 4.2.3.2). This might artificially lower the quadratic error for the linear regression model. Moreover, the MOVF performs clearly better for cases where a longer ear canal is combined with a shorter test bore.

4.5 Conclusion

This chapter reveals that an individual’s transfer function can be approximated with linear regression and MOVF if only the length of the test bore and of the residual part of the ear canal are known. Despite the fact that only two geometrical parameters are needed, the regression model is quite extensive and the MOVF approach even more. Nevertheless, once the model has been established, it takes negligible computer time to actually evaluate the individualized transfer functions. Simplifying the models might also reduce the accuracy and the two current approximations perform well for a considerable range of possible values.
4.5. Conclusion

for the independent parameters. However, in some cases the outcome of the MOVF model is clearly better, suggesting that this model is possibly preferable in practice.
4. Verification in situ: filter approaches
Chapter 5

Speech recognition in noise with active and passive hearing protectors: a comparative study

The content of this chapter is based on \cite{Bockstael2021}.

5.1 Introduction

When using personal hearing protectors to prevent noise-induced hearing loss, the actual sound pressure level at the workfloor remains unaltered, making consistent use crucial for effective prevention \cite{Winters2005}. However, the fright of missing verbal cues and warning signals not seldom paves the way to incorrect partial insertion or even temporal removal of the protector \cite{Hong2008, Okpala2007, Lwow2007, Nakashima2007}. This implies that the implementation of personal protection is only worthwhile if the protectors sufficiently preserve environmental awareness and – not forgetting – if this preservation is also perceived by the users.

When making attempts to handle this issue, one must always bear in mind that signal detection and understanding with hearing protectors depend on a complex of factors. First, influencing characteristics with regard to the listener include his gender \cite{Giguere2008}, his hearing abilities and his protector’s attenuation \cite{CasaliBerger1996}. Naturally, general communication skills and context as well as complexity of the message also play an important role \cite{CasaliBerger1996}. In addition, speech intelligibility with hearing protectors is influenced by whether or not the speaker wears hearing protectors \cite{TuftsFrank2003, Hoermann1984} and is exposed to noise \cite{Forshaw2007}. 
Speech recognition is further enhanced if listener and speaker stand relatively close (Giguère et al., 2008) and can actually see each other's facial expression and lip movements (Casali and Berger, 1996). Finally, the acoustical characteristics of both signal and noise (Studebaker et al., 1994; Laroche et al., 2003) (see Section 5.2.2.1 for more details) and reverberation time of the environment (Casali and Berger, 1996) also determine to a certain extent the ease of communication.

Given the large number of variables, it is not surprising that different studies yield to sometimes contradictory conclusions with respect to the influence of hearing protectors on speech intelligibility in noise. Moreover, the wide variety in types of hearing protectors (see also Section 1.3) increases the variability among research results. In this regard one can roughly distinguish on the one hand the ‘standard’ or ‘classical’ protectors that solely block the sound path to the eardrum whereas on the other ‘augmented’ protectors actually process the incoming sound with or without electronics, i.e. in an active or passive way.

For normal-hearing subjects, Abel et al. (1993) found that wearing classical hearing protectors is beneficial for speech perception in noise. However, studies designed to simulate on-the-job listening conditions have reported poorer audibility with standard earplugs (Casali et al., 2004). With respect to hearing impaired subjects, Abel et al. (1982, 1993) demonstrate that wearing earmuffs or earplugs deteriorates word recognition in noise when a high-frequency or flat sensorineural hearing loss is present. By contrast, Dolan and O’Loughlin (2005) suggest that occluding the ears does not necessarily hamper communication, even among people with hearing loss.

Whether or not standard protectors (slightly) alleviate speech intelligibility, the fact remains that communication in background noise is often though going (Nakashima et al., 2007). In addition, listeners with protectors might less benefit from relatively silent periods in fluctuating noise because of momentaneous overprotection (Robinson, 2003). As a consequence, ‘augmented’ hearing protectors have been developed (Casali et al., 2004; Casali and Berger, 1996), offering different possibilities to make sound attenuation more comfortable (Dantscher, 2007) and to diminish the masking effect of noise on signals.

Nowadays, a wide variety of augmented protectors is available, each with their own operating mechanism. To the authors’ knowledge, only limited research has been published concerning the effect of these protectors on speech intelligibility (Muchenje, 2008) and these studies usually fail to establish clearly the benefits on signal perception in noise (Dolan and O’Loughlin, 2005; Abel et al., 1993; Abel and Spencer, 1997; Dancer et al., 1999; Casali et al., 2004; Muchenje, 2008) over all standard passive protectors.

This chapter describes tests with prototypes of active hearing protector designed to be worn in similar conditions as standard passive protectors. Different types of background noise are selected and in each sound environment speech fragments are recorded using a HATS without any hearing protectors, with passive earplugs and with active ones. The speech intelligibility for each sound fragment is then determined from the intelligibility scores of normal-hearing sub-
5.2 Material and methods

5.2.1 Hearing protectors

Three types of hearing protectors are initially included: passive custom-made acrylic hearing protectors (PC), active custom-made acrylic hearing protectors with volume control and active foam earplugs (AF). For the hearing protectors with volume control, two settings are selected: full (AC 3) and minimal amplification (AC 1). These settings are chosen because they provide clearly different output levels for the input levels used in this study.

The passive earplugs with ST35 filter simply block the ear canal and thus reduce the sound with a fixed amount, regardless of the input level. The attenuation values measured by the German Institute for Occupational Safety and Health (BGIA) following ISO 4869-2 are tabulated in Table 5.1.

By contrast the active protectors contain a microphone and a loudspeaker so that the attenuation can be adapted to the incoming sound level. The two types of augmented protectors process the signals in different ways; the foam earplugs amplify the incoming sound with a fixed quantity whereas the custom-made ones have a brickwall limiter that imposes a hard ‘ceiling’ on the loudspeaker output – the signal can go no further once it reaches the safety threshold and this regardless of the position of the volume control.

5.2.2 Sound environment

This study aims to assess speech recognition in noise for different listening conditions. Therefore, realistic noise fragments are recorded using a Bruel & Kjær head-and-torso-simulator (HATS) type 4128 C with left and right ear simulator. Additionally, the same HATS is used to record the speech material in

<table>
<thead>
<tr>
<th>Frequency (Hz)</th>
<th>APV (dB)</th>
</tr>
</thead>
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<td>125</td>
<td>14.4</td>
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<td>17.6</td>
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<td>1000</td>
<td>21.9</td>
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<tr>
<td>2000</td>
<td>27.6</td>
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<td>4000</td>
<td>26.3</td>
</tr>
<tr>
<td>8000</td>
<td>24.0</td>
</tr>
</tbody>
</table>

Table 5.1: Assumed protection values (APV) in function of the octave bands center frequencies for the custom-made passive earplugs with ST35 filter measured by the German Institute for Occupational Safety and Health (BGIA).
an anechoic room. The rationales behind this approach and the characteristics of both speech and noise are discussed in this section.

5.2.2.1 Noise material

The influence of noise on speech largely depends on three major components. First, the signal-to-noise ratio plays a very important role. The louder the noise compared to the speech signal, the more speech recognition is hampered. Secondly, the frequency spectrum of the noise also influences speech intelligibility, in particular for people with normal hearing (Festen and Plomp [1990]). In this regard [Studebaker et al. (1994)] have shown that high-frequency noise decreases the intelligibility more than white noise. Thirdly, the temporal characteristics of the noise have to be taken into account since relatively silent periods enhance speech intelligibility for normal hearing subjects (Laroche et al. [2003]).

In accordance with this knowledge, three types of noise are chosen for this study. First recordings are made inside the alternators and turbines hangar of a power station where the noise incidence is more or less equally diffused in space. In contrast to this, a more focused sound source is included in the form of a bottle filling machine. Finally, moving fork-lift trucks are selected because they produce more fluctuating noise. All recordings are made with the HATS facing the sound source.

From the time averaged (Leq over 80 s) spectra depicted in Figure 5.1 it can be seen that the alternators and turbines produce more low-frequency energy whereas the bottle filling machine has more energy in the frequency region between 1000 Hz and 10000 Hz. The fork-lift trucks produce an overall lower sound pressure level. Recordings appear to be very similar for the left and the right ear simulator.

Naturally, speech intelligibility in noise is not only determined by the acoustical characteristics of the signals but also by the way they are listened to. In normal conditions, the nervous system can combine information from the two ears. In a background of maskers or detractors, this binaural processing can enhance the detectability of signals and improve the intelligibility of speech (Hafter and Trahiotis [1997]) especially if speech and noise come from different directions (Hawley et al. [1999] Giguère et al. [2008]). To include the possible effect of spatial separation between signal and noise, the noise of the bottle filling machine is also recorded with the left ear pointing at the sound source. This particular machine is chosen because its focused character produces the largest interaural difference if the HATS is no longer frontally facing the source. Figure 5.2 reveals that the sound pressure levels at the left and the right simulator differ most distinctly between 2000 Hz and 5000 Hz.

5.2.2.2 Speech material

For the speech material standardized recordings of the Dutch-language ‘Brugse Lijst’ are chosen. This material – read by a professional female speaker – is especially designed to perform speech audiometry and consists of consonant-
5.2. Material and methods

Figure 5.1: $\frac{1}{3}$-octave band spectra of the recordings with the HATS facing the sound source; ‘Electricity’ refers to the noise of alternators and turbines, ‘Line 3’ to the bottle filling machine and ‘Fork-lift trucks’ to moving fork-lift trucks. ‘L’ are the recordings made by the left ear simulator while ‘R’ refers to the right ear simulator.

Figure 5.2: $\frac{1}{3}$-octave band spectra of the bottle filling machine (‘Line 3’) recorded with the left ear simulator orientated toward the sound source. ‘L’ are the recordings made by the left ear simulator while ‘R’ refers to the right ear simulator.
vocal-consonant words spread among 20 lists (numbered from 31 to 50) with 17 words per list. Care is taken that each list is equally difficult, hence the specific list chosen for each test situation should not influence the result (Damman, 1994). The advantage of this type of words is their very low redundancy and the fact that the recognition of individual monosyllables in noise appears to be hardly improved by repetition (Sust et al., 2009).

The standardized recordings of the ‘Brugse Lijst’ are not suitable to be directly mixed with the noise fragments because the former do not include the head-related transfer functions whereas the latter do. Thus, the ‘Brugse Lijst’ is played with the audio equipment described in Section 5.2.3.2 and recordings are made in an anechoic room with a Renkus-Heinz (model CM 81) loudspeaker placed at 1 m from the HATS. All electronics are placed outside the room to prevent disturbance from reflection and equipment noise.

