Knowles®

TB5

WALK INDUCED HEAD VIBRATIONS AND HEARING AID DESIGN

THUMPS: Head Vibrations While Walking

Walking is a common physical activity that is usually not considered an important environmental parameter in the design of hearing aids. In particular, head supported hearing aids are excited by a low frequency transient impulse with each step of the user. Magnetic type microphones provide natural filtering against both low frequency sound and vibration. Ceramic microphones with their capability for response to low frequency sounds also have a uniform sensitivity to vibration extending to frequencies as low as 10 Hz. This is a report on investigation of the amplitude and frequency of vibrations generated on the head and on head worn hearing aids by a walking person. It is found that these vibrations, of substantial magnitude below 100 Hz, may overload hearing aid amplifiers and cause audible thumps. They must therefore be taken into account in the overall hearing aid design.

THUMP ELIMINATION

Basically, thump elimination requires reduction of seismic vibrations of the microphone and reduction of the microphone output to levels low enough to prevent overload of hearing aid amplifier stages. We have used the term seismic vibrations to distinguish externally generated low frequency (below 100 Hz) vibrations from internal receiver generated vibrations. Internally generated vibrations are usually at a higher frequency and can be dealt with by using suitable vibration isolators. External seismic vibration in general cannot be reduced with vibration isolators. Practical isolators have resonant frequencies above 100 Hz and hence offer no attenuation at seismic frequencies.

Two practical methods of thump elimination are available. Seismic vibration is most severe in the vertical direction and least severe in the direction perpendicular to the side of the head. Mounting the microphone in the hearing aid so that the axis of greatest sensitivity is perpendicular to the side of the head will reduce vibration received by the microphone by approximately 10 dB. For all Knowles microphones this requires that the microphone diaphragm be in a plane parallel to the side of the head. Another method would be to mount the microphone with its sensitive axis perpendicular to the front of the head, but this doesn't always give the same reduction as when the sensitive axis is perpendicular to the side of the head. Proper orientation of the microphone relative to the head is the only *practical* method of reducing the vibration input.

A second method of thump elimination is the use of a high pass filter in the first stages of the hearing aid amplifier to reduce low frequency gain. This technique has several shortcomings. Space may not be available for the filter components. It will change the acoustic response below 300Hz, the amount of change depending on how much attenuation is needed and how sharp the filter is. In the case of microphones with built-in preamplifiers, such as the Knowles BL, overload could occur in the microphone and the resulting thump frequencies would be high enough to pass through the filter. As the sample calculation will show, microphone overload is improbable.

THE BODY AS A VIBRATION TRANSMITTER (HISTORY AND TEST BACKGROUND)

The literature on vibrations of the body is heavily orientated toward motions necessary to produce injury or interfere with normal body functions. This information is not particularly useful for hearing aid design since motions generated by walking do not fit into either of these categories. One study* measured the transmission of acceleration to the head of a man standing on a vibration table. Below a frequency of approximately 2 Hz the body acts as a single mass. One resonance occurs between 4 to 6 Hz and another between 10 to 14 Hz. These resonances were thought to be suggestive of a mass-spring system consisting of the entire torso on the lower spine and pelvis in the first case, and the upper torso with forward flexing movements of the upper vertebral column in the second case. A third resonance between 20 and 30 Hz due to the head mass resting on the neck stiffness would, however, probably affect a hearing aid the most. At frequencies above 100Hz, head vibrations are attenuated 40 dB or more relative to the vibration table.

* Goldman and Von Gierke, Shock & Vibration Handbook; Harris & Crede, Eds. Chap. 44, pp. 12-13, McGraw-Hill, 1961.

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WALK ACCELERATIONS

In order to verify that the vibration transmission as reported applies to a walking person and to establish the amplitudes likely to be found on the head of a walking person, miniature accelerometers were attached directly to the foot, to the top of the head, to an in-the-ear hearing aid and to a behind-the-ear hearing aid of a walking person. The study was limited in scope in that no attempt was made to select a wide variety of physiques and all data was taken on subjects walking with hard soled shoes on a concrete backed, asphalt tile floor. The subjects were male, of medium height and between the ages of 20 to 40 years.

