ELSEVIER

Hearing Research 167 (2002) 1–12

HEARING RESEARCH

www.elsevier.com/locate/heares

Air versus bone conduction: an equal loudness investigation

Stefan Stenfelt *, Bo Håkansson

Department of Signals and Systems, Chalmers University of Technology, SE-412 96 Göteborg, Sweden

Received 6 March 2001; accepted 27 September 2001

Abstract

Air conduction (AC) versus bone conduction (BC) loudness balance testing was conducted at frequencies of 0.25, 0.5, 0.75, 1, 2, and 4 kHz for two groups: 23 normal hearing subjects and eight subjects with a mild to moderate pure sensorineural hearing loss. Narrow-band noise was presented interchangeably between earphones and a bone transducer fitted to the subjects. Loudness matching was carried out at each frequency and at the levels 30–80 dB hearing level (HL) (10 dB steps) in the following manner: the sound pressure from the earphones was fixed and the subject adjusted the output level of the bone transducer for equal loudness by bracketing the standard. The results revealed somewhat different loudness functions for AC and BC sound with a 6–10 dB difference in the AC and BC loudness functions for the normal hearing group over the dynamic range 30–80 dB HL at the frequencies 250–750 Hz. At the higher frequencies, 1–4 kHz, the difference was only 4–5 dB over the same dynamic range. Similar results were obtained for the sensorineural hearing-impaired group. The difference between the AC and the BC loudness functions may originate from changes with level of the AC sound path, e.g. contraction of the stapedius muscle, but also distortion from the bone transducer and tactile stimulation could have contributed to the results seen. © 2002 Elsevier Science B.V. All rights reserved.

Key words: Bone conduction; Loudness estimation; Acoustic reflex

1. Introduction

The concept of hearing by bone conduction (BC) in man has been known for several centuries. During this time, investigators have tried to characterize this sound pathway and to compare it with the ordinary hearing route known as air conduction (AC), involving the outer, middle, and inner ear. A fundamental question of BC stimulation was if it is a cochlear stimulation or a stimulation of some other end organ. The first one to address this question was Békésy whose hypothesis was the following: if a sound transmitted by either AC or BC stimulates the basilar membrane the same way, they should be able to cancel each other if they end up with the same amplitude but 180° out of phase. He was able to conduct a subjective cancellation of an AC and BC tone of 400 Hz in one subject (von Békésy, 1932). A couple of years later, Wever and Bray (1936) measured the cochlear microphonics in cats by both AC and BC sounds and found them to be similar, yet another indication that AC and BC stimulate the basilar membrane in a similar way. Wever and Lawrence (1954) extended Békesys findings by cancelling an AC tone with a BC tone over a frequency range from 0.1 to 15 kHz in animals while measuring the cochlear microphonics. Khanna et al. (1976) reported subjective cancellation of AC 1 kHz tone by BC 1 kHz tone over a range of 40–70 dB hearing level (HL); they did report a residue of the second harmonic at cancellation. Further, theoretical models of the wave motion in the cochlea show similar basilar membrane motion with AC as well as BC stimulation (Zwislocki, 1953; Tonndorf, 1957, 1962).

In a model experiment of the wave motion on the basilar membrane Békésy (1955) was able to show that, regardless of the location of the stimulation of the cochlea, the waves on the basilar membrane always travel from the stiffer part towards the more compliant one, i.e. from the base of the cochlea towards the hel-

* Corresponding author. Tel.: +46 (31) 772-1770; Fax: +46 (31) 772-1782.
E-mail address: stenfelt@s2.chalmers.se (S. Stenfelt).

Abbreviations: ABR, auditory brainstem response; AC, air conduction; BAHA, bone-anchored hearing aid; BC, bone conduction; HL, hearing level; THD, total harmonic distortion

0378-5955/02/$ – see front matter © 2002 Elsevier Science B.V. All rights reserved.
PII: S0378-5955(01)00407-5

icotrema as with normal AC stimulation. Other implications of similar excitation of the basilar membrane with AC and BC sounds are the ability to generate two-tone distortion products with both a BC tone and an AC tone (Purcell et al., 1998, 1999). However, there are differences between AC and BC stimulation as well. Except for the difference in sound transmission, where both cochleae are excited when the skull is stimulated by a vibration (BC) whereas one cochlea can be stimulated if the sound is presented by an insert earphone (AC), there seems to be a different level dependence for AC and BC sound. It was found when investigating BC auditory brainstem responses (ABR) that the latency of Jewett wave V increases more with a decrease of the stimulus level than AC ABR with the same decrease of stimulus level (Schratzenstaller et al., 2000). Further, in BC-evoked otoacoustic emissions the increase in emission amplitude with increase in stimulus level is much higher for BC than for AC (Rossi et al., 1988). Another indication of difference between AC and BC sound is the ability to hear ultrasonic sounds (up to 100 kHz) through BC while inaudible through AC (Lenhardt et al., 1991).

The level difference between AC and BC could be explained with non-linear distortion produced in the BC transmission path. Such non-linear distortion would also give an explanation for the hearing of BC ultrasound: the distortion in the transmission path would demodulate the ultrasound to audible frequencies. However, in a study measuring the transcranial transmission properties of BC sound in a living human skull, the transmission of BC sound in the human skull was reported to be linear for levels equal to 77 dB HL for the frequencies 0.1–10 kHz (Håkansson et al., 1996). Further, many investigators have reported the mechanical point impedance of the human skull to be linear for vibration frequencies and levels appropriate for hearing (Corliss and Coidan, 1955; Flottorp and Solberg, 1976; Smith and Suggs, 1976; Khalil et al., 1979; Håkansson et al., 1986). Thus, if both the BC and AC systems are linear, it can be anticipated that there is no difference in the perceived loudness by the two different routes, i.e. equal energy stimulus above threshold should yield the same loudness level for hearing whether stimulation is by AC or BC.

Another area that indicates a different level function of AC and BC sounds is in the use of BC hearing aids. As in an ordinary AC hearing aid, the bone-anchored hearing aid (BAHA) picks up the sound with a microphone and amplifies and filters it by means of electronics (Tjellström et al., 2001). However, instead of a receiver in the ear canal, the transducer of the BAHA transmits the vibrations to the skull by means of an osseointegrated titanium implant placed in the parietal bone of the skull. Usually, the patient using a BAHA has either pure conductive loss or a combined sensorineural and conductive hearing loss (for inclusion criteria see Tjellström et al., 2001). However, patients with external otitis or eczema in the ear canal are also candidates for a BAHA: they can have a more or less pure sensorineural hearing loss (the air-bone gap is less than 10 dB) with a PTA[1] of up to 45 dB for the ear level device. The typical functional gain of the ear level device BAHA is 5 dB for a patient without a conduction loss (Carlsson and Håkansson, 1997). If the patient, in addition to the sensorineural hearing loss, has a conductive hearing loss component, the functional gain is increased by the amount of the conduction loss.