For the first 16 lists, the HATS faces the loudspeaker so that the right and the left ear simulator receive the same signal. For the last 4 lists, the HATS is turned with its right ear toward the sound source.

To monitor the test setup, recordings are also made for the speech reference noise provided by the developers of the Brugse lijst. The resulting spectra are depicted in Figure 5.3.

5.2.3 Listening conditions

5.2.3.1 General considerations

The recorded speech and noise fragments (Section 5.2.2) can be combined into five different sound environments (Section 5.2.3.4). One of the aims of this project is the comparison of speech intelligibility in different listening conditions – i.e. under hearing protectors and with open ear – for these environments. For that reason test material is built by first recording speech and noise separately with unoccluded ear simulators (see Section 5.2.2), subsequently mixing speech and noise electronically to create several speech-in-noise fragments (see Section 5.2.3.4) and finally recording the fragments with the HATS under headphone in the required listening conditions (with and without hearing protectors). This approach requires specific processing of the speech and noise samples to keep the fragments as realistic as possible (see Section 5.2.3.3 and 5.2.3.5).

The different recording steps are necessary because the active hearing protectors are level-dependent, which implies that the amplification is based on the global sound pressure level. Since the speech is noticeably softer than the noise, the processing of the speech will be clearly different if the words are presented alone or in combination with noise. Because speech and noise are inherently mixed at the workfloor, the test fragments must mimic this situation to get a realistic idea of the active protectors’ performance. This explains why for the final recordings a mixture of speech and noise is preferred to the separate speech and noise fragments. An appurtenant benefit is that the sound environments are essentially identical for the different listening conditions. The composition of the test material is described in more detail in the following sections.
5.2. Material and methods

5.2.3.2 Audio equipment

All recordings and the presentation of the listening material to the test subjects is carried out with the same audio equipment. The fragments are played on a laptop PC using Audacity software and then the signal is sent to an open circumaural Philips headphone (type SBCHP890) via a Pioneer A-607 R direct energy MOS amplifier.

5.2.3.3 Pre-processing

The project aims to test speech intelligibility in sound environments that are as realistic as possible, thus different issues have to be considered in the processing of the material. First, unwanted influence of the HATS is filtered out of the sound material. This is necessary because the composing of the test material includes two recordings with the HATS, once with open ears at the workfloor (for the noise) and in the anechoic room (for speech) and once under headphone for the different listening conditions. The first recordings are deliberately made with the HATS to have realistic binaural recordings, but this implies that the HATS’s ear canal and pinna will influence the final test material twice which is of course undesirable. Moreover, the headphone used for the final recordings might also unwantedly mark the test fragments. Hence, the influence of the ear simulators and headphone is minimized by filtering noise and speech so that recordings under headphone made by the HATS with open ears resemble the original recordings as close as possible, especially between 500 Hz and 4000 Hz.
i.e. the frequency range most important for speech recognition. These filtered fragments, and not the original recordings, are used to create the speech-in-noise listening items.

Further, when mixing the sound files, the level of the speech is set at approximately 74 dB measured at the HATS’s eardrum, comparable to 68 dB(A) measured in free field [Hammershøi and Møller 2008]. This level is chosen to approach a normal communication situation where a female person would speak at free-field levels between 63 dB(A) (raised) and 71 dB(A) (loud) [Olsen 1998]. The calibration of the speech signals is done with continuous speech noise, especially developed for this particular set of speech material [Damman 1994]. Once the sound equipment is calibrated, the intensity of the filtered noise files is also checked. The loudest noise levels are slightly attenuated to protect the participant’s hearing. This is done by simply decreasing the overall level, preserving the spectral characteristics of the noise. Applying the 3-dB rule to the lowest safe sound level of 74 dB(A) [Mills and Going 1982] for eight-hour exposure, people can be exposed safely to a level of 94 dB(A) during 3 minutes 38 seconds. Since there are only five recordings with open ears and each recording lasts about 1 minute 15 seconds, it is possible to respect this safe exposure level with the sound pressure levels for open ear recordings tabulated in Table 5.2. In this table, the A-weighting is applied to allow comparison with the safety limits. Nevertheless, this type of weighting is less applicable for measurements at the eardrum since the A-weighting also accounts for the influence of the outer and middle ear.

<table>
<thead>
<tr>
<th>Noise Source</th>
<th>Left</th>
<th>Right</th>
</tr>
</thead>
<tbody>
<tr>
<td>Electricity</td>
<td>91.9</td>
<td>91.2</td>
</tr>
<tr>
<td>Line 3 f</td>
<td>92.7</td>
<td>93.0</td>
</tr>
<tr>
<td>Line 3 l</td>
<td>94.5</td>
<td>92.6</td>
</tr>
<tr>
<td>Fork-lift trucks</td>
<td>85.6</td>
<td>87.9</td>
</tr>
</tbody>
</table>

Table 5.2: Overall A-weighted sound pressure level (dB(A)) under headphone for the noise fragments to be mixed with speech measured with the left (‘Left’) and right (‘Right’) ear simulator of the HATS; ‘Electricity’ refers to the noise of alternators and turbines, ‘Line 3 f’ to the bottle filling machine with the HATS facing the sound source, ‘Line 3 l’ to the same bottle filling machine with the left ear simulator orientated toward the sound source and ‘Fork-lift trucks’ to moving fork-lift trucks.
5.2.3.4 Recording

For the recordings under headphone, the filtered fragments of both speech and noise are set at an appropriate level (see Section 5.2.3.3) and mixed using Audacity software. The first 16 lists of the ‘Brugse Lijst’ are combined with the noise from the alternators and turbines, from the fork-lift trucks and from the bottle filling machine recorded both frontally and sideways. The latter noise fragment is also combined with the last four speech lists, meaning that in these conditions noise predominantly comes from the left and speech from the right. In this way, five global sound environments are created. Within each sound environment one particular noise fragment is combined with four successive speech lists to establish in total 20 unique speech-in-noise fragments.

Subsequently, for each of the five sound environments, recordings with the HATS are made under headphone in five listening conditions; with open ears and with the ear simulators occluded by the earplugs described in Section 5.2.1, namely the passive earplugs, the active custom-made earplugs in both volume settings and the active foam earplugs. Since there are only four unique fragments per sound environment, each time one speech list has to be used twice.

Before each series of recordings, it is ensured that the unoccluded sound pressure level of the speech reference noise under headphone equals 74 dB at the left and right ear simulator. For both passive and active hearing protectors, it is also verified that the right and left ear simulator measure the same overall intensity level when the reference speech signal of 74 dB is presented as input signal.

5.2.3.5 Post-processing

It is well-known that measurements with a HATS tend to overestimate the attenuation of passive hearing protectors (Berger, 2005). To compensate for this effect, the approach proposed by Hiselius (2005) is followed, taking into account bone conduction thresholds. Since an absolute prediction of the earplugs’ attenuation is not within the scope of this work, this procedure seems allowed to make the test setup more realistic.

A summary of all test situations with the sound pressure level at the HATS ear simulators can be found in Table 5.3. Again, A-weighting is applied to allow comparison with the safety limits. This table clearly shows that the foam active hearing protector does not perform well since the sound pressure level is too strongly amplified, even beyond the unoccluded level. Therefore the fragments of the active custom-made hearing protector with minimal amplification are used instead as test material.

5.2.4 Speech intelligibility test

5.2.4.1 Test subjects

The project aims to compare speech intelligibility for different controlled listening situations and not to predict speech recognition in real wearing conditions. Thus, it is important that the general communication skills of the listeners
<table>
<thead>
<tr>
<th>List</th>
<th>Noise</th>
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<th>L</th>
<th>R</th>
</tr>
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<tr>
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<td>96.1</td>
</tr>
<tr>
<td>46</td>
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<td>78.5</td>
<td>79.2</td>
</tr>
<tr>
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<td>92.0</td>
</tr>
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<td>56.0</td>
<td>56.2</td>
</tr>
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<td>83.3</td>
</tr>
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<td>50 r</td>
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<td>AF</td>
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<td>96.0</td>
</tr>
<tr>
<td>50 r</td>
<td></td>
<td>AC 1</td>
<td>78.2</td>
<td>78.9</td>
</tr>
</tbody>
</table>

Table 5.3: Overall A-weighted sound pressure level (dB(A)) for the 20 different listening situations. ‘List’ stands for the track number of the different speech lists as provided by the developers of the speech material. The indicator ‘r’ is added when the list is recorded with the right ear toward the loudspeaker in contrast with the other frontal recordings. ‘Noise’ is the type of background noise and ‘LC’ are the five different listening conditions; both with abbreviations similar to the previous tables.
do not influence the outcome of tests and therefore only native Dutch-speakers who have at least successfully finished high school are included. All participated voluntarily and signed an informed consent.

Further, normal hearing is required with tonal hearing thresholds of 25 dB or better for all octave frequencies between 250 Hz and 8000 Hz [Roeser et al., 2000], normal tympanometric results and normal speech perception. Pure-tone audiometry is carried out by a qualified audiologist in accordance with the modified Hughson-Westlake technique [ISO 8253-1]. This takes place in a sound-proof audiometric cabin at the Audiology Center of the Department of Otorhino-laryngology of Ghent University Hospital, using a regularly calibrated Orbiter 922 audiometer. Speech perception is tested monaurally in silence using the standardized Dutch consonant-vocal-consonant NVA-list [Damman, 1993]. Furthermore, the status of the middle ear is verified by carrying out tympanometry at 226 Hz with a ZODIAC 901 tympanometer of Madsen Electronics. All ears show normal patterns, suggesting that the reflection of sound at the eardrum is not perceptibly influenced by abnormalities of the eardrum or middle ear [Fowler and Shanks, 2002].

These criteria yield to a final group of 60 test subjects (30 female and 30 male) who are on average 27.6 years old. According to Abel et al. (1993) age is an insignificant factor with respect to consonant discrimination and word recognition, at least for the working population under study (George et al., 2006; Gifford and Bacon, 2005; Gordon-Salant and Fitzgibbons, 2004). Further, no gender related significant differences in hearing level or in speech recognition are present in the test group.

5.2.4.2 Test setup

At the beginning of each test day, the audio equipment is calibrated so that the noise level of the speech reference noise under headphone is 74 dB at both HATS’s ear simulators.

