

# The balanced electromagnetic separation transducer: A new bone conduction transducer

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Conventional bone conduction transducers, which are relatively large, suffer from poor performance at low frequencies. A new type of electro-dynamic transducer, the balanced electromagnetic separation transducer (BEST), was developed to improve the performance of the conventional transducers. By using a balanced suspension principle, the quadratic distortion forces, as well as the static forces between the vibrating parts, are principally counterbalanced. Both the distortion and the size of the transducer can therefore be considerably reduced. Moreover, the static and dynamic magnetic fluxes are separated, except in the air gap regions, giving a more efficient transducer. For example, in comparison with a conventional B71 transducer, a prototype of the BEST has: Lower total harmonic distortion (THD), by 20–25 dB, and improved sensitivity by 10–20 dB for 100 to 1000 Hz and by 2–10 dB for 1 to 10 kHz. From a clinical point of view, the BEST offers a chance to measure bone thresholds, at 250 and 500 Hz, which are reliable at hearing levels not possible before. For example, at 250 Hz the BEST has 23 dB higher sensitivity than the B71; the THD is improved from 61% (B71) to 3.3% (BEST) at 40 dB HL (ISO 389-3, 1994). © 2003 Acoustical Society of America. [DOI: 10.1121/1.1536633]

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## I. INTRODUCTION

Hearing by bone conduction as a phenomenon, i.e., hearing sensitivity to vibrations induced directly or via skin or teeth to the skull bone, has been known since the 19th century. The interest in bone conduction was initially based on its usefulness as a diagnostic tool. In particular, it is used in hearing threshold testing to determine the sensorineural hearing loss or, indirectly, to determine the degree of conduction hearing loss by noting the difference between the air and the bone thresholds. A comprehensive description of bone conduction threshold testing can be found in Dirks (1994).

In recent decades the use of bone conduction hearing aids has increased. One reason for this increase is the promising results achieved with the bone-anchored hearing aid (BAHA). The BAHA uses a percutaneous titanium implant system to attach the transducer directly to the skull behind the ear. The principal design of the BAHA system is extensively presented by Tjellström *et al.* (2001).

Although bone conduction transducers are frequently used in hearing threshold testing and hearing aids, there is still a great need for improvements. A bone conduction transducer has the purpose to efficiently transform the electrical signal energy into mechanical vibratory energy without distorting the original (information carrying) electrical signal. Deterioration of the electrical signal can occur in the electro-mechanical conversion caused by nonlinearities; nonlinear distortion products then contaminate the mechanical vibratory output. Low distortion of these transducers is a prerequisite in obtaining reliable bone conduction threshold data and also to achieve high sound quality in bone conduction

hearing aids. Erroneous hearing threshold data, caused by distortion in the transducer, could lead to an incorrect diagnosis of a patient's hearing etiology and jeopardize subsequent hearing aid fitting procedures.

It is well known that conventional audiometric transducers, as well as transducers for bone conduction hearing aids, have some drawbacks associated with their generic design. These drawbacks arise because the force generated in the air gap(s) is approximately proportional to the magnetic flux squared. To make such transducers usable as a hearing transducer, a static magnetic flux must be superimposed on the signal to obtain an acceptable, although not perfect, linear behavior. This static magnetic flux gives the transducer a bias point in the magnetic flux to force characteristics. When the signal flux is operating around this bias point, along the slightly bent quadratic curve, nonlinear distortion is generated, in particular at low frequencies and high signal levels. Moreover, in bone conduction hearing aids, poor efficiency and large size are other limiting factors that originate from the generic design of conventional bone conduction transducers.

In this study a new transducer design, which simultaneously meets the demands of high linearity, high efficiency, and small size, is presented. The new design is called the balanced electromagnetic separation transducer (BEST). Its performance in terms of frequency response and total harmonic distortion is compared with the widely used conventional audiometric bone conduction transducer, the B71, from Radioear Corporation, in the USA.

## II. PRINCIPLE DESIGN

### A. Background

Transducers for bone conduction sound generation can be constructed by different technologies, such as electro-

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dynamic, piezoelectric, or magnetostrictive. However, piezoelectric and magnetostrictive transducers are often considered inconvenient and are not commercially available, basically because of poor low-frequency response.

Electrodynamic transducers of the moving coil type, which are the most common in ordinary loudspeakers, are not considered desirable in hearing aid applications; their electrical impedance is too low. The low impedance is a consequence of that the driving voice coil: (1) is suspended to the sound radiating diaphragm and therefore should have a low weight; and (2) it should have a small radial (bundle) thickness to allow a high transversal magnetic flux density.

