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Tuning and sensitivity of the human vestibular system to low-frequency vibration

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ABSTRACT

Mechanoreceptive hair-cells of the vertebrate inner ear have a remarkable sensitivity to displacement, whether excited by sound, whole-body acceleration or substrate-borne vibration. In response to seismic or substrate-borne vibration, thresholds for vestibular afferent fibre activation have been reported in anamniotes (fish and frogs) in the range −120 to −90 dB re 1 *g*. In this article, we demonstrate for the first time that the human vestibular system is also extremely sensitive to low-frequency and infrasound vibrations by making use of a new technique for measuring vestibular activation, via the vestibulo-ocular reflex (VOR). We found a highly tuned response to whole-head vibration in the transmastoid plane with a best frequency of about 100 Hz. At the best frequency we obtained VOR responses at intensities of less than −70 dB re 1 *g*, which was 15 dB lower than the threshold of hearing for bone-conducted sound in humans at this frequency. Given the likely synaptic attenuation of the VOR pathway, human receptor sensitivity is probably an order of magnitude lower, thus approaching the seismic sensitivity of the frog ear. These results extend our knowledge of vibration-sensitivity of vestibular afferents but also are remarkable as they indicate that the seismic sensitivity of the human vestibular system exceeds that of the cochlea for low-frequencies.

The otolith organs, the sacculus, utriculus and lagena, primarily respond to whole-body acceleration or tilt in gravity [9]. In fish these also are important auditory structures for acoustic near-field (particle motion) sensing [13,17]. Several studies have determined behavioral particle motion audiograms for non-specialist species of fish, e.g. the cod, plaice and dab [4]. These have indicated that the region of best sensitivity lies between 40 and 120 Hz, with threshold acceleration values of about −120 dB re 1 *g* at 80 Hz. During the course of evolution the amniote ear developed new structures for far-field (sound pressure) hearing in air, including the basilar papilla and the mammalian cochlea [6].

It has been established, however, that the otolith organs in terrestrial vertebrates have conserved a particular sensitivity to substrate- or bone-conducted sound [2,15,16,22] consistent with their function as near-field sound sensors in fish [4]. In some species of frog the saccule shows a fish-like band-pass response to acceleration with best frequencies between 20 and 160 Hz and thresholds between −90 and −120 dB re 1 *g*, while others show a low-pass response with best frequencies at 10–20 Hz [14]. Sensitivity to audio-frequency vibration has also been demonstrated in mammalian vestibular organs. In the monkey [27] best frequencies were between 125 and 177 Hz, with phase-locking threshold as low as −80 dB re 1 *g*, and in the guinea-pig at 500 Hz thresholds were 10 dB above the ABR threshold [3]. At present, however, no such threshold measurements have been obtained for the human vestibular system and this was the aim of our study.

Non-invasive assessment of human vestibular sensitivity can be accomplished by measurement of the powerful vestibulo-ocular reflexes (VOR) to head acceleration. The VOR normally serves to maintain eye gaze with head tilt or rotation and its main effects are mediated by a simple three-neuron arc connecting the vestibular portion of the 8th nerve to the motor neurones of the extraocular muscles [1]. In response to stimuli such as head movements, reflex activity occurs in the extraocular muscles, producing a compensatory eye movement. By placing surface electrodes around the eyes, synchronous muscle activity can be recorded in the form of ocular vestibular evoked myogenic potentials (OVEMPs) [23,25]. These responses are vestibular, rather than cochlear, in origin as they are present in deaf patients but are absent in patients with loss of vestibular function [18,24]. We aimed to measure the tuning and sensitivity of OVEMPs to whole head vibration in the transmastoid plane.

Four volunteers (2 females and 2 males between 31 and 52 years of age) with no auditory or vestibular deficits were stimulated using sinusoidal accelerations between 12.5 and 800 Hz (12.5, 25, 50, 100, 200, 400 and 800 Hz). The subjects were seated

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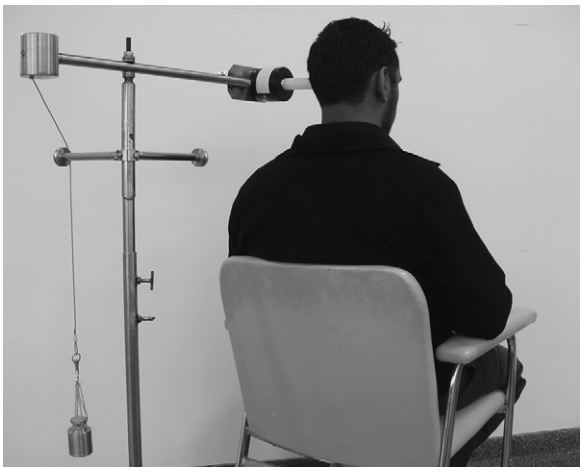

Fig. 1. A photograph of the experimental apparatus used to apply a perspex rod (attached to a B&K 4810 mini-shaker) with a constant tonic force to the mastoid of the subject. The stimulator is mounted on a 2:1 lever which can rotate freely in the horizontal plane and is pivoted on a vertical adjustable stand. A torque is applied at the opposite end of the lever by means of a mass M suspended by wire over a pulley below the fulcrum which can be adjusted so that the tension in the wire acts at an angle θ to the plane of the lever. The tonic force at the mastoid then is given by $F = 1/2\,Mg\cos\theta$, which for a mass of 1 kg and applied angle of 45° is 3.5 N.

upright and directed their gaze upwards (for frequency tuning) or straight ahead (for threshold determination), as this provided the best signal to noise ratio. A 70-year-old patient with autoimmune vestibular disease was tested as a control. He had no VEMP response measured from the neck muscles when stimulated with standard stimuli (air- and bone-conducted 500 Hz, 2 ms tone bursts at 142 dB peak SPL and 136 dB peak FL, respectively). There was no response to conventional caloric stimulation and minimal response to ice water stimulation. He was stimulated at 100 Hz only and OVEMPs were recorded during up-gaze.