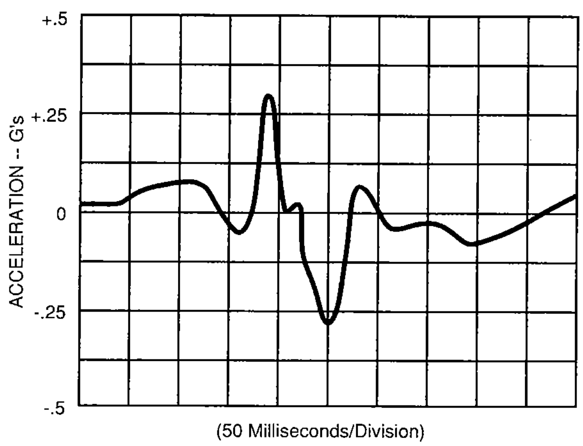

FIGURE 1. Oscillogram of a single step for vertical motion on the top of the head.

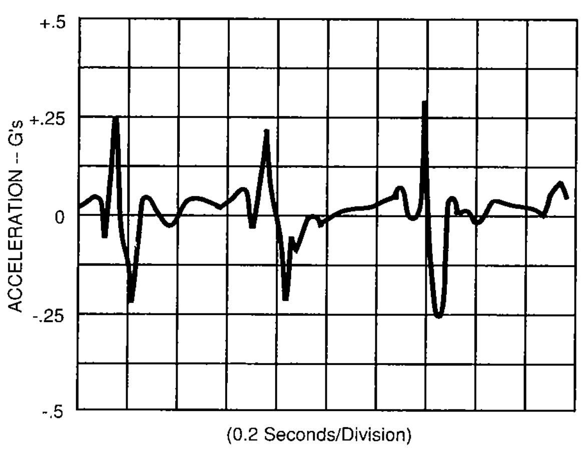

FIGURE 2. Oscillogram of a series of steps for vertical motion on the top of the head.

Measurements consisted of photographing the walk waveforms stored on an oscilloscope. The accelerometers were piezolelectric devices flat down to frequencies of 2 to 3 Hz. Several typical head waveforms are shown in Figs. 1 and 2. The data was converted to digital form and a Fourier analysis performed using a computer. The resultant shock spectrum for the oscillogram in Fig. 1 is shown in Fig. 3. (See Appendix for discussion of shock spectrum.) The walk period was normalized to 2 Hz for simplicity although actual step to step frequencies ranged from 1.6 Hz to 2 Hz. The roll-off above 50 Hz was verified by harmonic wave analysis of walk transients recorded on tape loops. The direction of measurement was vertical, the direction of maximum amplitude. The direction of least amplitude, 10 dB less than vertical, is perpendicular to the side of the head. Table I shows typical peak acceleration values for two subjects on an in-the-ear aid for up-down, side-side, and front-back motion. Table II gives vertical accelerations at four different points.

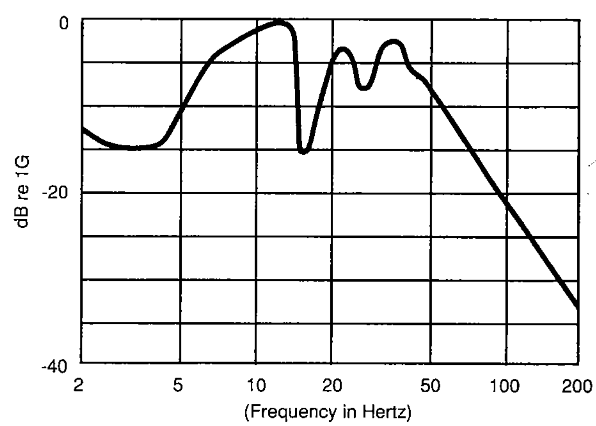

FIGURE 3. Shock spectrum for head motion waveform in Fig.1.

The peak head acceleration measured was 1 G or 980 cm/sec.^{2}. This occurred for subject I wearing a behind-the-ear aid. Accelerations in general were greater on the aids than on the top of the head. Average head accelerations were .34 G while average accelerations on hearing aids were .52 G. Frequencies of the acceleration peaks (See Fig. 3) ranged from 10 to 40 Hz.
The spectra showed 2 to 3 peaks in this frequency range.
Figs. 4 and 5 give the acceleration waveform and shock spectrum for a typical walk impulse measured on the shoe. Peak accelerations ranged from 6 to 26 G at frequencies of 200 to 400 Hz. The fundamental period was chosen to be 20 milliseconds. Average peak foot acceleration was 18 G. (See Table II.)