Electromagnetic transducers of the variable reluctance type combine properties such as small size, wide frequency range, high impedance, and efficient energy transformation; hence, they are almost exclusively used in hearing aid applications. In simple terms variable reluctance type transducers function according to the horseshoe magnet principle where there is a small air gap between the armature (basically the permanent magnet) and the yoke. By superimposing a signal magnetic flux (generated by a coil whose dimensions are not so critical) the force in the air gap, between the yoke and the armature, will vary accordingly. A more detailed description of present designs is presented below.

In hearing threshold testing, the B71 is today the most frequently used bone conduction transducer. A similar design is used in the BAHA system, in which the most conspicuous difference is that the internal suspension spring system has inherent damping (Håkansson *et al.*, 1990; Håkansson and Carlsson, 1985). Both of these transducers, the B71 and the BAHA, work according to the variable reluctance principle. However, even if these designs are well functioning there are some generic shortcomings in the conventional variable reluctance type transducers: these are related to poor frequency response and high level of distortion at low frequencies. The new balanced electromagnetic separation transducer (BEST) offers a significant improvement, especially in the application of bone conduction hearing threshold testing. The technical design differences between the B71 and the BEST are described below.

It should be noted that in transducers for bone conduction threshold testing, the consumption of current is of minor importance since they are driven by audiometers that are connected to a power line supply. This is probably the reason that there is a moving coil design as an alternative to the B71 for bone conduction threshold testing: the KH70 from Grahner Präcitronic GmbH, Germany. The KH70 has a better linearity and radiates less aerial sound than the B71. On the other hand, the KH70 has some disadvantages, such as a bulkiness (the weight is five times and its height is 2.5 times that of the B71); as a consequence it is more difficult to attach it to the skull without it touching the external ear (the pinnae). This is probably the reason that the KH70 is quite uncommon in the audiology clinics in most countries, and it is not dealt with in this paper.

## B. The conventional design—the B71

The B71 transducer has a plastic housing with a 1.75-cm<sup>2</sup> circular attachment surface toward the head, as

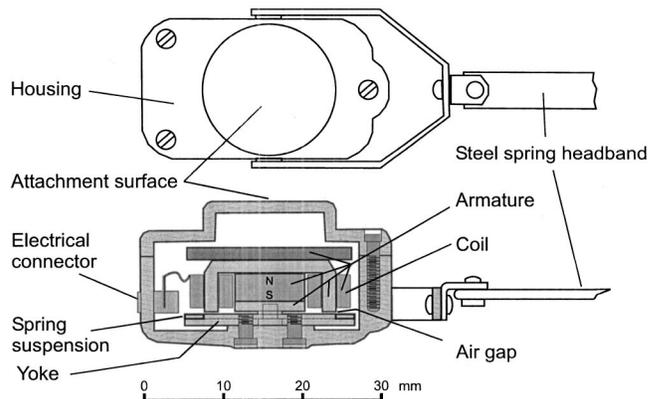


FIG. 1. External view and a cross-sectional sketch of the B71 transducer.

shown in Fig. 1. With a steel-spring headband, the transducer is pressed with a total force of approximately 5–6 Newton against the mastoid area behind the ear. The transducer consists of an armature, a yoke, and a small but essential air gap which disrupts the magnetic flux path. The magnetic flux is composed of the static flux generated by the permanent magnet and the dynamic flux generated by the current in two coils. The total weight of the B71 is 19.9 g. In the cross section in Fig. 1, the spring suspension that connects the yoke to the armature, thereby maintaining the essential air gap between them, cannot be shown in full.

Some details of the static and dynamic magnetic circuits of the B71 are shown in Fig. 2. According to a first-order analysis of the electromagnetic circuit, assuming that the leakage of magnetic flux is negligible and assuming that the dependence between the magnetic field and the magnetic flux in the air gap is linear, it can be shown that the total force between the yoke and the armature is approximately proportional to the total magnetic flux squared.

$$F_{\text{tot}} \propto (\phi_0 + \phi_{\sim})^2 = \phi_0^2 + 2 \cdot \phi_{\sim} \cdot \phi_0 + \phi_{\sim}^2, \quad (1)$$

where  $\phi_0^2$  represents the static force from the permanent magnet,  $2 \cdot \phi_{\sim} \cdot \phi_0$  represents the desired signal force, and  $\phi_{\sim}^2$  represents an undesired distortion force.

From Eq. (1) it is clear that in order to achieve a reasonably good linearity in the electro-mechanic transformation, a high static magnetic flux is required. By introducing a high static (biasing) magnetic flux  $\phi_0$  such that  $\phi_0 \gg \phi_{\sim}$ , it is possible to achieve a fairly linear behavior. However, when  $\phi_0$  is increased, the static force also increases, and the suspension spring compliance  $C$  (symbolically depicted in Fig.