Experience with AC hearing aids has shown that a gain of 5 dB is not sufficient to compensate for a sensorineural component of 45 dB; but it seem to be for the BAHA (Tjellström et al., 2001). One explanation could be that, since the typical BAHA patient has a conductive loss as well, they get more functional gain. However, we have also found that some patients with almost normal cochlear function combined with a maximum conduction hearing loss (for example patients with a malformed middle and external ear) use a volume control setting in ordinary listening situations that does not close their air-bone gap. Although this puzzling finding cannot be explained by current theories of AC versus BC, one possibility is that these phenomena could originate from different loudness functions of AC and BC sound. Therefore, the aim of this investigation is to compare the loudness function of BC sound with the loudness function of AC sound.

2. Materials and methods

2.1. Subjects

Two groups, one with normal hearing subjects and one with sensorineural hearing-impaired subjects, participated in the AC versus BC loudness experiment. The first group consisted of 23 normal hearing people, 16 males and seven females, mostly colleagues at our department. Their ages ranged between 18 and 43 years, with a mean of 28.0 years. Their AC thresholds were for both ears better than 15 dB HL in the frequency range 125 Hz to 8 kHz, and the symmetry between the ears for each subject was better than 10 dB. The second group comprised eight persons (six males, two females) with a mild to moderate pure sensorineural hearing loss. Their ages ranged between 47 and 63 years, with a mean of 56.8 years. Each subject had an air-bone gap

[1] PTA: pure tone average, the average of the hearing loss at 500 Hz, 1 kHz, 2 kHz, and 3 kHz.

Table 1
Individual data for the eight subjects with a high frequency sensorineural hearing loss

#	Sex	Age	PTA^{a}_{air}	PTA^{a}_{bone}	Hearing loss
1	M	47	22	23	NIHL[b]
2	F	63	45	37	Unknown cause, progressive
3	M	54	14	12	NIHL[b]
4	M	62	30	33	NIHL[b]
5	M	57	34	32	NIHL[b], hereditary
6	M	59	24	15	NIHL[b]
7	F	58	22	19	Unknown cause
8	M	56	36	29	NIHL[b]

[a]PTA: pure tone average of the best ear: 500, 1000, 2000 Hz.
[b]NIHL: noise-induced hearing loss.

better than 10 dB and his or her interaural BC thresholds were within 10 dB. The individual data of the subjects in this group are presented in Table 1. They were chosen because they had also participated in a study to evaluate an optimized BAHA for a pure sensorineural hearing loss (Stenfelt and Håkansson, 2000). In that study, they were equipped with a skin penetrating osseointegrated titanium fixture, in the parietal bone just behind the ear, for attachment of a BAHA.

2.2. Procedure

To obtain the loudness function for BC sound, loudness matching of AC and BC sound was performed. According to Nolan and Lyon (1981), the transcranial attenuation for BC is on the average around 10 dB between the two cochleae for the audiometric frequencies 0.25–4 kHz. Normally, masking procedures are used to evaluate each cochlea individually. However, masking procedures used in audiometric testing can affect the perceived loudness. To eliminate any bias from the masking of the non-test ear, the AC stimulus was presented binaurally whereas the BC sound was applied unilaterally giving stimulation to both cochleae. The BC sound path is made up of multiple processes that can produce sharp frequency dependencies that occur at different frequencies at the two ears (Håkansson et al., 1994). This phenomenon can affect the loudness estimation; to minimize this possibility, narrow bandwidth noise was used as sound stimuli. Another reason for using bilateral AC and unilateral BC stimulation was to compare the two types of conduction relevant for hearing aid wearers.

Loudness summation, i.e. the effect of binaural stimulation, can affect the loudness measurements. The outputs of the earphones were equal at both ears, whereas the actual BC stimulation of the two cochleae per se, excited from one point, was unknown. The loudness summation effect of the BC sound was, accordingly, somewhere between that of a monaurally applied sound and that of a binaurally applied one of equal amplitude. It has been shown that binaural loudness summation was almost constant with level for narrow-band stimuli as used here (Scharf, 1969). Hence, the loudness summation was constant for all test levels and only gave a constant offset of the measurement results.

All measurements were performed in a sound-insulated booth. The background noise was less than 22 dBA during the measurements, which was below the lowest test levels. Each subject's hearing thresholds were tested to ensure meeting the inclusion criteria to participate in the study. Thereafter, each person was equipped with the earphones (Koss portaPro, Koss corporation, Milwaukee, WI, USA) and a bone transducer; the Radioear B-71 for those with normal hearing or the Cordelle transducer (Entific Medical Systems AB, Göteborg, Sweden) for those with hearing impairments (Tjellström et al., 2001). The Cordelle transducer was attached to the titanium fixture whereas the Radioear B-71 was applied with a headband to the skull. A power amplifier (Sony TA-N220) was connected between the audiometer and the bone transducer. The Koss portaPro are small and light superaural earphones. BC thresholds with and without these earphones attached were obtained on three normal hearing subjects for frequencies between 0.25 and 1 kHz; there was no indication that they caused any occlusion effect. With the earphones and the bone transducer attached, the binaural hearing thresholds were tested for the frequencies 0.25–4 kHz by means of a Békésy sweep. These data were later used to compute the difference between the threshold levels and the loudness estimation at 30 dB HL.

Loudness matching data were obtained by the method of adjustment. Narrow bandwidth random noise stimuli of 30 dB HL centered around 1 kHz were presented to the subject through the earphones. The stimulus was generated by a modified two-channel audiometer (Interacoustics AC40, Interacoustics, Assens, Denmark) equipped with an external control unit placed in the sound-insulated booth; with it the subject could manipulate the level of the bone transducer in steps of 1 dB. The noise alternated between AC and BC at intervals of 1 s while the subject adjusted the BC stimulus (the variable) by the method of bracketing the AC stimulus (the standard). The stimuli had a 25 ms linear rise time, 950 ms on and 25 ms linear fall time. During the 1 s the stimulus was presented through the earphones, the bone transducer was silent. Then, the same stimulus (but with the level set by the subject) was presented through the bone transducer and the earphones were silent. After that, the cycle restarted. There were no silent periods for the test subject; a sound was always present, either through the earphones or through the bone transducer.

The listener was instructed to first adjust the BC sound to be louder than the AC sound, then softer, then louder again, and continued by decreasing the range of the BC sound until the two stimuli sounded equally loud. The subject was given as much time as needed to make the match, typically between 1 and 2 min for each match, i.e. the two stimuli (standard and variable) were presented between 30 and 60 times. When equal loudness was obtained, the BC level was stored and a new level and frequency were presented by the earphones. The investigation started with 30 dB HL at 1 kHz; the level was increased in 10 dB steps up to 80 dB HL. The same procedure was conducted at 0.25, 0.5, 0.75, 2, and 4 kHz.