The instruction to the test subject are in conformity with classical speech audiometry [Damman, 1993]. The subject is told that he will hear fragments of speech in noise of approximately 1 minute 15 seconds. Although the noise levels are not harmful because of their short duration, it is stressed that the test can be stopped at any time if the subject feels that he cannot stand the noise. Further, the subject is encouraged to repeat as much speech material as possible, even though he may have to guess or can only repeat one or two phonemes instead of a complete word. The investigator notes the correct phonemes on a score form and afterward the number of correctly understood phonemes is counted for each list, yielding to a specific speech recognition score. There is always a short break between the different listening fragments used to reinforce the subject and solicit to take a longer break if necessary. In this way, the concentration level is kept up during the entire test.

The 20 speech-in-noise fragments are presented through the headphones in random order. A random permutation of the lists numbers is carried out to determine the sequence of fragments for each subject. In the beginning, a test list is
played so that the participants become used to the whole test concept.

5.3 Results

In this section, the speech recognition scores will be compared between test fragments (Section 5.3.1) and with respect to those results, the acoustical characteristics of sound environments and listening conditions will be assessed in detail (Section 5.3.2).

5.3.1 Comparison of speech recognition scores

The mean speech recognition scores and standard deviations are depicted in Figure 5.4 for the 20 test fragments. On this data a mixed model analysis of variance (ANOVA) is carried out with three crossed factors, using the statistical software SPSS. For this analysis, the variables ‘sound environment’ and ‘listening condition’ are included as fixed factors whereas the variable ‘subject’ is included as a random factor. The full linear model for this design is

\[
y_{ijkl} = \mu + \alpha_i + \beta_j + \gamma_k + (\alpha\beta)_{ij} + (\alpha\gamma)_{ik} + (\beta\gamma)_{jk} + (\alpha\beta\gamma)_{ijk} + \epsilon_{ijkl} \tag{5.1}
\]

with

\[y_{ijkl}: \text{ the observation within each group,}\]
\[\mu: \text{ the group mean,}\]
\[\alpha_i: \text{ the fixed main effect of the variable ‘sound environment’,}\]
\[\beta_j: \text{ the fixed main effect of the variable ‘listening condition’,}\]
\[\gamma_k: \text{ the random main effect of the variable ‘subject’,}\]
\[(\alpha\beta)_{ij}: \text{ the fixed interaction effect of ‘sound environment’ by ‘listening condition’,}\]
\[(\alpha\gamma)_{ik}: \text{ the random interaction effect of ‘sound environment’ by ‘subject’,}\]
\[(\beta\gamma)_{jk}: \text{ the random interaction effect of ‘listening condition’ by ‘subject’,}\]
\[(\alpha\beta\gamma)_{ijk}: \text{ the random interaction effect of the three factors,}\]
\[\epsilon_{ijkl}: \text{ the random error.}\]

The first two variables are considered fixed because the effect of their specific levels is of interest. In this regards, ‘sound environment’ has five levels since the combination of speech and noise yields to five different sound environments (see Section 5.2.3.4). The factor ‘listening condition’ has four levels; open ear, passive earplug and active custom-made earplug at two volume settings (see Section 5.2.3.4). The factor ‘subject’ on the other hand is a random factor because its levels are a sample from a larger population of potential factor
Figure 5.4: Bar plot for the speech recognition scores of the 20 test fragments represented by the track number of the different speech lists. The center of the error bars is given by the mean speech recognition score whereas the width of the bars equals one standard deviation. All abbreviations are similar to the other figures and tables except for 'Line 3 l r' where 'r' indicates that the speech is recorded with the right ear simulator oriented toward the loudspeaker.
levels. Thus inferences are desired about the population of factor levels, not about the individual factor levels themselves (Kutner et al., 2004). Since there is only one observation per group, the three-way interaction effect is assumed insignificant based on visual inspection of the error bar chart (Kutner et al., 2004) and the ANOVA outcome is tabulated for the two-way interaction effects in Table 5.4. The main effects can not be interpreted because all two-way interaction effects are significant ($\alpha = 0.05$) (Kutner et al., 2004) and hence they are not included in this table.

Before this model is interpreted, the aptness of the assumption that the residuals come from a normal distribution is checked (Kutner et al., 2004). Both the Kolmogorov-Smirnov test and the Shapiro-Wilk normality test yield to insignificant p-values ($p > 0.01$) and the residuals turn out to be randomly distributed around zero, showing that the mixed model approach is valid for this dataset. Apparently, the influence of sound environment and listening condition does not apply to the same degree to all levels in the population of the random factor ‘subject’ because both random interaction effects are to be included in the model. Since the factor levels of these interactions effects constitute a sample from a larger population of factor levels, they are not of intrinsic interest in themselves (Kutner et al., 2004). Conversely, the levels of the fixed interaction effect between the variables ‘sound environment’ and ‘listening condition’ are important for further analysis and therefore a pairwise Tuckey-post hoc test is carried out with the interaction effect as independent variable (Hothorn et al., 2009). The most striking results are discussed below and the significant differences are summarized in Tables 5.5 and 5.6 and represented graphically in Figures 5.5 and 5.6.

5.3.1.1 Listening conditions in sound environments

In Table 5.5 and Figure 5.5 the statistically significant results are summarized in function of the sound environment. It becomes clear that the performance of the different hearing protectors strongly depends on the type of background noise. First, passive protectors seem to give the best results for the noise at the power plant and the noise from the bottle filling machine recorded from the left with speech coming from the right. These results are in agreement with previous studies suggesting that passive protectors might enhance communication for normal hearing subjects when the noise levels exceed 85 dB(A) (Abel et al., 2004).

<table>
<thead>
<tr>
<th>Source</th>
<th>$F$</th>
<th>$p$</th>
</tr>
</thead>
<tbody>
<tr>
<td>$(\alpha\beta)_{ij}$</td>
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<td>.000</td>
</tr>
<tr>
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<td>1.215</td>
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</tr>
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<td>$(\beta\gamma)_{jk}$</td>
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<td>.009</td>
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</tbody>
</table>

Table 5.4: Significant interaction effects for the speech recognitions scores analyzed with a mixed model analysis of variance (ANOVA). The column $p$ stands for the p-value associated with the calculated F-statistic.
5.3. Results

(a) Line 3 f.

<table>
<thead>
<tr>
<th></th>
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<th>AC 1</th>
</tr>
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<tbody>
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<td>(&gt;0.05)</td>
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(b) Fork-lift trucks.

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<th>AC 1</th>
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**Table 5.5**: Summary of the pairwise comparisons tabulated in function of the sound environments. Each cell represent the difference in speech recognition scores between the listening condition in the column and the corresponding condition in the row. The positive estimated differences are tabulated with between brackets their associated p-value. (To be continued on the next page.)
(c) Electricity.

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(d) Line 3 l.

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(e) Line 3 l r.

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Table 5.5: Post-hoc analysis in function of the sound environments: continuation.
### 5.3. Results

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(b) PC.

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**Table 5.6:** Summary of the pairwise comparisons of speech recognition scores tabulated in function of the listening conditions. All abbreviations are similar to the other tables except for ‘Fork-lift’ referring to the noise from the fork-lift trucks and ‘Electric’ to the noise from the power plant. (To be continued on the next page.)
5. Speech recognition and hearing protectors

(c) AC 3.

<table>
<thead>
<tr>
<th>Line 3 f</th>
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(d) AC 1.

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</table>

Table 5.6: Post-hoc analysis in function of the listening condition: continuation.
Figure 5.5: Graphical representation of the statistical pairwise comparison between listening conditions grouped in function of the sound environment. Two listening conditions with a significant ($\alpha = 0.05$) different speech recognition score are connected with a line and the listening condition with the higher score is placed above the other. A dashed line is used if the p-value lies between 0.05 and 0.1. All abbreviations are similar to previous tables and figures.
Figure 5.6: Graphical representation of the statistical pairwise comparison between sound environments grouped in function of the listening condition. Two sound environments with a statistical significant different speech recognition score are connected with a line and the listening condition with the higher score is placed above the other. All abbreviations are similar to previous tables and figures.
5.3. Results

For the noise of the fork-lift trucks, the passive earplugs appear to hamper speech recognition more than the unoccluded situation. In this environment, the noise level is quite soft compared to the other conditions and therefore it is not unthinkable that the attenuation of the passive earplug is overprotective, involving more difficulties in communication.

The performance of the active protectors clearly depends on the settings of the volume control. Whereas maximal gain seems to lead to the least performing listening condition for most sound environments, minimal gain enhances recognition for the frontal recordings of the bottle filling machine and is the best occluded listening condition for the fork-lift trucks.

5.3.1.2 Sound environments in listening conditions

In Table 5.6 and Figure 5.6, the statistically significant results are summarized in function of the listening conditions. From these analysis, two conclusions can be drawn directly: the bottle filling machine frontally recorded is clearly the most difficult listening situation whereas the fork-lift trucks appear to be the least disturbing for the unoccluded situation and the active protector with minimal gain. For the other sound environments, the ranking depends on the listening condition.

Further, spatial segregation between speech and noise seems to enhance speech recognition for all listening conditions if the sound environments with frontally recorded noise are compared to the sideways recordings. Changing the direction of the speech from frontal to sideways also induces a positive effect on speech perception for all listening conditions except the active protector with minimal gain.

In general, hearing target signals in noise is indeed much easier if speech and noise come from different directions (Bronkhorst and Brungart, 2005). This phenomena will be elaborated more in Section 5.3.2.6, as will the absence of this advantage for the active earplug with minimal gain in Section 5.4. With this, it should be noted that the so-called ceiling effect has to be considered first when investigating the lack of amelioration of phoneme recognition. The effect comprehends that no improvement can be seen because speech recognition has reached a maximum level for a given sound environment. However, the ceiling effect can already be ruled out given that in both sound environments with spatial segregation between speech and noise, the speech recognition scores of the active protector with minimal gain are exceeded by those of the unoccluded situation.

5.3.2 Acoustical analysis

5.3.2.1 Global sound pressure level

Assessing the global sound pressure level with the HATS is the first step to ensure that a hearing protector can effectively prevent noise-induced hearing loss. In this, the malfunction of the foam active earplug can be seen from Table 5.3.
Further, the post-hoc analysis reveals that the noise from the fork-lift trucks is the least disturbing (see Section 5.3.1.2). Little surprisingly, Table 5.2 clearly shows that this noise fragment has the lowest overall sound pressure level. Because of the fixed speech level in all sound environments, a lower noise level leads automatically to a better signal-to-noise ratio and hence more favorable listening conditions. Table 5.2 also suggests that the bottle filling machine yields to the lowest signal-to-noise ratio for the frontally recorded noise fragments, although the difference with the noise from the power plant is quite small (see also Section 5.3.2.3).