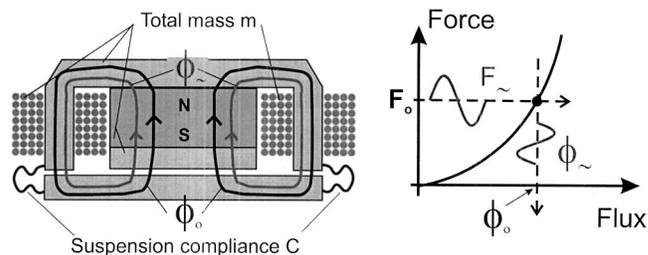


FIG. 2. Close-up of the magnetic circuit of the B71 and a corresponding flux to force characteristic.

2) must therefore be stiffer to avoid a collapse of the air gap. The consequences of a stiffer  $C$  (smaller value of  $C$ ) will be described in what follows.

The seismic mass  $m$ , that serves as a counterweight, consists of a coil, a magnet, a soft iron armature, and an additional mass body (not shown in Fig. 2). This counterweight is essential in these designs for two reasons. First, if the counterweight  $m$  is very small, this means that very little of the force generated in the air gap will be transmitted to a vibration fed into the skull. Second, the counterweight mass  $m$  interacts with the suspension compliance  $C$  and generates a resonance frequency  $f_r$ ,

$$f_r \approx 1/(2\pi\sqrt{m \cdot C}) \text{ Hz.} \quad (2)$$

A transducer is normally designed so that the resonance frequency falls at a frequency slightly above the lowest frequency of interest, i.e., in the range of 500–1000 Hz for a BAHA and in the range of 250–500 Hz for a transducer used in bone conduction hearing threshold testing. This resonance gives an essential boost at low frequencies in variable reluctance transducers.

With this background some drawbacks of conventional variable reluctance type transducers can be pointed out. The first drawback is evident from Eq. (2): a low  $f_r$  requires a large mass  $m$  for a given compliance  $C$ . If the low-frequency response has to be improved and  $C$  cannot be more compliant (to prevent air gap collapse), the counterweight mass must be increased in spite of the undesired greater weight and size.

Moreover, a second drawback of the conventional variable reluctance transducer is that the magnetic circuit is designed so the magnetic signal flux follows or coincides with the static magnetic flux (Fig. 2). As a consequence, the properties of the electrodynamic conversion suffer since high quality permanent magnets have inherently poor properties for conducting the dynamic magnetic flux.

A third drawback of a conventional variable reluctance transducer is the vulnerability related to the existence of the static force,  $F_0$  (Fig. 2). This force, generated by the permanent magnet, strives continuously to collapse the essential air gap between the yoke and the armature. This is critical, as this magnetic contraction force grows as the reciprocal of air gap length as air gap length decreases, whereas the counteracting suspension spring force grows linearly. A stable balance between the permanent magnet force and the force from the suspension spring is a prerequisite to maintaining this air gap. Furthermore, aging of the spring or severe external force/impact on the transducer can adversely affect the spring suspension. This is quite common in bone conduction hearing aids where such external forces can deform the spring suspension and the air gap might collapse partially or fully. If the air gap collapses, the sound from the transducer is severely distorted; the transducer must be repaired.

Finally, and perhaps most critically, since the total force output is approximately proportional to the total magnetic flux squared, a high level of primarily second harmonic distortion is generated. At low frequencies and at high signal levels, this distortion severely limits the use of conventional variable reluctance transducers in bone conduction hearing

threshold testing, especially at 250 and 500 Hz. For example, a patient being tested at 250 Hz may hear the second harmonic at 500 Hz instead. This is especially serious, as the normal-hearing threshold is better by 9 dB at 500 Hz than at 250 Hz (ISO 389-3, 1994). This distortion, described in several studies, severely limits the use of the B71 transducer at lower frequencies (Dirks and Kamm, 1975; Dolan and Morris, 1990; Parving and Elberling, 1982).

It is understandable that the second harmonic distortion of the B71 is high. The primary reason for this, as mentioned, is the quadratic flux dependence, but a contributing factor is that the resonance frequency is located near 500 Hz. This means that the second harmonic component, generated in the electromagnetic conversion, is further enhanced by some 12 dB when the fundamental frequency is 250 Hz (12 dB is the approximate difference in the frequency response of the B71 between 250 and 500 Hz). To overcome this, the Radioear Company has developed another transducer model, the B72, which has a lower resonance frequency. This was mainly achieved by increasing the mass and, as a consequence, also the size. Larger size implies not only that it is more difficult to attach the transducer to the skull, but also that more airborne sound is radiated. The airborne sound may then be heard as an aerial sound via normal air conduction, instead of via bone conduction, and the result will be misleading (Lightfoot and Hughes, 1993). Due to these drawbacks the B72 is not widely used (at least not to the author's knowledge).