There are reports indicating that in loudness estimation measurements, the variable sound (the sound controlled by the subject) is underestimated (Stevens, 1956; Scharf, 1959, 1969) or estimated towards the comfortable level (Buus et al., 1998; Florentine et al., 1998). Such biases in the estimation would severely corrupt the results. Therefore, the following measurement was conducted. Four subjects from the normal hearing group conducted a loudness balance test with both the AC and BC sound as the variables at two frequencies, 750 Hz and 1 kHz. First, the equal loudness levels with the BC sound as variable were obtained and stored as described previously. Second, the controls were switched, so the subject controlled the AC sound, and the previously stored BC levels were presented. If there is no bias, the original AC levels should appear again, i.e. by tens, 30–80 dB HL. For all four persons and for all levels, the result was within ± 2 dB of the original. Moreover, three subjects from the normal hearing group were presented with both the standard and variable by the earphones. When so, a perfect match between the variable and the standard was obtained (no match deviated more than 1 dB).

All measurements were conducted in an ascending order: the next loudness matching for the same frequency was conducted at a 10 dB higher level. According to Stevens (1975), a stimulus is judged lower when it is preceded by a lower stimulus than if it is preceded by a higher one (order bias). If order bias affects the measurements, the experiment where the variable and standard were switched should show a negative result since both loudness estimations were conducted in an ascending order. However, neither positive nor negative preference of the results was found. Moreover, in the test where both the standard and variable were presented by the earphones, an order bias would give a result of the variable that is lower than the standard, which is not the case.

There was no blinding of the subjects regarding the start level of the BC sound for the loudness estimations except at the beginning of each frequency, where the BC start level was set 5–10 dB below the expected equal loudness level. Otherwise, when a new AC level was presented the BC level was left at the previous equal loudness match. There was a risk that the subjects just step up the BC sound 10 dB (10 steps) between each loudness estimation. However, the subjects were never informed that the increase between each loudness estimation was 10 dB or how many estimations they were expected to perform, only that they should conduct a equal loudness task by the method of bracketing the standard every time they were presented with a new level or frequency of the sound.

2.3. Calibration

The earphones (Koss portaPro) were calibrated on a coupler (IEC 60318-3, 1998) according to the ISO 8798 values. For amplification of the signal to the bone transducer, a power amplifier was connected between the audiometer and the bone transducer. This enabled the maximum output from the bone transducers to be over 90 dB HL at all frequencies used. The audiometric bone transducer (Radioear B-71) used for the normal hearing group was calibrated on an artificial mastoid (Brül&Kjær type 4930) according to the ISO 7566 values. The bone transducer for attachment to a titanium implant (Cordelle transducer) used in the hearing-impaired group was measured with the Skull Simulator TU-1000 (Håkansson and Carlsson, 1989). There exists no standard for the BC thresholds directly on an implant as those in the hearing-impaired group. Therefore, the reference equivalent threshold force levels proposed by Carlsson et al. (1995) were used for the calibration of the Cordelle transducer.

The absolute value of the BC sound is not used in this study but relative values that are obtained in a normalization of the BC values. More information about the normalization process is given in Section 3.1. Since high stimulation levels of the BC sound are required (about 80 dB HL), and it is well known that bone transducers usually do not perform well at high levels and low frequencies, a thorough distortion analysis of the BC sound was conducted. The total harmonic distortion (THD) of a pure tone was measured for both bone transducers at 250, 500 and 750 Hz and at the equivalent narrow band noise levels 50, 60, 70 and 80 dB HL. The result of the distortion analysis is presented in Table 2.

For 250 Hz, only a couple of the subjects exceeded BC levels of 70 dB HL and, at the other frequencies, equal loudness was generally obtained for BC levels below 80 dB HL. Both transducers produce a high amount of distortion at 250 Hz. The performance of the Radioear B-71 is generally worse than the Cordelle transducer, with up to 165% THD for levels of 80 dB

Table 2
Distortion analysis of the bone transducers used in this study

Level (dB HL)	Radioear B-71 (%)			Cordelle transducer (%)		
	250 Hz	500 Hz	750 Hz	250 Hz	500 Hz	750 Hz
50	17	< 1	1	14	< 1	< 1
60	64	2	4	31	< 1	< 1
70	102	5	12	87	< 1	< 1
80	165	22	25	115	1	1

The figures represent the THD out from the bone transducers when applied to an artificial mastoid.

HL at 250 Hz; at the same level and frequency the THD from the Cordelle transducer was also large, 115%. It should be remembered that the test subjects nearly never reached these high levels. At the higher frequencies, the THD of the Cordelle was 1% or lower for levels of 80 dB HL or lower. The Radioear B-71 produced a larger amount of distortion than the Cordelle at the higher frequencies as well: a THD of 22% at 500 Hz and 31% at 750 Hz for an output level of 80 dB HL.

A THD of 31% indicates that the second harmonic can be as large as −10 dB of the fundamental. The result of the distortion is, if the THD is high enough, an increase in loudness. At 250 Hz and 70 dB HL where the THD is close to 100% for both transducers, the stimulation can be approximated with an increase from one-third octave band noise (the original stimulation) to octave band noise (if the second harmonic is dominating the THD). For such bandwidth increase an increase in loudness of 3 dB can be expected (Stevens, 1956). At the other frequencies, no effect on the perceived loudness due to the distortion is expected.

It should be noted here that the sensitivity to BC sounds is approximately 10 dB higher at 500 Hz than at 250 Hz, i.e. a 500 Hz vibration of equal force level as a 250 Hz vibration is 10 dB greater above the threshold of hearing. However, the loudness function at 250 Hz differs from that at 500 Hz. A 250 Hz tone at 70 dB HL is perceived as loud as a 500 Hz tone at 80 dB HL (Fletcher and Munson, 1933). Since these functions oppose each other, the total effect is approximately zero.

2.4. Vibrotactile sensation

The vibrotactile thresholds at the mastoid part of the temporal bone with a bone conductor of the Radioear B-71 type were measured by Brinkmann and Richter (1983). When their results are converted to HL for BC, the vibrotactile thresholds are the following: 43 dB HL at 250 Hz, 55 dB HL at 500 Hz, 66 dB HL at 750 Hz, and 72 dB HL at 1 kHz. Consequently, at the two lowest frequencies, most of the balance measurements are made at supra vibrotactile thresholds. There are no data available of the vibrotactile thresholds when the BC sound is applied directly to the titanium fixture. However, since the sensitivity to BC sound with or without skin penetration is similar for frequencies below 1 kHz (Stenfelt and Håkansson, 1999), the vibrotactile thresholds can be expected to be similar for the hearing-impaired group as well. A tactual sensation can add to the overall perceived magnitude of the sensation, i.e. a BC sound would be judged louder than if the stimulus excites the cochlea alone. This type of cross-modal stimulation is present at the lowest frequencies and at the higher levels.