In a second stage, the influence of the overall level at the eardrum (over the frequency range of interest, i.e. between 100 Hz and 10000 Hz) on speech recognition scores is addressed in Figure 5.7. At first glance, little connection is seen between both variables. This is not surprising within this relatively small range of sound levels because speech recognition depends on a complex of acoustical parameters (see below) and the global sound pressure level is only one of them. However, things are more clear if the particular sound environment is taken into account. For the sound of the fork-lift trucks where the unoccluded situation has a relatively lower overall level with higher recognition scores, the passive protectors deteriorate speech intelligibility although the results remain quite good compared to the other sound environments. Nevertheless, this is a highly unwanted situation because people are tempted to remove their earplugs if this enhances communication. If the sound pressure level is remarkably higher in the unoccluded ear canal, chances increase that implementation of passive protectors equals or even ameliorates speech intelligibility.

For the active protectors, the link between sound pressure level and speech recognition is much less clear, suggesting that other mechanisms like for instance distortion (see Section 5.3.2.4) govern the outcome. However, the active protectors with minimal gain seem to do a good job for the noise from the fork-lift trucks, possibly because they keep the sound pressure level higher than their passive counterparts.

### 5.3.2.2 Time pattern

Figure 5.8 depicts the one-second Leq-values of the test fragments with unoccluded ear simulator. From this graphic it becomes clear that the noise of the fork-lift trucks is not only generally lower, but also more fluctuating than the other samples. As stated previously, short periods with lower background noise might indeed improve speech intelligibility for normal-hearing subjects.

In this regard, it is also notable that the active hearing protector with maximal amplification does not preserve the fluctuating character of the noise (see Figure 5.9). Artificially equalizing the noise levels naturally cancels the benefits from silent gaps and hence might explain the low intelligibility scores for this test fragment.
5.3. Results

Figure 5.7: Line plot for the speech recognition scores of the 20 test fragments in function of the global sound pressure level over the frequency range between 100 Hz and 10000 Hz. All abbreviations are similar to the other figures and tables.
Figure 5.8: Overall sound pressure level per second (Leq per 1 s) for the unoccluded test fragments registered with the right ear simulator. All abbreviations are similar to previous tables and figures.

Figure 5.9: Overall sound pressure level per second (Leq per 1 s) for the noise of the fork-lift trucks. All abbreviations are similar to the other figures.
5.3. Results

5.3.2.3 Spectral analysis

Apart from the global sound pressure level and the time pattern, the spectra of the test fragments also influence speech intelligibility. For instance, the difference between the global sound level at the bottle filling machine and at the power plant is quite small, but the speech recognition scores clearly differ. Figure 5.1 shows indeed that the bottle filling machine produces higher levels above 1000 Hz. The fact that this relation is reversed around 50 Hz is less important since high-frequency noise appears to be more disturbing ([Studebaker et al., 1994]).

To illustrate the effect of the listening conditions on the sound, the spectra of the four fragments with the fork-lift truck noise are depicted in Figure 5.10. One of the most striking features is the interaural difference for low-frequency attenuation of the active protector. This is most prominent for the situation with minimal gain and will be discussed in Section 5.3.2.6. Further, the active protector with maximal amplification seems to emphasize strongly the frequencies between 1000 Hz and 4000 Hz. This region is indeed important for speech perception, but an excessive amplification might distort the balance between the different frequencies and hence hamper speech intelligibility instead of improving it. The possibility of distortion will be elaborated in Section 5.3.2.4.

5.3.2.4 Distortion

To quantify possible distortion, the amplitude of speech-in-noise fragments is investigated more closely between 1000 Hz and 4000 Hz. In the concrete, the magnitude in the unoccluded condition is subtracted from each occluded condition within the same sound environment and this for each $\frac{1}{3}$-octave band in the frequency region of interest. The spread of the differences is assessed by calculating the coefficient of variance $cv$ as

$$
    cv = \frac{s}{\bar{y}} \cdot 100\%
$$

with $s$ the standard deviation of the observations and $\bar{y}$ the mean.

If the hearing protectors respect the relation between the different frequency bands seen in the original signal, the amplification or attenuation should be more or less constant across this frequency range and therefore the coefficient of variation should be quite low. This is the case for the passive protectors and the active protectors with minimal amplification; over the different sound environments $cv$ is maximal 16 %. By contrast, for the active protectors with maximal amplification, $cv$ is minimal 20 % and even reaches 44 % for the noise of the fork-lift trucks. These results suggest that the low intelligibility scores for maximal amplification might be associated with distortion of the original spectrum.
5.3.2.5 **Loudness**

A key element when studying the influence of noise on speech is the amount of energetic masking. This is most commonly assessed by comparing the A-weighted sound pressure level of speech to that of noise yielding to a certain signal-to-noise ratio. Yet this approach is less useful for the hearing protectors under study because they apply the same amount of attenuation or amplification to both speech and noise. Hence they can not distinctly alter the ratio of sound pressure levels and thus differences in signal-to-noise ratio are not suitable to explain differences in speech recognition between listening conditions.

By contrast, the sensation of sounds depends on much more variables than the sole sound pressure level. For instance, the beneficial effect of passive protectors on speech recognition in noise might be explained with the generally accepted theory that by lowering the total incident energy of both speech and noise, passive protectors alleviate cochlear distortion that occurs at high sound levels (Casali and Berger, 1996). This effect and other important variables like bandwidth, frequency content and duration of sounds can be included by working with ‘loudness’ (in sone) instead of the A-weighted sound pressure level (Zwicker and Fastl, 1999; Fastl, 1997).

The loudness of speech and noise is calculated following the ISO 532-1975 standard completed with improvements proposed by Zwicker and Fastl (1999) for sounds with strong low-frequency components. From the loudness calculated for the left and right ears, the binaural loudness can be derived following the approach of Moore and Glasberg (2007). One important issue in these calculations is that speech and noise are mixed beforehand (see Section 5.2.3.1), so...
they are not directly separately available for the different listening conditions. For the unoccluded situation and the recordings with passive hearing protectors, speech and noise are processed linearly and independently. Therefore recordings of reference speech noise in both listening conditions can be used to calculate the loudness of the speech in absence of masking noise. Speech noise is preferred over the original fragments with monosyllabic words because the silent intervals in those samples complicate the correct estimation of the overall sound pressure level of the target signals. The loudness of the noise is based on the test fragments themselves. This approach is valid because the level of the noise greatly exceeds the level of the speech so that the final spectrum is solely determined by the characteristics of the noise. Loudness calculations of noise and speech-and-noise fragments confirm this way of thinking by leading to negligible differences that exceed by no means the variation in loudness between successive recordings of identical fragments.

Things are more complicated for the active protectors because here the processing of the speech depends on the present noise. To estimate the influence of these hearing protectors on speech loudness, the difference is made between the noise registered by the ear simulators with open ear canals and with protectors. This difference is then applied to the calibration speech noise recorded with open ear simulators and for the processed speech fragments the loudness is calculated. The loudness of the noise is again based on the original test fragments. These procedures give for each test fragment the loudness of speech and noise in function of the critical band-rate (in bark) approximating the frequency selectivity of the hearing system \cite{Zwicker1999}. The signal-to-noise ratio expressed in loudness can then be derived easily by dividing the loudness of speech and noise.

Comparing these ratios within the five sound environments confirms the conclusions drawn earlier in Section \ref{sec:5.3.2.3} an instance for the unoccluded test fragments is given in Figure \ref{fig:5.11}. It is clear that the test situations with the fork-lift trucks is the most favorable, followed by the noise from the power plant. The noise from the bottle-filling machine all yield to lower signal-to-noise ratios but spatial separation of speech and noise source seems to have a somewhat beneficial effect (see also Section \ref{sec:5.3.2.6}). When the ratios are mutually compared between the different listening conditions, the most striking results are found for the noise at the power plant where the ratio for the passive protectors (on average 0.54) clearly exceeds the ratio for the others in the higher critical band rates (on average 0.47 for open condition, 0.27 for active protectors with maximal gain and 0.31 for the same protectors with minimal gain).

\subsection*{5.3.2.6 Interaural differences}

The binaural listening system relies on two important cues to detect signals in background noise, namely the interaural time difference (ITD) and the interaural level difference (ILD) \cite{Haater1997}. The ILD is related to the head-shadow effect where the signal might be louder in one ear (head-shadow advantage) and at the same time masking sound can be attenuated by
the shadowing of the head if they originate from directions that are away from that ear \cite{Stern2006}. Hence, spatial segregation of speech and noise might indeed yield to a more favorable signal-to-noise ratio in one ear which can improve speech recognition \cite{Bronkhorst2005}. The influence of the ILD on speech recognition is clearly visible from the results in this study. Changing the recording position at the bottle filling machine from frontal to sideways (left) significantly increases the speech recognition scores for all listening conditions (see Table 5.6 and Figure 5.6). For most of these conditions, moving the noise source indeed lowers its level at the right ear due to head-shadow (see Table 5.3) and since the level of the speech remains constant, the signal-to-noise ratio will become more advantageous. Further relocation of the speech source to the right will similarly increase the speech level at this ear (see Figure 5.3) which in its turn ameliorates the signal-to-noise ratio and the speech recognition scores for most listening conditions. Improvements of the signal-to-noise ratio in the right ear due to spatial separation of speech and noise sources are illustrated for the open condition in Figure 5.12. However, the ILD fails to account for all observed variation in speech recognition scores. In case of the active protectors with maximal amplification, the level of the noise increases (instead of decreasing) when the noise source is moved from the front to the left but the speech recognition does improve. Conversely, moving the speech source from the front to the right does ameliorate the signal-to-noise ratio when the amplification is set to minimal, but the positive effect on speech intelligibility fails to occur. These findings compel the study of other binaural parameters like the ITD.

**Figure 5.11:** Binaural loudness signal-to-noise ratio for the unoccled test fragments registered with the right ear simulator. All abbreviation are similar to previous tables and figures.
The ITD can be assessed by calculating the normalized interaural cross-correlation function ICF

\[
\text{ICF} = \frac{\Phi_{LR}(\tau)}{\sqrt{\Phi_{LL}(0)\Phi_{RR}(0)}}
\]

with \( \tau \) the difference in arrival time between the left and the right ear and \( \Phi_{LL}(0) \) and \( \Phi_{RR}(0) \) the auto-correlation functions at \( \tau = 0 \) for each ear. Further \( \Phi_{LR}(\tau) \) is calculated as

\[
\Phi_{LR}(\tau) = \frac{1}{2T} \int_{-T}^{+T} f_L(t)f_R(t+\tau)dt,
\]

\( f_L \) and \( f_R \) being the continuous sound signals at the left and right ear (Tohyama et al., 1995).