### C. The balanced electromagnetic separation transducer (BEST)

It is clear that in the optimization procedure of a conventional variable reluctance type transducer some aspects are counterproductive. To achieve low distortion, a high static flux is needed: a high static flux requires a stiff suspension, and finally, a stiff suspension requires a heavy counterweight mass to avoid reducing the low-frequency response. This means that a design having low distortion often has, as consequence, a poor low-frequency response. The opposite is also true, i.e., a design with a good low-frequency response suffers from high distortion. The basic idea of the new transducer design (BEST) is to counterbalance or cancel the static forces, thus avoiding the strong requirement of a stiff spring suspension and a high mass to get an acceptably low distortion and good low-frequency response.

This counterbalance of the static forces is accomplished by introducing a second, opposing, air gap. A solution to implementing two opposing air gaps with only one magnet is shown in Fig. 3. The transducer in this figure has circular symmetry; it was the first attempt to implement the principles of the BEST.

It is shown here that the static forces of the upper and lower air gaps cancel or counterbalance each other, as they are of equal magnitude but act in opposite directions. The signal forces are generated in a push–pull fashion as follows. At one instance, the total magnetic flux is decreased from the static value in the upper air gap  $[(\phi_0/2) - \phi]$ , while it is

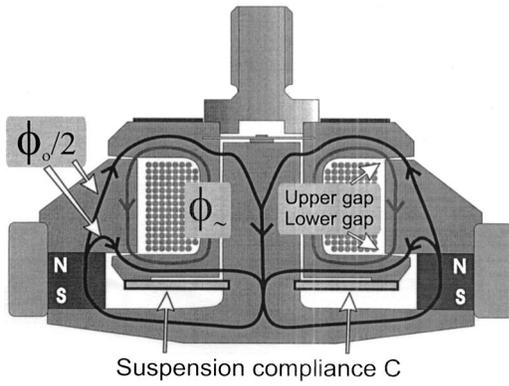


FIG. 3. Cross-sectional view of the BEST, in which one circular magnet and two opposing air gaps are used.

simultaneously increased by the same amount in the lower air gap  $[(\phi_0/2) + \phi_-]$ .

Some prototypes were built and it was proved that the idea underlying the present design worked. This implementation is extremely compact, which might be useful in an implantable bone conduction hearing aid where size is of the utmost importance. However, in these prototypes it was difficult to find the balanced position; also, the components had to be extremely precise, especially the suspension spring system.

A modified design of the BEST, especially adapted for bone conduction hearing threshold testing where size and power consumption is not so critical as in an implantable bone conduction hearing aid, was therefore developed (Fig. 4). This design is slightly larger, as it uses four magnets instead of one. Also, in addition to the two opposing internal upper and lower air gaps in the magnetic signal path, there are two external air gaps that affect only the static magnetic path. These additional external air gaps help to stabilize the static magnetic flux, i.e., to make it independent of the position of the bobbin arms in the air gaps. If the leakage of static magnetic flux is assumed negligible, the total air gap length for the static flux passing through the external and internal air gaps is constant and independent of the actual position of the bobbin relative to the yoke. Hence, it is much easier to maintain the balanced position, with the equal air

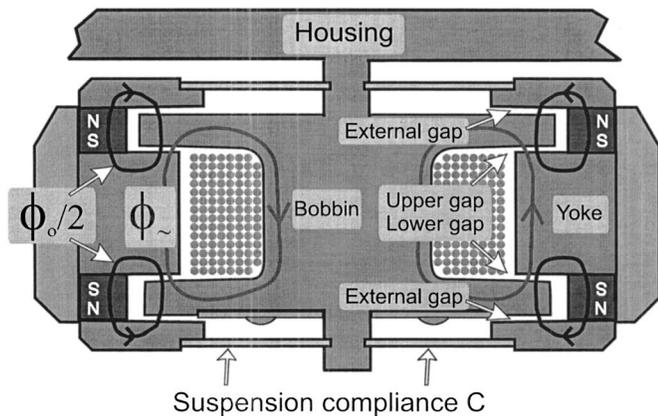


FIG. 4. Cross-sectional view of the BEST, in which four magnets and four internal and four external air gaps are used.

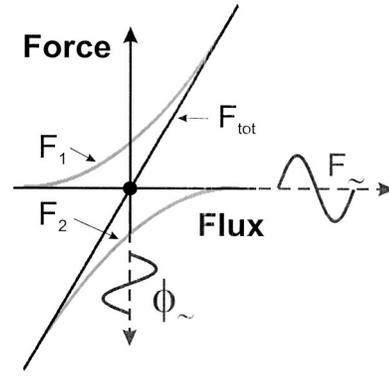


FIG. 5. The force versus magnetic flux characteristics of the BEST.

gaps, as compared with the design presented in Fig. 2. The BEST prototype used in all of the measurements presented in this study were, in principle, of this design.