3. Results

In what follows, the notation dB HL_{AC} stands for the HL produced by the earphones and transmitted to the cochlea by way of the AC route. Similarly, dB HL_{BC} stands for the normalized BC HL; more details on the normalization procedure are given below. The notations dB_{BC} and dB_{AC} indicate the relative levels of BC and AC sound, respectively.

3.1. The normal hearing group

The results from the loudness balance test of the normal hearing group are presented in Fig. 1A–F for the frequencies 250 Hz to 4 kHz. Each person's BC results are normalized to be 30 dB HL_{BC} at each frequency when the loudness matching is conducted with the standard at 30 dB HL_{AC}. The increase of the BC stimuli for each 10 dB_{AC} increase is then added to this reference value and plotted in the graphs. All of the graphs include the individual results as dots at each level, presented as the required bone transducer output level (dB HL_{BC}) to match the loudness level of the standard (dB HL_{AC}). The AC levels are restricted to the tens 30–80 dB HL_{AC}; a dot between these levels indicates multiple responses of that particular BC level. The mean BC stimuli level is shown as a cross at each level of AC stimuli and the total mean dB_{BC} increase per dB_{AC} increase, calculated as the average increase in BC level between 30 and 80 dB HL_{AC} divided by 50, is also shown.

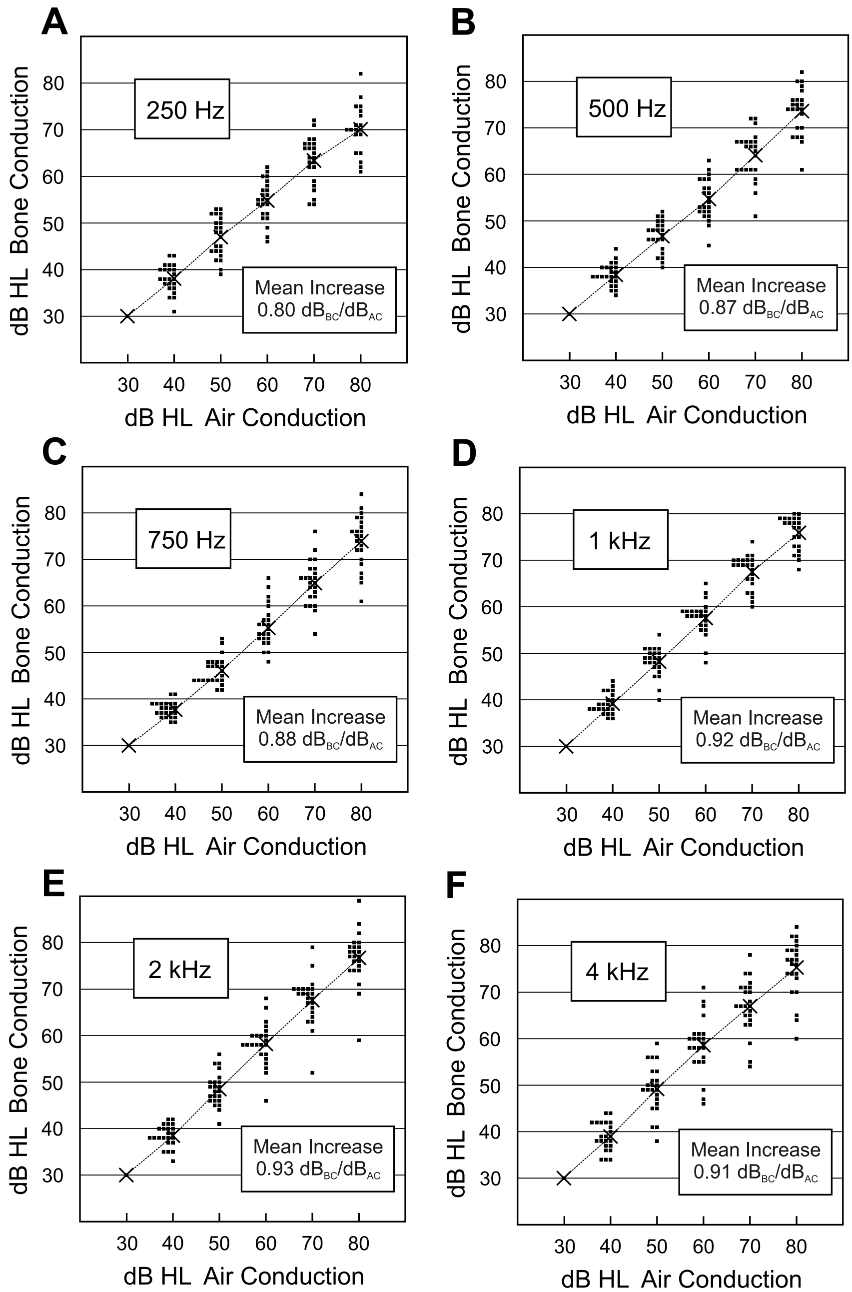

Fig. 1. The results of the loudness balance test for the normal hearing group. The dots are the individual results (measured at the tens of AC stimuli level) and the crosses spaced along the dotted lines are the BC mean for each level: (A) 250 Hz; (B) 500 Hz; (C) 750 Hz; (D) 1 kHz; (E) 2 kHz; and (F) 4 kHz.

Fig. 1A presents the results of the loudness balance test with the narrow-band stimuli centered around 250 Hz. The total average increase is 0.80 dB$_{BC}$/dB$_{AC}$ (S.D. 0.11), the lowest value for the normal hearing group in this investigation. The mean BC increase for every 10 dB$_{AC}$ increase is just above 8 dB$_{BC}$, except at the highest levels where the increase is only 6.7 dB$_{BC}$. Fig. 1B,C shows the results for the stimuli centered around 500 Hz and 750 Hz, respectively. The two plots show nearly the same result: the total average increases are 0.87 and 0.88 dB$_{BC}$/dB$_{AC}$ (S.D. 0.10 and 0.11), the mean BC increases for every 10 dB$_{AC}$ are around 9 dB$_{BC}$. Results with the stimuli centered around 1, 2, and 4 kHz are plotted in Fig. 1D–F. These three graphs constitute nearly equal results with an overall average increase of 0.91–0.93 dB$_{BC}$/dB$_{AC}$ (S.D. 0.07, 0.11, and 0.12). The mean BC increase per 10 dB$_{AC}$ increase is above 9 dB$_{BC}$ all over the whole range except at the highest intensities, where the average BC increase is slightly less. It should be pointed out that for the normal hearing group, at all frequencies and over the range 30–80 dB HL_{AC}, perceived loudness is more sensitive to BC than AC, i.e. $dB_{BC}/dB_{AC} < 1$.