Cross-correlation is a useful function for comparing two signals, when they are shifted relative to each other by some time called lag. An important property of cross-correlation is that it is maximized when the signals after shifting are identical. Therefore \( f_L(t) \) and \( f_R(t+\tau) \) are shifted relatively to each other with different lags to find such lag for which the cross-correlation is maximized, i.e. the time lag corresponding to the ITD. A graphical example can be found in Figure 5.13 where the amplitude of the ICF is depicted in function of time lag and frequency.

Comparing ITD and ICF between sound environments makes most sense for the two conditions with identical noise fragments - i.e. the noise from the bottle filling machine recorded sideways - and speech respectively coming from the front and from the right. For these environments, one can see whether spatial separation of speech and noise markedly influences the ITD and/or the ICF.
Figure 5.13: Interaural cross-correlation in function of frequency and time lag for the noise from the bottle filling machine frontally recorded for the unoccluded listening condition.

A similar comparison between frontally and sideways recorded noise from the bottle filling machine is much less useful since it is not guaranteed that the input noise is inherently the same. In these situations, changes in ICF and ITD are not necessarily attributable to changes in the relative positioning between head and sound source. It is for instance not unthinkable that shifts in ICF and ITD are caused by the presence or absence of other noise sources. Hence, the connection between alternations in ICF and ITD on the one hand and binaural cues for spatial unmasking on the other is then much less clear-cut.

Changes in time lag and ICF due to relocation of the speech source are investigated by comparing the global ICF in function of the lag for the sound environments where the noise of the bottle filling machine comes from the left. The time interval is set between $-1$ ms and $+1$ ms because it takes about 1 ms for a sound wave impinging perpendicular to one side of the head to travel to the other side (Tohyama et al., 1995). An example of the resulting graph is shown for the open ear condition in Figure 5.14; the conclusions for the other listening conditions are completely similar.

The most striking feature of all graphs is that ICF for negative time lags slightly lowers in favour of the ICF at positive lags as the speech source is moved from frontal to right. In Figure 5.14 the negative ITD indicates that the sound predominantly comes from the left which is naturally the case when the left ear points at a noise source substantially louder than the speech. The slight
increase in ICF for positive time lags and the slight decrease for negative lags reveal the presence of a sound source coming from the right compared to the previous situation, in this case the allocation of the speech source from the front to the right. This phenomenon is also seen for the active protector with minimal amplification hence it provides no explanation for the lesser speech intelligibility in this situation.

5.4 Discussion

The ongoing evolution in signal processing gradually relieves the technological constraints in the development of active hearing protectors. In theory these protectors might alleviate the perceived negative influence of hearing protection on speech intelligibility. Since communication difficulties are often reported as an argument for not wearing hearing protection (Nakashima et al., 2007), active protectors can enhance the safety at the workfloor in two ways; by preventing hearing loss and by lowering the risk of accidents due to malcommunication (Casali et al., 2004; Casali and Berger, 1996). The key question is of course whether these protectors can actually come up to the expectations.

The first task of any hearing protector is the prevention of noise-induced hearing loss. This implies that active protectors with electronic circuits must be able to fulfill their safety function at all times, even when the electronic components fail (Buchweiller et al., 2003). Although the requirement is a matter of course, measurements with the foam earplug in this study reveal that it can not be taken for granted.

Investigating the cause of the high sound levels under the particular protector was outside the scope of the current work. Therefore it is not beyond question whether the overamplification is due to failure of the electronic system, or, for instance, induced by leakage around the earplug itself. Anyway, it is clear that assessing the attenuation offered by both passive and active components of an active hearing protector should always be the first step.

Once it is verified that the protectors reduce the sound pressure level sufficiently, their influence on speech perception can be addressed. Here this is done by recording samples of speech and noise with and without hearing protectors and then scoring the speech recognition for a group of volunteers. It should be noted that the chosen method is not representative for communication in real working conditions. There are for instance no competing tasks to perform (Laroche et al., 2003), the level of the speech is independent of the noise level (Hormann et al., 1984) and the listening material is clearly read by a professional speaker but is at the same time much less redundant than normal sentences (Damman, 1993). The somehow unrealistic listening conditions are not a drawback since the study’s major aim is a direct comparison of speech perception under different protectors. Since the quality of their signal processing is an sich not influenced by the variables mentioned above, the more realistic conditions can be excluded from the test protocol. Moreover, a fair comparison assumes that the observed variations can be attributed to true differences between the protectors, not to
other confounding factors. This requires a very controlled test design with well-defined parameters.

To respect this requirement, equality of input signals is a key factor. Therefore lists for speech audiometry are preferred over realistic utterance since they are designed to be equally difficult (Damman, 1993). This is of major importance because each test fragments naturally consists of another set of words.

When it comes to the noise sets, samples are first recorded in realistic conditions and only then presented in different listening conditions. Second recordings are made with a HATS with occluded and unoccluded ear simulators instead of presenting the material directly to the test subject wearing the different protectors. The followed approach averts that intersubject variation in the protectors’ fit either suggests or masks the effects of a particular listening condition (Wagoner et al., 2007). In the same line of thinking a headphone is preferred to a loudspeaker to present the signals because in the latter case the influence of head movements or minor variations in the subject’s position can be as little excluded as unwanted influence from the test environment (Bronkhorst and Plomp, 1988).

However, using two recordings does increase the influence of the audio equipment and in particular of the HATS. Although distortion of the original signal is partially controlled by filtering the signal, it can not be guaranteed that the final input signals resemble the noise at the factories exactly. Nevertheless, in this project it is more important that the input signals are identical and this aspect can be very well controlled when following the described approach. Another advantage is that the different noise recordings can be easily used for other types of (active) hearing protectors, making new results comparable to those of the current study. Nevertheless, the double influence of the HATS

![Figure 5.14: (Color online) ICF in function of the time lag for the open ear listening condition.](image)
could have been reversed by using free-field microphones for the first recordings at the workfloor. However, this would clearly diminish the possibility of making realistic binaural recordings.

Within all these constraints of a controlled test design, the test fragments themselves are kept as close to reality as possible. First this implies that the protectors’ mode of operation should resemble the real working conditions. In the concrete, the level-dependency of the active protectors under study requires the mixing of speech and noise beforehand. Although the procedure somehow complicates the acoustical analysis of the different listening conditions, it does ensure that the influence of the active protectors on both speech and noise is for certain correctly included. Conversely if speech and noise are recorded as individual signals in different listening conditions, precise knowledge about the signal processing is indispensable for the mixing and the accuracy of the assumptions can never be fully guaranteed.

The test fragments are further kept as realistic as possible by a careful selection of speech and noise. This implies that the speech material provided by the developers of the ‘Brugse lijst’ is not used directly, but recordings are made in an anechoic room with the HATS placed at an appropriate distance of the sound source (Olsen, 1998). The free-field conditions in the anechoic room mimic the situation where communication partners stand relatively close to each other, minimizing the influence of environmental reverberations on the speech signal. In addition, possible effects of head-and-torso transfer functions and of binaural listening are included.

For the level of the speech, it is clear that a speaker has at least to raise his voice in all selected sound environments. Therefore a fixed level is chosen between a raised (63 dB(A)) and a loud (71 dB(A)) female voice (Olsen, 1998). Alternatively, the actual speech level for a speaker with unoccluded ears could have been estimated with the formula of Plomp (Corthals, 2004). However, tests are performed for noise levels at which the use of hearing protectors is obliged and Tufts and Frank (2003) have shown that people wearing protectors do not adequately adapt their voice level. Hence, the formula of Plomp is not entirely suitable in these conditions. Since the realistic level could not be estimated accurately, a fixed level is considered an equally valid approach, all the more because varying speech levels would complicate the comparison between sound environments.

To create the sound environments, speech fragments are combined with a variety of industrial noise samples. Different noise features are selected to include these aspects that mostly determine the masking effect of noise on speech perception; different levels are chosen, together with different spectral characteristics, different temporal patterns and different angles of incidence, the latter to include the advantage of binaural listening. Because the focus lies on noise conditions for which the hearing protectors are effectively suitable, no extremely loud or soft situations are included nor are any impulsive sounds. The fact that the different sound environments indeed have different acoustical characteristics (see Section 5.3.2) and yield to statistically significant differences in speech recognition (see Section 5.3.1.2) confirms that the protectors are tested in a certain
variety of situations. Statistical analysis also reveals that variation in speech intelligibility across listening conditions depends on the sound environment under study. In addition, the acoustical characteristics of the test fragments are analyzed to gain insight in the connection between the protector’s operating mechanism and speech intelligibility. In this regard objective acoustical parameters and speech recognition have been linked by different indexes like the Speech Intelligibility Index (SII) ([George et al. 2006] and the Speech Transmission Index (STI) ([Payton et al. 1994]. However, most of these indexes and their extensions are in one way or another based on the signal-to-noise ratio. Since most hearing protectors do not distinctly alter the signal-to-noise ratio compared to the unoccluded situation, this approach seems less appropriate to compare different types of protectors. Instead, the acoustical analyses described in Section 5.3.2 are directly used to elucidate the performance of test subjects in different listening conditions. In this matter, some findings are quite striking and might be useful not only to explain the results in this study, but also to bear in mind for further development. Nevertheless, one should be aware that there exist a lot of possible interesting acoustical parameters, all interacting closely. Therefore, conclusions upon the effects of the particular features addressed in this study are drawn with certain caution.

First, the active protector with maximal gain seems to decrease speech intelligibility, possibly because the original signal is distorted. Calculation of the cv in Section 5.3.2.4 strongly supports this hypothesis. Distortion in general is a highly unwanted side-effect of active hearing protectors ([Casali and Berger 1996] and although lowering the amplification seems to solve this problem, it remains an important issue since one can not guarantee that the maximal gain setting will not be used in practice. Moreover, excessive amplification also seems to equalize the temporal pattern of the signal (see Section 5.3.2.2), making it more difficult to benefit from the relatively silent periods for communicating. Leaving aside these problems, the active protectors with minimal amplification clearly do a good job in the most silent condition. Although they do not completely alleviate the hinder of wearing protectors, they seem to be a better choice than passive protectors when sound levels barely exceed safety limits. In these circumstances, passive protectors can be overprotective and therefore might enhance inconsistent use which in its part clearly increases the risk of noise induced hearing loss ([Voix and Laville 2005]. For louder noise conditions, the hinder caused by the passive protectors is much less prominent and they might even be beneficial for normal hearing listeners. The link between global sound pressure level and speech recognition as described in Section 5.3.2.1 is also confirmed by literature findings ([Abel et al. 1993]. However, one must bear in mind that the positive effects of the passive protectors are found for normal-hearing subjects and this excludes by no means that they could degrade communication if a hearing loss is present ([Abel et al. 1982].