The walk frequency is low compared to the well damped natural resonances of the body. Therefore, the body responds to each step impulse as if it were a transient, the head motions decaying to near zero before the next step. The resultant head motion is therefore the impulse response of the head-torso-leg transmission path. The body is a low pass filter, and at frequencies above 100 Hz, walk vibrations are attenuated sufficiently that they should no longer be a noise source for a hearing aid.

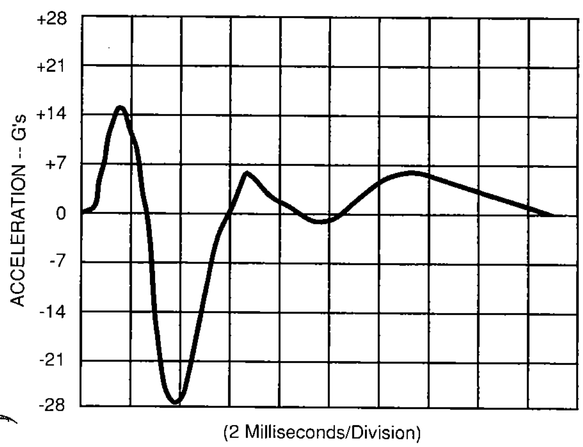

FIGURE 4. Oscillogram of vertical shoe motion for a single step.

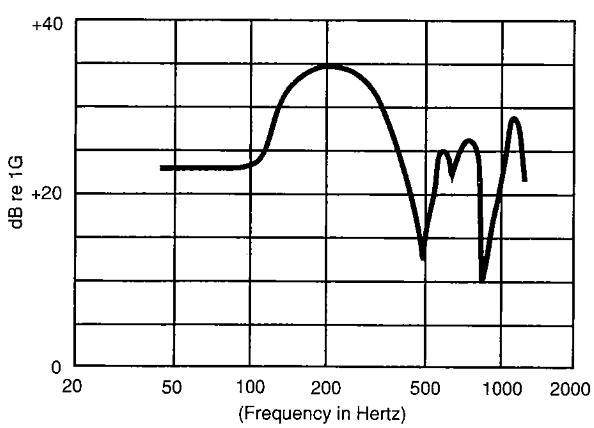

FIGURE 5. Shock spectrum for shoe motion waveform in Fig. 4.

The important result is that acceleration levels at frequencies between 10 to 40 Hz can approach values where they would interfere with hearing aid function by causing overload and generation of higher frequency harmonics in the hearing range. This potential problem can be eliminated by designing aids with sufficient dynamic range and by mounting transducers so that their vibration sensitive directions are perpendicular to the vertical direction.

Measurements on a larger, more diverse sample, may indicate trends and give a better idea of the distribution of amplitudes among individuals. These measurements are difficult to perform because individuals in our group frequently altered their walk pattern from day to day. For example, differences between right and left steps were not detectable because variations among all left or right steps were larger than variations between successive left and right steps.

SAMPLE CALCULATION OF THUMP

The following calculation is given for a BL-1680 type microphone. We first convert vibration outputs into equivalent acoustic levels in dB SPL.

BL-1680 characteristics -

(See Fig. 6 for typical response curves) Acoustic Sensitivity -57 dB re 1 volt/microbar

Vibration Sensitivity -26dB re 1 volt/G Overload Sound Pressure 105 dB SPL minimum.

Table II lists the peak accelerations measured at four different positions on four different subjects. For simplicity, we will use the highest head acceleration value measured, 1G, for subject I and a behind-the-ear hearing aid.

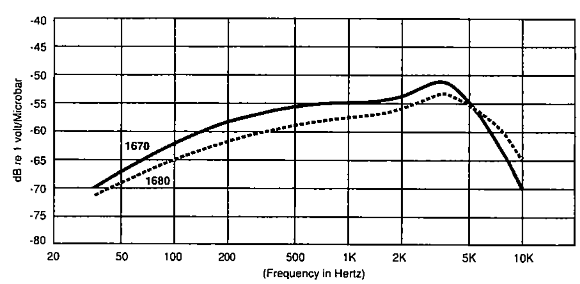

FIGURE 6A. Acoustic Response of BL-1680 and BL-1670.