3.2. *The hearing-impaired group*

The results from the hearing-impaired group, presented in Fig. 2, are given in a way similar to that of the normal hearing group, except that only the averages are presented. Since the hearing-impaired subjects have a hearing threshold of 60–75 dB HL_{AC} at 4 kHz, this frequency is omitted for this group. Moreover, their thresholds at 2 kHz were somewhere between 20 and 35 dB HL_{AC}; the starting level is set to 40 dB HL_{AC} at this frequency, a level audible for all subjects. The normalization of the BC level is computed exactly as for the normal hearing group at the frequencies 0.25–1 kHz whereas for the 2 kHz sound, the results are normalized to 40 dB HL_{BC} at 40 dB HL_{AC}.

The results for the hearing-impaired group, in Fig. 2, are fairly equal to the mean results of the normal hearing group, except at 250 Hz. Results at this frequency have the lowest increase ratio (dB_{BC}/dB_{AC}) of the frequencies tested for this group; they yield a mean of only 0.51 dB_{BC}/dB_{AC} (S.D. 0.18). The results at the other frequencies in Fig. 2 seem to be clustered in pairs. With the stimuli centered around 500 and 750 Hz, the total mean increase ratios are 0.79 and 0.84 dB_{BC}/dB_{AC}, respectively (S.D. 0.19 and 0.09), and for 1 and 2 kHz, the total mean increase ratios are 0.92 dB_{BC}/dB_{AC} for both (S.D. 0.08 and 0.09). Fig. 3 illustrates a comparison of the normal hearing and the hearing-impaired groups. On the left side of each vertical frequency line, the AC levels are displayed, i.e. 30–80 dB HL_{AC}, while on the right side, the mean of the equal loudness balance levels by way of BC are plotted. This figure clearly shows that, with an exception of 250 Hz, the results for the normal hearing and the hearing-impaired groups are similar.

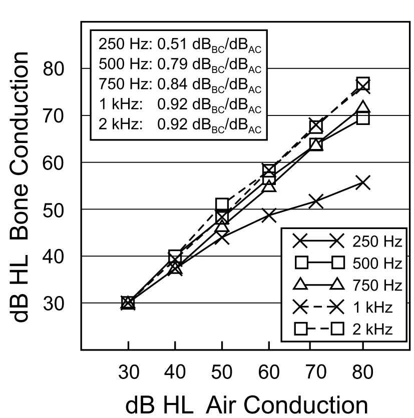

Fig. 2. The mean results of the loudness balance test for the sensorineural hearing-impaired group. 250 Hz: crosses connected with solid lines; 500 Hz: squares connected with solid lines; 750 Hz: triangles connected with solid lines; 1 kHz: crosses connected with dotted lines; and 2 kHz: squares connected with dotted lines.

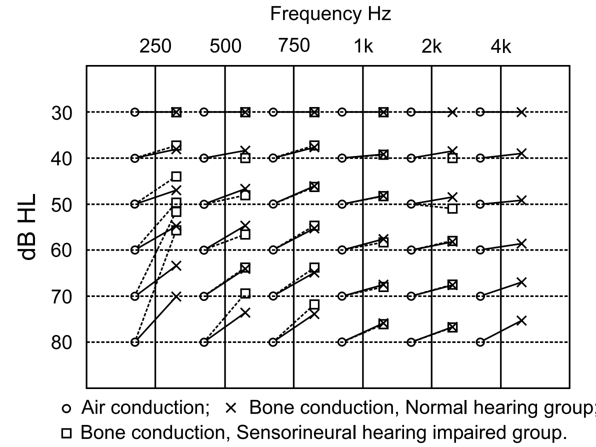

Fig. 3. The mean results from both the normal hearing group and the sensorineural hearing-impaired group. The circles indicate the AC levels and the crosses indicate the BC levels for the normal hearing group; the squares show the BC levels for the sensorineural hearing-impaired group.

4. Discussion

4.1. *Possible bias*

When conducting an equal loudness estimation task, care must be taken to avoid bias contaminating the results. The procedure of bracketing the standard yields valid data when performed correctly. However, there are several issues that could give results that can be misinterpreted as differences in the loudness function between AC and BC sound. When the earphones or the bone transducer present the stimulus, both cochleae are stimulated. This leads to a loudness summation effect. For wide-band stimuli, the binaural loudness summation effect increases with stimulation level whereas this effect is almost constant for narrow-band stimulation. Since narrow-band noise is used in these tests, the only difference in binaural loudness summation between the AC and BC stimulation is the unequal BC stimulation at the two cochleae. This results in an offset between the AC and BC loudness functions that disappears in the normalization process.

The loudness measurement was conducted in an ascending order, i.e. from 30 to 80 dB HL_{AC}. This can lead to order bias, i.e. the BC stimulus is judged lower since it is preceded by a lower stimulus. It was verified in the experiment where both the AC and BC sound were used as the variable and standard that there was no indication of an order effect in the loudness estimation procedure. One explanation may be the time it takes for each estimation. The sound at each level is presented 30–60 times before the estimation is finalized: the stimulus that is judged is then not preceded by a lower stimulus but a stimulus of similar magnitude.

Other problems that can occur are underestimation or estimation towards a comfortable level of the variable. At the start of each measurement session, the binaural hearing threshold levels, with the earphones and the bone transducer, are measured by means of a Békésy sweep. These binaural hearing threshold levels are used to compare the AC hearing threshold versus the 30 dB HL_{AC} and the BC hearing threshold versus the normalized 30 dB HL_{BC}. It is assumed that, if there are preferences in estimating the loudness towards a comfortable level, the difference between the BC hearing threshold and the normalized 30 dB HL_{BC} should be larger than the same difference for AC. If, on the other hand, the BC sound is underestimated, the result would be the opposite: the difference between the BC hearing threshold and the normalized 30 dB HL_{BC} should be smaller than the same difference for AC. It was found, however, that the average difference between AC and BC hearing thresholds and the lowest loudness estimation level is less than 2 dB for all of the frequencies without any preference towards a negative or positive value. This is taken as an indication that bias due to estimation towards a comfortable level or underestimation of the variable has a negligible influence on the results presented here.

The Radioear B-71 and the Cordelle transducer consist of a variable reluctance transducer contained in a plastic housing. Such a transducer has an inherently non-linear characteristic, especially at low frequencies and high levels. This was shown in the distortion analysis in Section 2.3. The distortion spreads the signal energy over a greater frequency interval; especially the second harmonic component is significantly large at 250 Hz and at the higher levels. This high distortion produces sounds outside the critical band that is perceived as an increase in loudness. This non-linear behavior of the bone transducer causes a function similar to the loudness function seen in this study. For the dynamic range in this test, the distortion at 250 Hz can lead to a loudness increase of over 3 dB. Even if this distortion can explain some of the difference between the AC and BC loudness level function, it is not the sole explanation. The difference between the AC and BC loudness function exists over the whole dynamic range whereas the distortion increases with level. It is only at the highest levels at the lowest frequencies that the distortion affects the perceived loudness of BC sound. At the higher frequencies, there is not enough distortion produced with the bone transducer that can explain the difference between the AC and BC loudness function. Hence, the different AC–BC loudness functions exist without any distortion from the bone transducer. The earphones did not produce any measurable distortion for the dynamic range used in this study.