Despite the positive results in the more silent sound environment, the active protectors with minimal gain fail to let the user fully benefit from binaural listening in contrast to the other listening conditions (see Section 5.3.2.6). In this
regard it must first be noted that preservation of the binaural unmasking effect does not necessarily guarantee good sound localization under hearing protection. Several lines of research seem to indicate that the relationship between sound localization and perceptual segregation is not as tight as one might expect despite the fact that they depend upon the same acoustical cues (Edmonds and Culling, 2005; Brungart et al., 2004).

Nevertheless, studying the ILD and ITD can to a good extent explain the observed variation in speech recognition scores due to spatial separation of speech and noise. However, they do not provide clarification of the results for active protectors with minimal gain. A possible cause is the complexity of binaural listening that can not fully be described as the effect of ILD or ITD. In reality, ILD and ITD do not simply add up (Bronkhorst and Brungart, 2005) and models designed to capture binaural hearing are much more complicated than the analysis conducted in Section 5.3.2.6 might have suggested (Stern et al., 2006). A more thorough assessment is outside the scope of the work here presented, but might in future yield to better insights, for instance in the aberrant results for the active protector with minimal gain.

In this respect, their apparent unequal attenuation in the lower frequencies might also be a lead of departure for research. Conversely the increased difference between the left and right ear does not necessary hamper communication, given the positive results of the active protector for the fork-lift trucks and for the frontally recorded noise of the bottle filling machine. This is reassuring because at the workfloor it is probably even more difficult to achieve a perfectly balanced fitting.

The distinct connection between acoustical parameters and speech recognition described above could open the door to a more formalized relationship, for instance based on regression analysis. If a standard set of sound material is used, this approach would allow to predict the influence of new protectors on communication from recordings with a HATS. In this research project, there are insufficient fragments to establish such a model because of the complex interaction of the influential acoustical parameters. However, the test setup does permit to enlarge the existing data because the same fragments can be used for new recordings with different protectors and then the speech intelligibility can be scored again for test subjects. If the test protocol is conscientiously followed, more protectors can be mutually compared and the speech recognition scores can be combined with results from acoustical analysis into a predictive model. These efforts seem surely advantageous because once a valid model is built, the effect of innovations in hearing protectors on speech perception can be assessed more easily before time-consuming tests with human subjects are to be carried out.

From the acoustical analysis on the 20 test fragments here included, the global sound pressure level in the unoccluded situation seems an important parameter to decide upon the use of standard or augmented protectors. For the latter protectors, possible distortion must certainly be addressed, together with the preservation of the more silent periods in the noise time pattern and the benefits from binaural listening. Nevertheless, tests with human subjects seem
indispensable in the end because it will be very difficult to include all influential factors for speech recognition for all communication situations.

5.5 Conclusion

From the research conducted in this chapter, it becomes clear that speech recognition with hearing protectors strongly depends on the interaction of sound environment and hearing protector under study. In general, passive protectors do not seem to have a clear adverse effect on speech intelligibility and might even offer some improvement as long as the input level is sufficiently high. In more silent conditions, active protectors tested in this study might be a better choice if the amplification is set sufficiently low. However, a critical issue with these protectors seems preserving the beneficial effects of spatial unmasking. The apparent variation among results strongly suggest that newly developed augmented protectors should be rigorously tested beforehand in different sound environments. The protocol here proposed allows such testing, in addition it permits to reuse the input material for similar recordings with other protectors so that more protectors can be mutually compared. Establishing a predictive model would make it easier to judge the benefits of certain innovations but in the end tests with human subjects still seem essential.
Chapter 6

General discussion

This chapter will first discuss the major aspects of effective hearing conservation and the contribution of the conducted research towards the existing knowledge. Further, possible applications are sketched and general conclusions are drawn.

6.1 Effective hearing conservation

Using personal hearing protectors is a very common practice in the prevention of noise-induced hearing loss at the workfloor (Arezes and Miguel, 2002). Although the ‘hierarchy of hazard control’ states that control at the source is preferred over less reliable human controls (El Dib et al., 2007), the latter approach is often more feasible due to technological, practical and economical constraints (Arezes and Miguel, 2002; El Dib et al., 2007). However, this implies that the sound pressure level at the workfloor remains unaltered and hence hearing protectors are only effective if they are worn both correctly and consistently. Therefore, hearing conservation programs have been developed to enhance successful implementation. In this, two major cornerstones can be distinguished. First, the protector must be adapted to the user in terms of comfort, acceptability (possibly with other personal protective equipment), necessary environmental awareness and other specific conditions (Bernon and Rogez, 2007). Secondly, training, education and control seem indispensable for proper use (Trabeau et al., 2008; Casali and Park, 1991).

6.1.1 General comfort aspects

The currently conducted research merely focuses on custom-made hearing protectors. This is a well-thought choice because in hearing conservation, the importance of comfortable and wearable protectors can not be underestimated (Hsu et al., 2004). Concerning these issues, Neitzel et al. (2006) and Steenkamp (2007) report that custom-made hearing protectors are more positively rated by workers with respect to comfort, perceived protection and general acceptabil-
ity. Along with this, the necessary attenuation seems easier realized and the reported variability is lower, both in time as between subjects (Abel and Rokas, 1986; Neitzel et al., 2006). Moreover, custom-made earplugs may fit ear canals that other plugs may not (Voix and Laville, 2009b).

6.1.2 Environmental awareness and communicative abilities

Closely connected to the concept ‘comfort’ is the environmental awareness and the communicative abilities under hearing protection, this subject is addressed in Chapter 5. It is a matter of course that a hearing protector should preserve an individual’s auditory sensitivity, but if the device can additionally provide augmented auditory perception, it will more likely be worn, improving the hearing conservation effort as well as general safety (Casali and Berger, 1996). In theory, properly selected standard protectors do not necessarily affect the perception of warning signals or speech for normal hearing subjects and can even improve it in some cases (Giguère et al., 2008). When the levels in the unoccluded ear clearly exceed 85 dB(A), the passive protectors tested in Chapter 5 confirm these statements.

Nevertheless, this positive outcome with standard protectors reported both here and in literature (Abel et al., 1993; Dolan and O’Loughlin, 2005) should be nuanced by taking into account the more demanding aspects of on-the-job communication (Casali et al., 2004). In particular, speech production under hearing protectors appears to be an important issue as the speech level is less adapted to the background noise, the frequency content shifts (Tufts and Frank, 2003), the speech rate is elevated and the silent breaks are shortened (Hörmann et al., 1984). All this contributes to the fact that verbal communication with standard hearing protectors is often found cumbersome (Hong et al., 2008; Okpala, 2007; Lwow et al., 2007; Nakashima et al., 2007). The above reasoning partially explains the need for augmented hearing protectors. Furthermore, the currently conducted research confirms that augmented protectors might be a better choice when the background noise is relatively low and more fluctuating (Casali and Berger, 1996). Other fields of application are persons with sensorineural hearing loss. Abel et al. (1993; 1982) have shown that sound attenuation and hearing impairment might introduce severe communication problems, although Dolan and O’Loughlin (2005) somehow contradict this point of view. Anyway, adapted sound attenuation is extra challenging for this population since both audibility problems and suprathreshold deficits like reduced temporal resolution are present (George et al., 2006).

Besides all possible benefits, the results described in Chapter 5 also hold a clear warning for blind faith in new technologies as one of the protectors amplified the sound level above safety limits and the other distorted the signal when the gain was maximal.

To sum up, standard protectors do not by definition hamper communication, but augmented protectors might offer added values for specific users in well-defined environments. In this regard, many advanced signal processing algorithms may
6.1. Effective hearing conservation

potentially solve some problems, but extensive research is needed to assess appropriateness and cost-effectiveness (Chung et al., 2008; Chung, 2007). For development, theoretical models might address the influence on speech perception in a first stage, but intelligibility depends on such a complex of factors (described in Section 5.1) that finally tests with human subjects seem indispensable.

6.1.3 Education, training and monitoring

A large part of the research here described is conducted to gain insight in and to elaborate the objective MIRE method, verifying the performance of hearing protectors in situ. This also plays an important role in hearing conservation since providing a perfectly comfortable and suitable hearing protector is one thing, teaching the user to wear it correctly and consistently appears another. About this subject, it seems very difficult to predict the attenuation realized by an individual wearer from the values measured in laboratory conditions. Therefore adapted laboratory tests are developed to enhance similarity with the situation in practice, but they still lead to upper achievable attenuation limits (Berger et al., 1998) and can not account for the variability at the workfloor. The discrepancy can be explained by certain factors like comfort, fitting and training, education and motivation and program management. All are influential in practice but difficult to incorporate in laboratory conditions (Berger et al., 1996).

The above-mentioned problems can partially be solved when estimating the actual protection by qualitative in situ measurements (Neitzel et al., 2006). These efforts seem necessary because the labour to select the right protectors and to monitor the worker’s hearing status is quite useless if one can not verify that the hearing protectors will do their job for the individual user. Furthermore, in situ tests are not only advisable when the protectors are first introduced, but might also be included in the broader training procedure (Tsukada and Sakakibara, 2008; Gaudreau et al., 2008) and/or in the monitoring of the protector’s aging process (Kotarbinska et al., 2007).

The attention for measurements at the workfloor has yield to a broad gamut of subjective and objective in situ techniques (Hager, 2006) among other things dictated by the variety in hearing protectors and available test facilities. Each test protocol has its own strengths and weaknesses, but aspects like speed and required acoustical environment gain importance if tests are to be conducted on a more regular basis (Berger et al., 1996). In this, objective measurements outclass subjective procedures, all the more because some authors report higher variability with subjective approaches due to inconsistencies in the subject’s threshold response (Casali et al., 1995).

For those reasons the conducted research has focused on the application of the objective MIRE technique. Moreover, the implementation for custom-made hearing protectors has been elaborated because these devices have certain advantages (see Section 6.1.1) but still merit from individual testing (Berger et al., 1996; Voix and Laville, 2009b). In the concrete, the approach described by Voix (2006) is chosen as a starting point to determine an individualized relationship
between the sound recorded by the MIRE measurement microphone and the level of interest at the eardrum. The selected procedure is based on a convenient setup with stable equipment, allowing to test the protector as in real working conditions.