Since the upper and lower magnets (Fig. 4) each generate the static magnetic flux  $\phi_0/2$ , the forces  $F_1$  (lower gaps) and  $F_2$  (upper gaps) on the bobbin, with reference to the yoke, are approximately following the equations:

$$F_1 \propto \left( \frac{\phi_0}{2} + \phi_- \right)^2 = \frac{\phi_0^2}{4} + \phi_- \cdot \phi_0 + \phi_-^2, \quad (3)$$

$$F_2 \propto \left( \frac{\phi_0}{2} - \phi_- \right)^2 = \frac{\phi_0^2}{4} - \phi_- \cdot \phi_0 + \phi_-^2. \quad (4)$$

The total force  $F_{\text{tot}}$  can be obtained by subtracting  $F_1$  from  $F_2$ , which clearly shows that not only the static forces, but also the quadratic distortion forces, are cancelled or counter-balanced.

$$F_{\text{tot}} = F_1 - F_2 = F_{\text{signal}} \propto 2 \cdot \phi_0 \cdot \phi_- \quad (5)$$

That the quadratic magnetic flux to force characteristic of the upper and lower air gaps appear as a linear curve after summation is also illustrated in Fig. 5. It should be noted in Fig. 5 that there is no remaining static force between the bobbin and the armature in the balanced position, as is also shown by Eq. (5). It is important to point out that the ideal behavior described by Eq. (5) requires that the assembly has perfect symmetries; i.e., equal air gaps, equal magnetization of magnets, bobbin exactly centered, etc. In a final implementation the degree of asymmetry will determine the amount of harmonic distortion and static force imbalance.

In addition to the principle of balanced suspension, there is another important feature of the BEST. The static and the dynamic magnetic fluxes are separated except in the vicinity of the air gaps where they must be superimposed. This separation of the static and dynamic magnetic fluxes (except in the air gap regions) has two specific advantages. First, the dynamic flux does not need to pass the permanent magnet, which generally has very poor dynamic properties. Second, as seen in Eq. (5), a high static flux is important in obtaining a high gain factor in the electro-mechanical conversion. However, a high static flux can result in local saturation of the soft iron material in the signal flux path. If there is a local saturation somewhere in the soft iron material, the dynamic

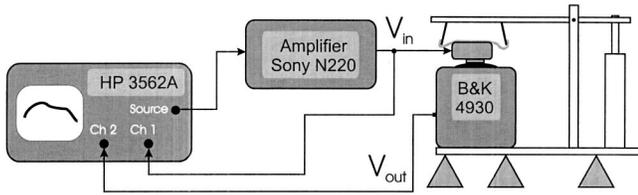


FIG. 6. Measurement setup for the frequency response and distortion measurements.

properties may deteriorate significantly. This problem is minimized in the BEST design, since the static and dynamic fluxes are separated.

Finally, it should be noted that there are some further minor design differences between the B71 and the BEST. To give the resonance of the BEST a better-controlled shape, some internal damping of the mechanical system is used. Furthermore, to improve efficiency and to reduce heat generation, the magnetic circuit is designed to minimize the magnetic losses.

In summary, the *balanced* suspension principle and the *separation* of the fluxes are key features of the new transducer shown in its name, the balanced electromagnetic separation transducer. An international patent application of this new transducer principle has been approved (Håkansson, 2001).

Finally, it should also be pointed out that Hunt (1982) described the general principles of a “balanced armature” transducer design. This principle design has also been successfully used in subminiature sound transducers for hearing aids; see, for example, Sebesta and Carlisle (1969).

### III. METHODS

The setup that was used in the frequency response and distortion measurements is shown in Fig. 6. The equipment used includes: a two-channel signal analyzer (Hewlett Packard 3562A), a power amplifier (Sony TA-N220), and an artificial mastoid (Brüel & Kjær type 4930).

The B71 transducer was chosen arbitrarily from one of the audiometry test rooms at Sahlgren hospital (Göteborg, Sweden). The BEST was manufactured according to the previous description (see Fig. 4) and was incorporated into the housing of a B71 transducer. Hence, from the outside there was no visible difference between the BEST and the B71, but the weight of the BEST transducer element (without housing) was lighter: 10 g versus 15.3 g.

#### A. Frequency response

In the frequency response measurement, a swept sinusoid from 0.1 to 10 kHz with peak amplitude of 100 mV was used. The frequency response function  $G_{\text{measured}}(j\omega)$  was calculated with the Hewlett Packard 3562A as

$$G_{\text{measured}}(j\omega) = G_{12}/G_{11}, \quad (6)$$

where  $G_{11}$  is the power spectrum of the input voltage to the transducer, and  $G_{12}$  is the cross-power spectrum between output voltage from the artificial mastoid and the input voltage to the transducer.