4.2. Vibrotactile stimulation

At the two lowest frequencies, 250 and 500 Hz, the vibrotactile threshold for stimulation with the Radioear B-71 is 43 and 55 dB HL_{BC}, respectively. This means that at these frequencies, the loudness balance testing was conducted mainly at tactile supra threshold levels. Given two sensation inputs, the hearing stimulus and the tactile stimulus, the perceived loudness can be affected by both. The tactile sensation may add to the overall perceived magnitude of sensation and be misinterpreted by the subject as an increase in loudness. Such an increase in loudness would give a similar function of AC–BC loudness estimation as found in the results here.

Even if this is a plausible explanation for the difference found between the AC and BC loudness function, similar arguments as with the distortion of the bone transducer indicate that it is not the sole explanation of the difference. The vibrotactile thresholds at 750 Hz and 1000 Hz are close to 70 dB HL_{BC}. This means that it is only above this level that the multi-modal stimulus would affect the loudness estimation. As seen in the results, the difference in AC–BC loudness functions is present for levels well below the 70 dB HL_{BC} level. Consequently, even if the loudness estimation conducted in this study is affected by tactile stimulation, it is only at the low frequencies and the high levels this effect gives a contribution. The tactile sensations were observed by the subjects, four persons in the normal hearing group and five persons in the hearing-impaired group did not perform the balance test at the highest level at 250 Hz (80 dB HL_{AC}): they complained that the tactile sensation was too strong to be able to judge the loudness of the sound.

4.3. Physiological aspects

The two ways to deliver sound energy to the cochlea used in this investigation, the AC and BC paths, are schematically illustrated in Fig. 4. The transmission for the AC path is rather straightforward; by way of the outer ear canal to the middle ear with the ossicles, resulting in a motion of the stapes that gives a motion of the cochlear fluid. The sound pressure at the different parts of the ear induces some BC sound, as indicated in Fig. 4. However, due to the large impedance difference between air and bone, this sound is mostly reflected; consequently this part of the conduction path is considered negligible for a normal hearing ear. If an AC hearing aid is used, the earmould itself vibrates due to the vibration of the receiver connected to the hearing aid housing. These vibrations of the mould can, if the mould is seated deep down in the canal and is in contact with the bony part of the external ear canal, induce a BC component much larger than possible with free field stimulation (Hayes and Chen, 1998).

The BC path has, as indicated in Fig. 4, a rather complex form. The transducer is applied either directly to the bone or to the skin covering the bone. In Fig. 4, the place to apply the stimuli can be the temporal bone behind the ear, but it could also be either bone of the skull without violating the schematic model. The vibrations at the temporal bone radiate sounds into the outer ear canal by way of relative jaw movements and sound radiation from the cartilage and soft tissues (von Békésy, 1960; Tonndorf, 1966). Also, vibrations of the temporal bone can induce sounds in the middle ear cavity by compression of the cavity (von Békésy, 1960). More important is the relative motion of the middle ear ossicles (Bárány, 1938), which is believed to account for the major BC stimuli in the low frequency range (0.5–2 kHz) (Kirikae, 1959; Tonndorf, 1966; Khanna et al., 1976). Finally, there is compression of the cochlea and the vestibular system together with the fluid inertia of the two, which results in a displacement of the basilar membrane (von Békésy, 1960; Tonndorf, 1966).

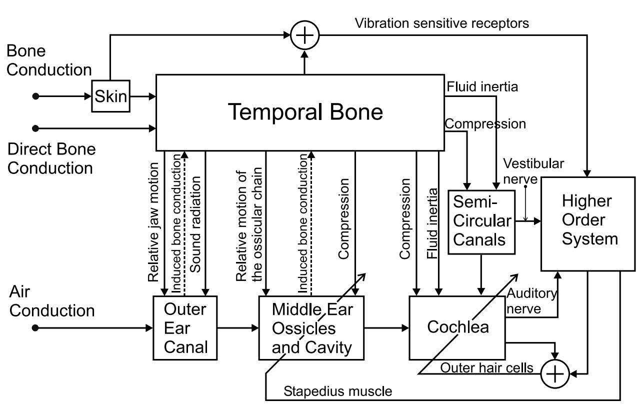

Fig. 4. A schematic illustrating possible AC and BC paths. The AC path is straightforward with a sound outside the ear that is transmitted through the outer ear canal, via the middle ear ossicles to the cochlea. The BC sound transmission path is rather complex: inertial effects and compressional effects transmit a vibration of the temporal bone to the outer ear canal, middle ear ossicles and cavity, and to the semicircular canals. All are transmitted to the cochlea and added to the direct stimulation of the cochlea by BC sound. The vibration of the temporal bone and skin influences vibrotactile receptors that transmit stimulation to the higher order system that can influence other parts of the system and/or add to the total perceived sound. Further, there are some transmission parameters, such as the stapedius muscle and the outer hair cells, that are affected by the stimulation of the higher order system.

The sound, either as a pressure outside the outer ear canal or as a vibration of the temporal bone, induces a wave motion of the basilar membrane (von Békésy, 1960) activating the neurons and is finally interpreted as sound in the acoustical cortex. The sound level affects some parameters that influence the transmission of the sound. These parameter changes are controlled locally as well as from the higher order system. An example is the contraction of the stapedius muscle. The parameter changes, which can affect the BC and AC paths differently, can explain, at least partly, the findings in this study. As discussed previously, a difference between the two paths is the vibrotactile receptors sensitive to vibrations of the skin and the temporal bone. These sensations are also transmitted to the higher order system and integrated in the whole sensation; they can also affect the parameters of the hearing paths.

It is known that the stapedius reflex can cause up to 20 dB_{AC} attenuation for frequencies below 1 kHz. It has generally been accepted that the stapedius reflex is activated first at 70–90 dB HL_{AC} for pure tones and at about 10 dB lower levels for narrow-band noise (Flottorp et al., 1971). However, it has been shown lately that the stapedius muscle starts to contract at lower levels; at least 20 dB lower than measurable with ordinary stapedius reflex measurement techniques (Neuman et al., 1996). Accordingly, the stapedius muscle can start to contract at a level as low as 40 dB HL with narrow-band noise stimulation. When the stapedius muscle contracts, the stiffness of the stapes footplate annular ligament is increased. Such a stiffness increase affects the low frequency AC transmission considerable whereas it leaves the BC sensitivity almost unaffected (similar to clinical results with otosclerosis of the stapes footplate). Hence, the contraction of the stapedius muscle can, at least partly, explain the difference between AC and BC loudness function found in this investigation.