Predicting the sound pressure at the eardrum from measurements with the MIRE measurement microphone is only meaningful if the transfer functions are actually stable and reproducible. Measurements with a HATS and human subjects described in Chapters 2 and 3 clearly show that this requirement is fulfilled. Further, the transfer functions are only applicable for in situ verification of hearing protectors if they are independent of the achieved attenuation. Otherwise the correct attenuation should be known beforehand to allow the selection of the applicable transfer function, making the implementation utterly impossible. Fortunately, Chapters 2 and 3 also reveal that the transfer functions are not affected by the earplugs’ attenuation.

The above reveals that the measurement results were very promising, but they do not provide thorough insight in the acoustical principles that govern them. However, this is indispensable if the transfer functions should actually be implemented, either to predict the sound pressure at the eardrum for an arbitrary user or to extend the field of application to other earplugs’ designs. Therefore a numerical FDTD model is established for the acrylic hearing protector under study. These simulations reveal that the earplug’s material impedance, the dimensions of the test bore, the length of the hearing protector (determining the length of the residual cavity between hearing protector and eardrum), the interaction between earplug and ear canal and the shape of the plug all influence the fluctuation of sound pressure amplitude between the MIRE microphone and eardrum. Naturally, also general characteristics of the outer and middle ear like the impedance at the eardrum, the impedance of the ear canal and the dimensions of the residual cavity play a part.

The fact that the transfer functions depend on the dimensions of earplug and ear canal explains why their exact spectrum is strongly related to the particular ear with its custom-made earplug. In this matter, Chapter 3 reveals that the general numerical model for an ear canal occluded by an earplug with two inner bores can be adapted for a specific ear if the geometrical features of ear canal and earplug are included in detail.

Hence, for frequencies below 5500 Hz, the sound pressure at the eardrum can be accurately predicted from the registrations of the MIRE measurement microphone by applying the FDTD simulated transfer functions. In the higher frequency range, the uncertainty increases because the differences between the measured and simulated transfer functions become larger as does the variation among the measured transfer functions themselves. As elaborated previously, this might be related to impreciseness of the numerical approach, but the question remains whether it is actually possible to perform valid measurements for higher frequencies given the sensitivity to the exact position of the earplug (Voix and Zeidan, 2009; Berger et al., 2007).

It should be noted that the transfer functions from this study are highly individualized to account for the large intersubject variability reported in Chapter 4.
This is not only important from an experimental and academical point of view, but also in practice it might help to stay as close to the individual situation and measurement as possible since working with average correction data might increase uncertainty.

However, the necessary geometrical measurements and computation costs of the FDTD simulations make their practical implementation very difficult. Instead the simulations serve as a base for individualized filters that can be implemented in measurement equipment so that the sound pressure at the eardrum can be calculated from the MIRE measurement as elaborated in Chapter 4. If the length of the test bore and of the residual cavity are known, the specific filter characteristics can be determined with multiple linear regression or the MOVF technique. The models themselves appear quite complicated, but once they are established it takes negligible computer time to accurately determine the transfer function for a particular ear if the length of the test bore and the residual cavity are known. Determining these two dimensional parameters seems also feasible in practice as explained in Chapter 4.

Naturally, the influence of variability between humans will decrease with decreasing frequency resolution. However, the advantage of the proposed approach is that no assumptions have to be made concerning the accuracy of the final measurement application. The established correction method is therefore as well suited for detailed FFT analyses (up to 6400 points) as for single number ratings. Moreover, an individualized approach seems even beneficial for less narrow frequency spectra like octave bands. To illustrate this, the noise reduction is calculated for all the subjects included in Chapter 3 for the test situations where the Aurical microphone is not present. More concretely, \( H^{(3)}_{m,F} \) (see Section 3.2.3.2) is on the one hand corrected with the individualized values and on the other with an average correction obtained from the transfer functions simulated in Chapter 3. In Table 6.1 the percentage of subjects is summarized for which the difference between both calculations exceeds a certain threshold. Only the higher octave bands (from 2 kHz to 8 kHz) are included because deviations are very small for the lower frequency bands.

Nevertheless, Table 6.1 reveals that systematic errors occur if average data are used. For instance, for almost one-fifth of the subjects, the attenuation with average correction differs more than 3 dB from the individualized calculations for the 2000 Hz-band, up to 8 dB for one participant. As expected, the errors become even larger with increasing frequency. Such aberrations in frequency regions where hearing is most vulnerable to noise exposure (Mills and Going, 1982) might largely compromise the efforts of in situ measurements.

The procedure described previously is also followed to assess overall attenuation ratings like the Single Number Rating (SNR, (ISO 4869-2, b)) and the Noise Reduction Rating (NRR, (ANSI S12.6-1997)). Again, the influence of the intersubject variability decreases as expected, but for some participants a deviation of 5 dB is still found. Bearing in mind the 3 dB-rule where a 3 dB-increased sound pressure level halves the permissible exposure time (NIOSH), it seems better to avoid similar deviations when assessing an individual’s noise exposure.
6. General discussion

Table 6.1: Cumulative percentage of participants for which the absolute value of the difference (in dB) in noise reduction calculated with individualized or averaged transfer functions exceeds 1 dB (> 1), 2 dB (> 2) and 3 dB (> 3) for the octave bands under study (2000 Hz, 4000 Hz and 8000 Hz). The maximal difference (max(diff)) is also tabulated. The calculations are made for the earplugs with closed and open valve.

(a) Closed valve.

<table>
<thead>
<tr>
<th>Band (Hz)</th>
<th>&gt; 1 (%)</th>
<th>&gt; 2 (%)</th>
<th>&gt; 3 (%)</th>
<th>max(diff) (dB)</th>
</tr>
</thead>
<tbody>
<tr>
<td>2000</td>
<td>71.1</td>
<td>36.8</td>
<td>18.4</td>
<td>8.0</td>
</tr>
<tr>
<td>4000</td>
<td>76.3</td>
<td>42.1</td>
<td>23.7</td>
<td>7.5</td>
</tr>
<tr>
<td>8000</td>
<td>81.6</td>
<td>65.8</td>
<td>52.6</td>
<td>11.5</td>
</tr>
</tbody>
</table>

(b) Open valve.

<table>
<thead>
<tr>
<th>Band (Hz)</th>
<th>&gt; 1 (%)</th>
<th>&gt; 2 (%)</th>
<th>&gt; 3 (%)</th>
<th>max(diff) (dB)</th>
</tr>
</thead>
<tbody>
<tr>
<td>2000</td>
<td>55.2</td>
<td>34.2</td>
<td>10.5</td>
<td>8.0</td>
</tr>
<tr>
<td>4000</td>
<td>71.0</td>
<td>47.4</td>
<td>23.7</td>
<td>6.7</td>
</tr>
<tr>
<td>8000</td>
<td>84.2</td>
<td>73.6</td>
<td>50.0</td>
<td>12.1</td>
</tr>
</tbody>
</table>

From the results and knowledge acquired in this project, different extensions and applications are feasible. The major examples will be sketched in the following section. In this, it is important to realize that the current transfer functions are established for a specific earplug made in acrylic. Extension to other materials and geometries can be made by adapting the existing numerical models accordingly.

In addition, adaptations to the MIRE setup itself might also be necessary if new adoptions are of interest, given for instance the results obtained for silicone plugs (see Section 2.3.1.2). Here, the discrepancy between simulated and measured transfer function suggests that in the current setup leakage of sound – either around the MIRE probe or through the hearing protector itself – raise the sound pressure level at the MIRE measurement microphone. This does not render the MIRE approach unsuitable for silicone, but it points out that the concrete implementation should be altered so that the registrations of the measurement microphone become more valid. At the same time, the findings show once more that thorough research – combining experimental and numerical methods – is indispensable when broadening the current technique’s field of application.

6.2 Extensions and applications

6.2.1 Monitoring noise exposure under earplug at the work-floor

As stated in Chapters 2 and 3, the MIRE setup allows normal wearing of the hearing protector and inserting the probe does not alter the attenuation. This
implies that for the MIRE measurement realistic noise can be used as input signal. Additionally, the relatively simple instrumentation could allow to perform the tests at the user’s actual workplace. Moreover, the ongoing miniaturization of data acquisition systems might even enable registration over a longer time period.

The latter seems particularly interesting because the consistency with which a protector is worn, largely determines its effectiveness (Gaudreau et al. 2008) and testing the earplug at one moment does not guarantee overall correct use (Berger 1986). In addition, exposure under earplug might vary in time due to fluctuations in background noise and (facial) movements (Pääkkönen 2007; Abel and Rokas 1986).

The conducted research allows to predict the sound pressure at the eardrum from registration by the MIRE measurement microphone and this on an individual basis. A new challenge would be to incorporate this approach in an earplug that can monitor the sound pressure under the hearing protector continuously, for instance with a built-in microphone. In this matter, the specific position of the microphone will influence the transfer functions necessary to assess the sound pressure at the eardrum, but this can be addressed by studying the pressure fluctuations in the ear canal with the available FDTD models. Further, the filter approach proposed earlier might help to put the necessary corrections into practice.

The numerical approach is not only useful to facilitate the correct implementation of built-in microphones, it might also help to study possible interventions lowering the difference between the sound pressure level at the MIRE measurement microphone and the sound pressure level at the eardrum. For instance, the design of the test bore can be adapted, as well as the absorbing characteristics of the earplug. As for the substance, it might also be right to investigate the transfer functions for fabrics that are rated high with respect to comfort and usability.

All here described procedures allows to measure the exposure level and variability for each individual, but does not reveal whether the noise at the eardrum is actually harmful. Although the question is evident, the answer is less clear because to-date all legislation is based on free-field measurement and therefore conversion from measurements at the eardrum to free-field values seems indispensable. Conversely, it can be assumed that the effects of noise on humans are a result of the actual exposure at the eardrum and hence it seems strange to measure at the eardrum and then adding uncertainty by extra calculations (Hammershøi and Møller 2008). In the future, technological achievements might lead to an increased insight into actual eardrum exposure and the relation to noise-induced hearing losses. This may eventually lead to criteria that are more accurate, but in the meantime, conversion seems the only practical possibility (Hammershøi and Møller 2008).

For the sound exposure under hearing protector, calculating the free-field equivalent is little intuitive given the influence of the hearing protector on the transfer function. Instead, it seems more logical to convert harmful thresholds, for instance the critical levels from Mills and Göing (1982), to their level at the
eardrum using average head-related transfer functions. In this regard, Hammershøi and Møller (2008) report no lower statistical uncertainty when individual rather than literature data are used, a point of view confirmed by Casali et al. (1995).