The artificial mastoid has a bone conductor attachment area or bilaminar rubber pad simulating skin and subcutaneous tissue properties. As the force gauge of the artificial mastoid is located below the bilaminar rubber pad, some corrections have to be made. To calculate the true frequency response function (force output/voltage input) of a transducer attached to the artificial mastoid, the measured frequency response function was corrected for the transmission through the rubber pad, and for the force to voltage sensitivity of the force gauge of the artificial mastoid. This correction required the frequency response function  $A(j\omega)$  calculated as the output voltage of the artificial mastoid divided by the input force level to the surface of the rubber pad; and it is also required a measurement of the mechanical impedance (force/velocity) of the artificial mastoid rubber pad  $Z(j\omega)$ . The mechanical impedance  $Z(j\omega)$  is needed to be compensated for the mass  $m \approx 1.1$  g over the force gauge in the impedance head Brüel & Kjær 8000 which was used to measure  $A(j\omega)$ . The total response  $G_{\text{real}}(j\omega)$ , that is the output force level at the pad of the artificial mastoid divided by input voltage to the transducer, was then calculated as

$$G_{\text{real}}(j\omega) = G_{\text{measured}}(j\omega) \cdot \frac{Z(j\omega)}{Z(j\omega) + j\omega m} \cdot \frac{1}{A(j\omega)}. \quad (7)$$

#### B. Distortion

In the measurements of nonlinear distortion, a sinusoid of fixed frequency at 250, 315, 500, 750, 1000, 1500, 2000, 3000, and 4000 Hz was used. For each frequency the power spectrum of the input voltage and output force were measured and the Hewlett Packard 3562A calculated the total harmonic distortion (THD), where the harmonics up to 12 kHz was taken into account

$$\text{THD} = \frac{\text{Total harmonic power}}{\text{Fundamental power}} \cdot 100\%. \quad (8)$$

#### C. Electrical input impedance

The electrical input impedance,  $Z_{\text{in}}$ , of both transducers was also measured, using the Brüel & Kjær artificial mastoid type 4930 as the load. By using a resistor  $R = 10 \Omega$  in series with the transducer and measuring the voltage at both the source side ( $V_S$ ) and the transducer side ( $V_T$ ) of the resistor,  $R$ , the impedance,  $Z_{\text{in}}$ , was calculated from a swept sinusoid measurement from 100–10k Hz

$$Z_{\text{in}} = \frac{V_T}{V_S - V_T} \cdot R = \frac{V_T}{V_S} \cdot \frac{1}{1 - \frac{V_T}{V_S}} \cdot R, \quad (9)$$

where  $V_T/V_S$  was the measured frequency response function.

### IV. RESULTS

#### A. Frequency response

The frequency response functions of the B71 and the BEST are shown in Fig. 7 as the magnitude of  $G(j\omega)$  in dB. From this figure, it is clear that the resonance frequency is

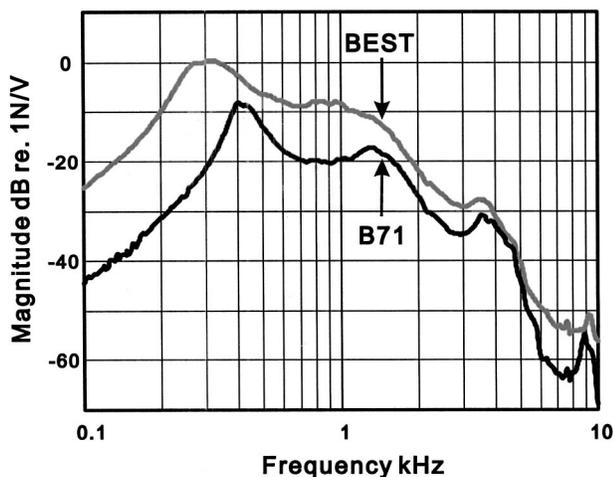


FIG. 7. Magnitude of the frequency response function  $G(j\omega)$  of the B71 and the BEST.

reduced from approximately 420 Hz for the B71 to approximately 300 Hz for the BEST. Note that this decrease by a factor of 1.4 corresponds to a change by the same factor squared, in terms of mass, according to Eq. (2). As the BEST is actually lighter than the B71 by a factor of 1.53, the total mass factor is approximately threefold. This means that the BEST can be designed, at least theoretically, with a counterweight that is 3 times lighter than that of the B71 and still have the same resonance frequency.

Below 1000 Hz, the sensitivity is improved by 10–20 dB, except near the resonance frequency of the B71, where it is approximately 5 dB. Above 1000 Hz, the sensitivity is improved by 2–10 dB.