A contraction of the tensor tympani will also increase the stiffness of the middle ear ossicles; especially the stiffness of the TM will be increased. This will, in a similar way as with the contraction of the stapedius muscle, impede the low frequency AC sound while leaving the transmission of BC sounds almost unaffected. The great difference between the contraction of the stapedius muscle and the tensor tympani muscle is that the stapedius muscle contraction is initiated by a sound stimulus whereas a contraction of the tensor tympani muscle is of tactile origin (Northern and Gabbard, 1994). As was shown previously, most of the low frequency loudness estimations were conducted above the vibrotactile thresholds. This indicates that due to the tactile sensation the tensor tympani can be contracted for some of the frequencies and levels in this study. It must be remembered that this effect is due to a tactile sensation caused by the BC sound that affects the transmission properties of the AC sound specific for this test situation. It does not indicate a difference in AC–BC loudness functions.

4.4. Comparison with previous investigations

Khanna et al. (1976), when investigating the sensitivity to BC sound, performed a cancellation experiment with AC and BC sound. They started out with a cancellation at 40 dB SL, continued in 10 dB-steps to 70 dB SL, and reported the BC response to be linear. In their study, it looks as if the loudness function for both AC and BC is the same. However, as it was a sound cancellation investigation, the perceived sound was reported by the test subject to be zero; there was no excitation of the cochlea. If the difference in the loudness functions between AC and BC originates in a parameter change due to the excitation of the cochlea, this difference is not revealed in that investigation since they achieved cancellation and there is no stimulation of the cochlea. Accordingly, no parameter change is present that affects the transfer function.

In a more recent study by Khanna and Decraemer (1996), the motions of the middle ear ossicles were investigated. They found that the ossicles move in a very complex way. 'The position of the rotation axis changed widely with frequency, and even within each cycle.' A similar kind of behavior could also be present in the amplitude, manifesting itself as a non-linear amplitude transfer function of the ossicles between the tympanic membrane and the cochlea. Such a non-linear function could give results similar to those found here. However, most investigations of the sound transmission through the middle ear in human cadaver temporal bones have shown it to be linear for the levels used in this investigation (Merchant et al., 1996).

A further indication of the higher relative sensitivity of BC sounds over AC sounds can be found by investigating the amplification provided by the BAHA. For a normal AC hearing aid, there are many different fitting algorithms. Usually, however, a half gain or third gain rule is utilized to compensate for the sensorineural part of the hearing loss. A half gain rule means that the sensorineural hearing loss is compensated in the hearing aid with a frequency specific gain that equals half that of the sensorineural loss in dB. Similarly, a third gain rule uses a frequency specific gain of the hearing aid that equals a third of the sensorineural hearing loss in dB. In an investigation of the gain settings in patients using a BAHA it was concluded that the hearing aid was used for closing the air-bone gap and almost no compensation for the sensorineural loss is used (Snik et al., 1995). One explanation for this can be that since they receive the sound through BC and according to the results in this study, the sound level at ordinary speech levels is perceived louder than if it is transmitted by AC. However, it can also be explained by the limited output levels by these hearing aids or that the ordinary BAHA user has a conductive hearing loss as well, eliminating environmental noise through the normal hearing path; the signal-to-noise ratio is not increased by a higher gain and the patient merely needs a comfortable level of the sound.

The latency of the Jewett wave V in ABR is a measure of the sound level: the shorter latency the louder stimuli. In BC ABR it is found that the latency of Jewett wave V increases more with a decrease of the stimulus level than an AC ABR with the same decrease of stimulus level (Schratzenstaller et al., 2000). This can be a result of a difference in magnitude transmission of the AC and BC path, but it can also be due to the lack of high frequency energy in the evoking BC stimulus compared with AC stimulus. A higher frequency component stimulates the cochlea more basally and, consequently, gives an earlier response (Mauldin and Jerger, 1979). Another level difference between AC and BC stimulation is found in evoked acoustic emissions. With BC-evoked otoacoustic emissions, the increase in emission voltage with increase in stimulus level is much higher than with AC stimulation (Rossi et al., 1988). Here, the reasoning is that BC sounds reach the inner ear by different routes that may affect the otoacoustic emission generation differently than AC sounds. These two reports are further indications that there can be different level responses of AC and BC stimulation.

In a study comparing loudness estimation, Keidser et al. (2000) reported a 10 dB difference when the ear canal was open compared with it occluded. They measured the ear canal sound pressure with a probe microphone in front of the eardrum. With this setup, they measured a 10 dB higher sound pressure level with the ear occluded than when the ear canal was open while the subject conducted an equal loudness task for octave band noise centered around 0.5 and 1 kHz. No explanation for this difference was given but several possible phenomena that can influence the loudness estimation were listed. One possible explanation was the enhanced low frequency effect with BC sounds when the ear was occluded; however, there was no evidence in their study for a large BC component when the ear was occluded (they used AC stimulation). They rejected contraction of the stapedius muscle as the explanation since there was no evidence that an occluded ear activated the muscle and an open ear did not. Also, the tensor tympani was rejected as the origin of the difference since it is mainly activated by tactile stimulation. It can be concluded that loudness estimation can give puzzling results and its origin, as seen in their as well as in our investigation, is difficult to explain.

A difference in loudness function with AC and BC sounds was found in this study when loudness levels for the frequencies 0.25–4 kHz and over a dynamic range of 30–80 dB HL were investigated. The difference is most pronounced in the low frequencies but exists for all frequencies; it is not influenced by a sensorineural hearing loss. The results seen are probably a product of the following:

1. Distortion from the bone transducer spreads the sound energy over a wider frequency range and gives rise to a higher perceived loudness level at the low frequencies.
2. Vibrotactile stimulation from the bone transducers at the low frequencies giving rise to a multi-modal stimulation increasing the perceived loudness level.
3. Differences in the transmission paths affected by the stimulation level, e.g. contraction of the stapedius muscle or contraction of the tensor tympani.

References

Bárány, E., 1938. A contribution to the physiology of bone conduction. Acta Otolaryngol. Suppl. 26.

Brinkmann, K., Richter, U., 1983. Determination of the normal threshold of hearing by bone conduction using different types of bone vibrators. Part I. Audiol. Akust. (Germany) 22, 62–85.

Buus, S., Florentine, M., Poulsen, T., 1998. Temporal integration of loudness, loudness discrimination, and the form of the loudness function. J. Acoust. Soc. Am. 101, 669–680.

Carlsson, P., Håkansson, B., 1997. The bone-anchored hearing aid: Reference quantities and functional gain. Ear Hear. 18, 34–41.

Carlsson, P., Håkansson, B., Ringdahl, A., 1995. Force threshold for hearing by direct bone conduction. J. Acoust. Soc. Am. 97, 1124–1129.

Corliss, E.L.R., Coidan, W., 1955. Mechanical impedance of the forehead and mastoid. J. Acoust. Soc. Am. 27, 1164–1172.

Fletcher, H., Munson, W.A., 1933. Loudness, its definition, measurement and calculation. J. Acoust. Soc. Am. 5, 82–108.