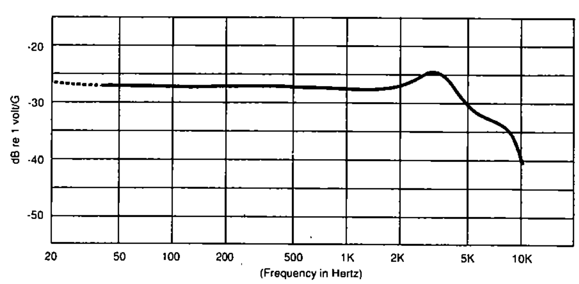

FIGURE 6B. Vibration Response of BL-1680 and BL-1670.

Using the microphone data we see that a 1G acceleration input to the microphone gives an output of -26 dB re 1 volt. This is equivalent to the output in a 104 dB SPL sound field. In this extreme case, the overload point of 105 dB SPL is approached. In general, acceleration values are lower (Table II) and the microphone would not overload and would not cause a thump. By properly positioning the microphone relative to the head, the vibration experienced by the microphone and the corresponding output can be reduced by 10 dB or more, making the probability of overload small.

Continuing with the example, we see that because the microphone does not overload, a high pass filter may be used effectively. Assuming

	Motion Direction		
	Vertical	Front to Back	Side to Side
Subject A	.82	.26	.29
Subject B	.51	.16	.11

Table 1: PEAK ACCELERATIONS, G's observed on the shell of an in-the-ear hearing aid in three directions.

Subject	Foot (Shoe)	Head	In-the-ear Aid	Behind-the-ear Aid
I	27	.51	.86	1.0
II	21	.20	.31	.43
III	19	.36	.57	.31
IV	5.3	.28	.34	.31
Average	18	.34	.52	.51

Table 2: PEAK ACCELERATION IN G's. In the vertical direction at several places on the body and on two head worn hearing aids.

the worst case, i.e., 104 dB SPL equivalent vibration output, and a hearing aid whose overload point is 90 dB SPL at its input, we calculate the parameters necessary for a simple RC filter. Referring to the shock spectrum plot, Fig. 3, we see that above 50 Hz, head vibrations are small. Thus, we must calculate the necessary filter parameter based on an effective frequency of 50 Hz. The required attenuation at 50 Hz to insure a level below 90 dB SPL is 14 dB. A simple RC filter rolls off at 6 dB/octave. To get 14 dB of attenuation, the 3 dB point must be about two octaves above 50 Hz, or 200 Hz. This would give at least 12 dB of attenuation at 50 Hz, about the amount needed. If a double RC filter were used, which rolls off at 12 dB/octave, the 3 dB down point would be about 100 Hz, and would affect the low end acoustic performance above 100 Hz much less.

In applying the above thump elimination techniques, the best results will be achieved if both methods are used together. It is interesting to note that acceleration levels on the hearing aids worn by the subjects in Table II were greater than acceleration levels on the top of the head. This could indicate a mild resonance effect between the aid mass and the ear tissue compliance.

APPENDIX

Let $f(t)$ represent the acceleration-time waveform being examined. The Fourier spectrum is then found by using the Fourier transform:

$$F(\omega)=\int_{-\infty}^{+\infty} f(t)\, e^{-j\omega t}\, dt \qquad \omega = 2\pi f$$

$F(\omega)$ was computed by expanding the waveform in a Fourier series over a period T_B.

Then the computer calculated Fourier coefficient $F(\frac{n}{T_B})$, multiplied by $\frac{T_B}{2}$, gives the value of the Fourier integral at that point. Each point of the Fourier amplitude spectrum, $F(\omega)$ was multiplied by $2\pi f$ to give the shock spectrum. The shock spectrum represents the peak response in G's that an undamped mass-spring system of resonant frequency f would experience when excited by the waveform from which the shock spectra was calculated. Further information on this method can be found in the following references:

1. Charles Morrow, *Shock and Vibration Engineering;* Vol.1, Chaps. 5 and 9, Wiley & Sons, 1963.

2. Rubin, *Shock and Vibration Handbook;* Harris and Crede, Eds; Chap. 23, McGraw-Hill, 1961.

3. Hans Olesen, "Frequency Analysis of Single Pulses", *Bruel and Kjaer Technical Review,* No. 3, 1969, pp. 3-16.