## B. Distortion

The distortion measurements were made frequency by frequency, by measuring the force output power spectra of the B71 and the BEST, as for example at 250 Hz (fundamental frequency) where the spectra obtained are shown in Fig. 8. At this particular frequency the output force level was 40 dB HL (107 dB re: 1  $\mu$ N). The THD was obtained by adding

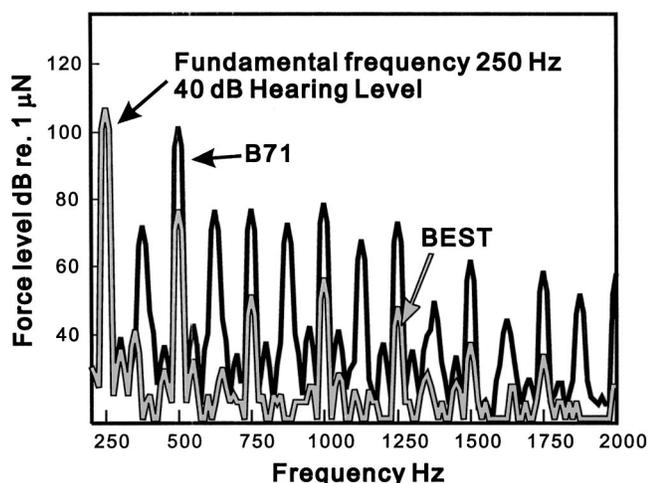


FIG. 8. The power spectrum of the output force when the fundamental frequency of the input signal is 250 Hz.

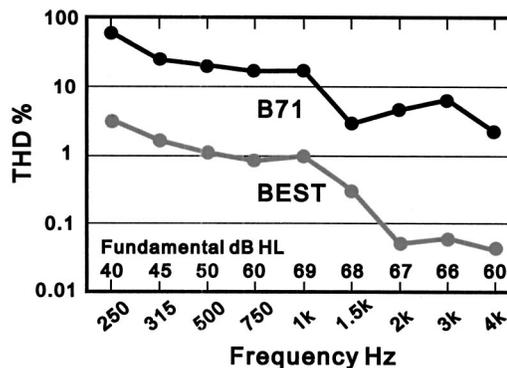


FIG. 9. The total harmonic distortion (THD) of the B71 and the BEST.

the power of all the harmonics (all peaks at multiples of 250 Hz) divided by the power of the fundamental frequency (250 Hz) according to Eq. (8). At 250 Hz, the THD was found to be 61% for the B71 and 3.3% for the BEST.

The major contribution to the THD at 250 Hz for the B71 and the BEST is from the second harmonic peak at 500 Hz, as is clear in Fig. 8. A peculiar phenomenon, appearing for the B71 only, is that additional peaks between the harmonics, i.e., at 375, 625 Hz, etc. can be easily seen. These peaks between the harmonics were present only in the measurement with the fundamental frequency of 250 Hz (Fig. 8). Also found was a small peak at the subfundamental frequency 125 Hz (not shown in Fig. 8). These nonharmonic peaks are obviously not intermodulation components between the harmonic frequencies and the power line frequency (e.g.,  $250 \pm 50 \text{ Hz} \neq 125$  or  $375 \text{ Hz}$ ). No explanation for this phenomenon has been found so far.

The THDs for all fundamental frequencies tested are presented in Fig. 9. The level of the fundamental (dB hearing level) for each test frequency is shown by the figure above the frequency scale. Although it was originally planned to use the minimum output level requirements for type 1 audiometers (IEC 645-1, 1992), small changes were made at 250 and 500 Hz: a lower level was used to avoid overloading the B71 transducer. At 315 Hz no particular level is specified in the standard but, for the other frequencies, the specified level was used, although some of them were corrected for the transmission through the rubber pad of the artificial mastoid afterwards (at 1000–3000 Hz a reduction of 1–4 dB was used).

From Fig. 9 it can be seen that the THD in percent is more than 10 times lower (reduction by 20–25 dB) for the BEST than for the B71. For verification purposes, the THD of the input voltage fed to the transducers was also measured. It was found that this input THD was lower by a factor of 10 than at the output, for all frequencies. Hence, it can be concluded that all of the distortion presented in Fig. 9 is generated in the transducer.

## C. Electrical input impedance

The magnitude of the electrical input impedances of the B71 and the BEST is shown in Fig. 10. At low frequencies the impedances are resistive and mainly determined by the ohmic losses in the coil wires. The dc impedances of the

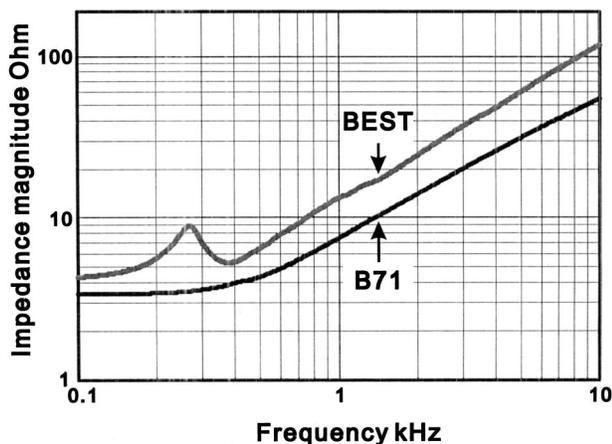


FIG. 10. The magnitude of the electrical input impedance of the B71 and the BEST.