6.2.2 Verifying noise exposure under earplug in test conditions

The transfer functions derived in this research project are directly useful if the sound level under the hearing protector is of interested as described in the previous section. However the MIRE measurement is not always performable with realistic noise and then the insertion loss might be more informative than the sole level at the eardrum.

The reason is that the insertion loss can be calculated from tests with artificial signals like pink noise because the final result is independent of the input noise. In this, the established transfer functions still allow to determine individually the sound pressure at the eardrum from the MIRE measurement. Subsequently, the sound pressure at the eardrum can be combined with the applicable transfer functions to calculate the insertion loss. Here a general scheme is drawn, leaving the possibility for optimization and individualization open.

In the remainder, all acoustical quantities are represented by one single symbol referring to the global sound pressure level and this for easy reference. Naturally, similar calculations are also possible for separate frequency bands. This will enhance accuracy, but the level of obtainable precision naturally depends upon the possibilities of the data acquisition system.

Berger (1986) defines the insertion loss (IL) as the difference between the sound pressure in the unoccluded \(B\) and occluded \(B'\) ear canal or from Figure 6.1

\[
IL = B - B'.
\]

The noise reduction (NR) on the other hand is the difference between the sound pressure in the undisturbed field \(A\) and in the occluded ear canal or

\[
NR = A - B'.
\]

However, in practice the reference microphone included in the MIRE probe measures the sound pressure at the earplug \(A'\) for reasons of convenience and stability. The approach yields to the measured noise reduction \(NR_m\) as

\[
NR_m = A' - B'.
\]

The difference between NR and \(NR_m\) is the transfer function from the free-field to the blocked entrance \(T_b\)

\[
T_b = A - A'
\]

which can in theory be measured individually, but in practice one might prefer an average transfer function as for instance tabulated by Hammershøi and Møller (2008).
6.2. Extensions and applications

(a) Points of interest for the open ear.

(b) Points of interest for occluded ear.

Figure 6.1: Points of interest for sound pressure distribution around the open and occluded ear canal. To calculate Insertion Loss and Noise Reduction, the sound pressure in the undisturbed field (A), at the blocked entrance (A’) and at the eardrum for both the unoccluded (B) and the occluded ear canal (B’) should be used.

When applying these average values, one must take into account the directional dependency of \( T_b \) – in contrast with the transfer functions between MIRE measurement microphone and eardrum (see Section 2.4). In this, Hammershøi and Møller (2008) distinct free-field-front and diffuse-field transfer functions.

Despite the reasonable accuracy with which the transfer function at the blocked entrance can be standardized across humans (Hammershøi and Møller 2008), the fact remains that uncertainty increases by introducing non-individualized data. The variability is clearly more distinct for higher frequency regions; combined with the decreased accuracy for the MIRE microphone-to-eardrum transfer functions above 5500 Hz (see Section 3.4), the NR seems less reliable for higher frequencies. But again, the reliability of attenuation measurements for higher frequencies has already been questioned (Voix and Zeidan 2009; Berger et al., 2007).

From the noise reduction the insertion loss can be derived with the transfer function of the open ear \( T_o \)

\[
T_o = A - B
\]  

(6.5)

by combining Equations 6.3, 6.5 and 6.1, namely

\[
\text{IL} = \text{NR} - T_o.
\]  

(6.6)

Casali et al. (1995) have verified that the insertion loss and noise reduction are essentially equivalent quantities when the noise reduction attenuation values are corrected for the individual’s transfer function of the unoccluded ear canal, so either can be used to estimate the other. In addition, the variation of \( T_o \) seems limited, suggesting that average constants may serve as an adequate correction (Casali et al., 1995; Hammershøi and Møller, 2008), bearing in mind that these computations will nevertheless increase uncertainty. Finally, the remarks con-
cerning directional dependency and increased variability with frequency that are made for \( T_b \) also hold true for \( T_o \).

One might notice that none of the above calculations include bone conduction as sound path. This seems acceptable for the earplug under study since different authors report that the ear canal is the dominant route of sound conduction to the cochlea if either earplugs or earmuffs are worn (Ravicz and Melcher [2001]; Behar [2004]). This is confirmed by the comparison of the maximal attenuation values in Table 3.1 and the limits for bone conduction in Table 6.2. In other cases, the possible influence of bone conduction can be judged by comparing the perceived attenuation with limits for bone conduction given by Hiselius [2005] and Berger and Kerivan [1983]; Berger et al. [2003]; Berger [1986].

Finally, no pronouncement is made upon the transducer used to present the test signal. In this regard, using headphones instead of free-field loudspeakers might make testing considerable more portable and less sensitive to the acoustical environment, provided of course that the cups are sufficiently large i.e. circumaural (Berger [1986]). Deciding upon the preferable transducer is beyond the scope of this work, but it is verified in Section 2.3.2 that the individualized transfer functions between MIRE measurement microphone and eardrum are transducer independent.

### 6.2.3 Verification and development of augmented custom-made earplugs

One of the major shortcomings of the current laboratory standards is their inability to test augmented protectors such as the active protectors described in Chapter 5 (Casali and Robinson [2003]). If those protectors are equipped with a test bore, the sound pressure at the eardrum can be individually determined with the approach investigated in this project. Moreover, insight in the sound pressure fluctuations in an occluded ear canal might be useful for augmented earplugs in general, but also for active communication sets where the signal is captured at the speaker and electronically transmitted to the loudspeaker in the listener’s earplug. FDTD simulations that model the transfer function between the loudspeaker and the eardrum can provide insights on how to enhance sound quality and speech perception.

As for the protector’s operating mechanism itself, amplification should always respect the safety limits. Further, distortion of the input signal is to be avoided at all times and the benefits of binaural hearing should be preserved as much as possible. In the hearing protectors under study, signal processing is enhanced by amplifying the frequencies with speech information. However, one could also

<table>
<thead>
<tr>
<th>Freq (Hz)</th>
<th>125</th>
<th>250</th>
<th>500</th>
<th>1000</th>
<th>2000</th>
<th>3150</th>
<th>4000</th>
</tr>
</thead>
<tbody>
<tr>
<td>Att (dB)</td>
<td>54</td>
<td>49</td>
<td>52</td>
<td>51</td>
<td>38</td>
<td>47</td>
<td>46</td>
</tr>
</tbody>
</table>

Table 6.2: Bone conduction estimate as reported by Hiselius [2005].
opt to alter amplification in time, for instance by an increased amplification if speech signals are present.

Although new algorithms might theoretically enhance communication, rigorous testing seems indispensable to verify their actual impact on speech recognition in noise. Since those tests are in general time-consuming, devoting extra attention to the establishment of a stable and valid protocol might increase the overall efficiency.

At the first stage, the performance of new protectors can be addressed with a HATS. In this matter, a variety of realistic background noise samples should be selected to assess fully the processing of the input signal. Further, the repeatability should be maximized so that different recordings are mutually comparable in time. The thus obtained sound samples can be analyzed in different ways; Chapter 5 has revealed certain acoustical characteristics that should be taken into account.

Measuring the global sound pressure is of outermost importance to verify that the exposure safety limits can be respected at all times. Comparing this parameter under hearing protector with the unoccluded situation might also indicate whether lowering the global level will be beneficial for speech intelligibility. In addition, more detailed spectral analysis might among other things reveal excessive amplification in certain frequency ranges which could distort the signal. Further, unbalances in signal processing of the left and right hearing protector might also come to light, but the effect on speech intelligibility seems not clear-cut from the research conducted in Chapter 5. Based on the spectral analysis, the loudness ratio between speech and noise can be calculated. A more favorable ratio can naturally enhance speech intelligibility.

Apart from the above parameters, the temporal pattern of the fragments might also be investigated. It seems for instance important to respect the relatively silent periods in more fluctuating noise.

Finally, the benefits from binaural hearing might be addressed by comparing changes in ILD and ITD between well-defined fragments. In addition, a more complex model for binaural listening might clarify the speech recognition results that are to-date not explainable by the investigated acoustical properties of speech and noise (see Section 5.3.2.6).

It should be noted that the interaction between the here mentioned acoustical characteristics is quite complex and further speech intelligibility tests with human subjects might even enlarge the set of influential variables. In this matter, a stable protocol is again of outermost importance to make the results from different test episodes mutually comparable so that knowledge can be gathered in an efficient way. A broader dataset might permit to formalize the relationship between speech recognition under hearing protectors and certain acoustical aspects of the fragments. This information can help in the development, assessment and selection of new algorithms, but in the end tests with human subjects seem indispensable due to the complexity of influential variables on verbal communication.
6.3 General conclusion

Previous research has clearly demonstrated that hearing protectors might be effective in the prevention of noise-induced hearing loss if implementation is covered by a well-thought hearing conservation program. The key factor in this is the individual’s ability and motivation to wear the protectors both correctly and consistently, which seems only feasible if the protector is selected in accordance with the user’s preferences and working environment. Eventually, the balance has to be found between sufficient attenuation on the one hand and preserved environmental awareness on the other.

The here presented procedure to verify the attenuation in situ starts from a highly individualized approach and is kept transparent to allow other applications. The current models are established for acrylic custom-made earplugs, but certain parameters - for instance the plug’s material - can be changed quite easily in the numerical simulations, enlarging the field of application. In addition, the gained knowledge can serve as a steppingstone for measurements at the workfloor, continuous registration of exposure under hearing protectors and the optimization of augmented protectors.

The variation in hearing protectors and industrial conditions paves the way to different needs and possibilities for in situ measurements, bringing along a variety of techniques. This is not a priori problematic as long as the approaches are well-thought, reliable and yield to comparable results. Making the investigated technique suitable for multiple settings and requirements is one of the reasons for aiming at compliant models in which the different components and calculation steps are clearly identifiable and verifiable.

As for environmental awareness, new technologies and insights raise hope for better speech intelligibility and detection of warning signals at the workfloor, even for hearing-impaired users. Nevertheless, a blind faith in the theoretical improvements seems fundamentally wrong, making rigorous testing indispensable. At a first stage, important acoustical parameters can be addressed from HATS recordings under new hearing protectors. In this matter, a valid and stable protocol can help to gather knowledge over different test episodes and an extended dataset can open the door to a more formal relationship between acoustical parameters and speech intelligibility, thus providing information for the development, assessment and selection of new technologies. However, verbal communication appears so complex that in the end tests with human subjects are essential.


Fowler, C. G. and Shanks, J. E. (2002). Handbook of clinical audiology, chapter Tympanometry, 175–204 (Lippincott Williams & Wilkins, Maryland).


NIOSH (????).