Florentine, M., Buus, S., Robinson, M., 1998. Temporal integration of loudness under partial masking. J. Acoust. Soc. Am. 104, 999–1007.

Flottorp, G., Solberg, S., 1976. Mechanical impedance of human headbones (forehead and mastoid portion of temporal bone) measured under ISO/IEC conditions. J. Acoust. Soc. Am. 59, 899–906.

Flottorp, G., Djupesland, G., Winther, F., 1971. The acoustic stapedious reflex in relation to critical bandwidth. J. Acoust. Soc. Am. 49, 457–461.

Håkansson, B., Carlsson, P., 1989. Skull simulator for direct bone conduction hearing devices. Scand. Audiol. 18, 91–98.

Håkansson, B., Carlsson, P., Tjellström, A., 1986. The mechanical point impedance of the human head, with and without skin penetration. J. Acoust. Soc. Am. 80, 1065–1075.

Håkansson, B., Brandt, A., Carlsson, P., Tjellström, A., 1994. Resonance frequency of the human skull in vivo. J. Acoust. Soc. Am. 95, 1474–1481.

Håkansson, B., Carlsson, P., Brandt, A., Stenfelt, S., 1996. Linearity of sound transmission through the human skull in vivo. J. Acoust. Soc. Am. 99, 2239–2243.

Hayes, D., Chen, J., 1998. Bone-conduction amplification with completely in-the-canal hearing aids. J. Am. Acad. Audiol. 9, 59–66.

Keidser, G., Katsch, R., Dillion, H., Grant, F., 2000. Relative loudness perception of low and high frequency sounds in the open and occluded ear. J. Acoust. Soc. Am. 107, 3351–3357.

Khalil, T.B., Viano, D.C., Smith, D.L., 1979. Experimental analysis of the vibrational characteristics of the human skull. J. Sound Vib. 63, 351–376.

Khanna, S.M., Decraemer, W.F., 1996. Vibrations modes and the middle ear function. In: Hüttenbrink, K.-B. (Ed.), Middle Ear Mechanics in Research and Otosurgery. UniMedia GmbH, Dresden, pp. 21–26.

Khanna, S.M., Tonndorf, J., Queller, J., 1976. Mechanical parameters of hearing by bone conduction. J. Acoust. Soc. Am. 60, 139–154.

Kirikae, I., 1959. An experimental study on the fundamental mechanism of bone conduction. Acta Otolaryngol. Suppl. 145.

Lenhardt, M., Skellett, R., Wang, P., Clarke, A., 1991. Human ultrasonic speech perception. Science 253, 82–85.

Mauldin, L., Jerger, J., 1979. Auditory brainstem evoked responses to bone-conducted signals. Arch. Otolaryngol. 105, 656–661.

Merchant, S., Rovicz, M., Rosowski, J., 1996. Acoustic input impedance of the stapes and cochlea in human temporal bones. Hear. Res. 97, 30–45.

Neuman, J., Uppenkamp, S., Kollmeier, B., 1996. Detection of the acoustic reflex below 80 dB HL. Audiol. Neurootol. 1, 354–369.

Nolan, M., Lyon, D.J., 1981. Transcranial attenuation in bone conduction audiometry. J. Laryngol. Otol. 95, 597–608.

Northern, J., Gabbard, S., 1994. The acoustic reflex. In: Katz, J. (Ed.), Handbook of Clinical Audiology, 4th edn. Lippincott Williams and Wilkins, Philadelphia, PA, pp. 300–316.

Purcell, D., Kunov, H., Madsen, P., Cleghorn, W., 1998. Distortion product otoacoustic emissions stimulated through bone conduction. Ear Hear. 19, 362–370.

Purcell, D., Kunov, H., Cleghorn, W., 1999. Objective calibration of bone conductors using otoacoustic emissions. Ear Hear. 20, 375–392.

Rossi, G., Solero, P., Rolando, M., Olina, M., 1988. Delayed otoacoustic emissions evoked by bone-condition stimulation: experimental data on their origin, characteristics and transfer to the external ear in man. Scand. Audiol. Suppl. 29, 1–24.

Scharf, B., 1959. Critical bands and the loudness of complex sounds near threshold. J. Acoust. Soc. Am. 31, 365–370.

Scharf, B., 1969. Dichotic summation of loudness. J. Acoust. Soc. Am. 45, 1193–1205.

Schratzenstaller, B., Janssen, T., Alexiou, C., Arnold, W., 2000. Confirmation of G. von Bekesy's theory of paradoxical wave propagation along the cochlear partition by means of bone-conducted auditory brainstem responses. J. Oto Rhino Laryngol. 62, 1–8.

Smith, J.B., Suggs, C.W., 1976. Dynamic properties of the human head. J. Sound Vib. 48, 35–43.

Snik, A., Mylanus, E., Cremers, C., 1995. The bone-anchored hearing aid compared with conventional hearing aids. Otolaryngol. Clin. N. Am. 28, 73–83.

Stenfelt, S., Håkansson, B., 1999. Sensitivity to bone-conducted sound: Excitation of the mastoid vs the teeth. Scand. Audiol. 28, 190–198.

Stenfelt, S., Håkansson, B., 2000. A bone-anchored hearing aid for patients with pure sensorineural hearing impairment: A pilot study. Scand. Audiol. 29, 175–185.

Stevens, S.S., 1956. Calculation of the loudness of complex noise. J. Acoust. Soc. Am. 28, 807–832.

Stevens, S.S., 1975. Psychophysics: Introduction to its Perceptual, Neural, and Social Prospects. Wiley, New York.

Tjellström, A., Håkansson, B., Granström, G., 2001. Bone-anchored hearing aids. Current status in adults and children. Otolaryngol. Clin. N. Am. 34, 337–363.
Tonndorf, J., 1957. Fluid motion in cochlear models. J. Acoust. Soc. Am. 29, 558–568.
Tonndorf, J., 1962. Compressional bone conduction in cochlear models. J. Acoust. Soc. Am. 34, 1127–1131.
Tonndorf, J., 1966. Bone conduction. Studies in experimental animals. Acta Otolaryngol. Suppl. 213.
von Békésy, G., 1932. Zur theorie des hörens bei der schallaufnahme durch knochenleitung. Ann. Phys. 13, 111–136.
von Békésy, G., 1955. Paradoxial direction of wave travel along the cochlear partition. J. Acoust. Soc. Am. 27, 137–145.
von Békésy, G., 1960. Experiments in Hearing. McGraw-Hill, New York.
Wever, E.G., Bray, C.W., 1936. The nature of bone conduction as shown in the electrical response of the cochlea. Ann. Otol. Rhinol. Laryngol. 45, 822–830.
Wever, E.G., Lawrence, M., 1954. Physiological Acoustics. Princeton University Press, Princeton, NJ.
Zwislocki, J., 1953. Wave motion in the cochlea caused by bone conduction. J. Acoust. Soc. Am. 25, 986–989.