BEST and the B71 are 4.0 and 3.3 ohms, respectively. At 1000 Hz the impedance magnitude of the BEST is some 1.8 times higher than that of the B71; above 1000 Hz, it is slightly more and below it is slightly less.

It is interesting to note that although the BEST has a generally higher sensitivity (Fig. 7), it consumes less current for a given voltage input. Hence, the BEST design is more efficient than the B71 by a factor that may be determined from the level difference in sensitivity in Fig. 7 ( $BEST_{dB}$  minus  $B71_{dB}$ , where dB is defined as  $20 \times \log[\text{force}/\text{voltage}]$ ), plus a level correction for impedance from Fig. 10 that is calculated as  $10 \times \log[\text{impedance magnitude ratio BEST/B71}]$ . Although this improvement is of moderate importance in audiometric applications, where the audiometer is powered from a power line, in hearing aids it is a dramatic one. Also, under some conditions, the thermal heating of the transducer element is crucial; this is lower with the more efficient BEST design.

## V. DISCUSSION

Even if the improvements presented in Figs. 7–10 look promising, there are other requirements that have to be fulfilled to assure that the BEST is a successful new design. First, the transducer must be reliable over time. Although this has not yet been proven, one can assume that the balanced design, with static forces between moving parts counterbalanced, has the potential to be as reliable as the conventional designs. Furthermore, the coil in the BEST design is placed at the side opposite to that of the counterweight mass, in contrast to in the conventional design. The reason for doing so is that the counterweight side of the transducer is the most vibrant one. Consequently, there is a risk of fatigue in the electrical connection wires, if the coil is not rigidly attached to the driving side (skull side) of the transducer.

Second, it must be possible to manufacture the transducer at a reasonable cost and with a sufficient degree of reproducibility. To evaluate these aspects, a larger number of transducers must be manufactured. From the limited experience we have so far, it seems that the BEST is easier to assemble with respect to the air gap stability. In particular, the use of both internal and external air gaps, stabilizing the

static magnetic flux, appears to provide a robust design and individual adjustments were not needed in the assembling of the prototypes.

It should also be noted that the present design is optimized for bone conduction hearing threshold measurements. Specifically, the present design is optimized to produce high output forces at low frequencies. An important aspect taken into account is that the transducer must have large enough air gaps to allow the larger deflections that occur at lower frequencies.

The balanced electromagnetic separation transducer (BEST) can be used in other applications and the optimization may then be different. Model parameters that may be modified are related to: air gap size, coil wires, mechanic and magnetic material properties. For example, designs for hearing threshold testing purposes (present study), implantable bone conduction hearing aids, ear level BAHA, and body-worn BAHA would all be optimized differently.

Planned and suggested future research includes: a clinical evaluation of the BEST for hearing threshold testing, development of better test and calibration methods, and finally, further improvement of the models of the transducer and the calibration equipment. The aim with this bone conduction methodological project is to significantly reduce the prevailing relatively large intra- and intersubject variability as well as the large interclinic variability in bone threshold testing.

In a concurrent project at our department, dealing with bone conduction physiology, attention is directed to understanding the underlying bone conduction mechanisms in an effort to improve the interpretation of bone conduction hearing threshold data.

## VI. CONCLUSION

A new transducer design, the Balanced Electromagnetic Separation Transducer (BEST), is presented and evaluated. The results show that this transducer has the following advantages over the conventional B71 transducer:

- (i) Lower total harmonic distortion (THD) by 20–25 dB;
- (ii) Lower counterweight mass by a factor of 3 for the same resonance frequency;
- (iii) Improved sensitivity by 10–20 dB for 100 to 1000 Hz and by 2–10 dB for 1 to 10 kHz; and
- (iv) Improved efficiency, in addition to improved sensitivity, since the electrical input impedance is higher by some 1.8 times.

The BEST offers the opportunity to measure bone thresholds at 250 and 500 Hz with sufficiently good accuracy at hearing levels not possible before. For example, at 250 Hz the BEST has 23 dB higher sensitivity than the B71, and the THD is improved from 61% (B71) to 3.3% (BEST) at 40 dB HL for the transducers used in this study.

A clinical evaluation of the BEST and the development of improved test methods are the next steps in this project. The final goal is to improve accuracy so as to reduce the patient and calibration influenced variability in bone conduction hearing threshold data.

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