Stimuli were delivered by a cylindrical perspex rod (diameter 2.5 cm, length 9.2 cm) attached to a vibrator ("Minishaker", model 4810, Bruel & Kjaer P/L, Denmark). The rod was placed normal to the skull in the horizontal plane just above the mastoid in order to produce primarily translational head acceleration (Fig. 1). A constant force was maintained by a pulley system with a weight of 1–2 kg. Stimuli were generated by means of customized software, using a laboratory interface (1401plus, Cambridge Electronic Design, Cambridge, UK).

Head acceleration along the y-axis was measured using two accelerometers (model 751-100, Endevco, California, USA) placed normally to the skull on the temporal bone directly superior to the ears and held in place by tight elastic bandages. For all stimulus frequencies, skull acceleration was kept constant at −20 dB re 1 g (0.1 g) by adjusting the driving voltage to the stimulator as required. This excluded effects of skull resonance. OVEMPs were measured using pairs of Ag/AgCl electrodes in bipolar montages. The first electrodes were placed on the orbital margin inferior to the eye and the second 2–3 cm below on the cheek, with an earth electrode over the sternum. Stimuli were presented for 100–1000 repetitions at a fixed rate of approximately 3 Hz. EMG was amplified and bandpass filtered (5–1000 Hz), then sampled at 10 kHz from 10 ms before to 290 ms following stimulus onset and averaged. Blink artefacts were automatically rejected. Amplification and analog filtering were performed by a bank of D150 amplifiers (Digitimer Ltd., Welwyn Garden City, UK) and their outputs sampled by means of a second CED 1401plus using SIGNAL software.

For frequency tuning the results for the four normal volunteers were averaged and a spectral analysis carried out. This was done digitally by a bank of filters or resonators (damped sinusoids) with impulse response $y(t) = \sqrt{2\alpha}\exp(-\alpha t)\exp(i\omega t)$ spaced logarithmically on the frequency-axis (24 per octave) and with a constant-Q tuning such that $\alpha = \omega/2Q$ where the sharpness of tuning was defined by $Q = 32$. The output was obtained by taking the root peak power over time for each frequency channel scaled by $\omega/2$ to normalise for the different duration of the stimuli and to make the peak power in time- and frequency-domain equivalent. The frequency response was fitted using a second-order velocity resonance curve such that the OVEMP magnitude as a function of frequency $V(\omega)$ was approximated by

$$V(\omega) = G\left[\frac{\omega^2}{(\omega_0-\omega)^2+\gamma^2\omega^2}\right]^{1/2} \quad (1)$$

where G is the gain, $\omega_0 = 2\pi f_0$ is the resonance frequency and $\gamma = \omega_0/Q$ is the damping factor.

OVEMP and auditory perceptual thresholds were measured in the four normal volunteers using the best frequency (100 Hz). Responses close to threshold were considered present if they had at least two clear peaks at the correct latency. Threshold measurements were obtained from the average of the OVEMP responses across the subjects in order to increase the signal-to-noise ratio. Amplitudes were measured using the clearest peak. Near threshold, the presence of a response was confirmed by a cross-correlation between the signal and response. The OVEMP magnitude as a function of acceleration $V(a)$ was plotted against the stimulus level using a log-log scale, assuming a power law of the form

$$V(a) = ka^{\beta} \quad (2)$$

where k is a scaling constant and β is the power law parameter. When transformed logarithmically the power law becomes linearized, so that β may be obtained from the slope of the linear regression.

The mean auditory threshold x_T and dispersion σ were obtained by fitting a logistic function:

$$p(x) = \left\{1+\exp\left[-\frac{(x-x_T)}{\sigma}\right]\right\}^{-1} \quad (3)$$

where $x = 20\log(a)$ and $p(x)$ is the probability of hearing the stimulus. This may be linearized by letting $y = \ln(p/(1-p))$ so that the logistic function becomes $y = x/\sigma - x_T/\sigma$. The parameters x_T and σ may then be obtained from the slope m and intercept c of the linear regression such that $\sigma = 1/m$ and $x_T = -c \times \sigma$.

When the head was vibrated at different frequencies it produced OVEMPs which differed in amplitude and morphology (Fig. 2). The best frequency in all four volunteer subjects was 100 Hz, which evoked responses which were clearly out of phase in the two eyes. The 100 Hz tuning can be seen in both the morphology, which displays ringing (Fig. 2), and when the response is analysed in the frequency domain (Fig. 3). In the patient with vestibular hypofunction only very small responses were seen, despite adequate head acceleration (Fig. 4).

The response obtained in normal subjects was indicative of a system resonance and indeed the data was fitted quite well using the velocity resonance equation with $f_0 = 80$ Hz and quality factor $Q = 2$ (Fig. 5a, Eq. (1)). At or close to the best frequency it was possible to measure responses to very small head accelerations which required averaging even to read the peak acceleration. When stimulus intensity at 100 Hz was systematically reduced, responses were recorded in the averaged response (across the four subjects) to accelerations as low as 70.2 dB below 1 g (Fig. 5b). When transformed logarithmically the OVEMP amplitude growth curves were well-fitted with a linear regression with slopes (β) ranging from 0.6 to 0.8 (Fig. 5b, Eq. (2)).

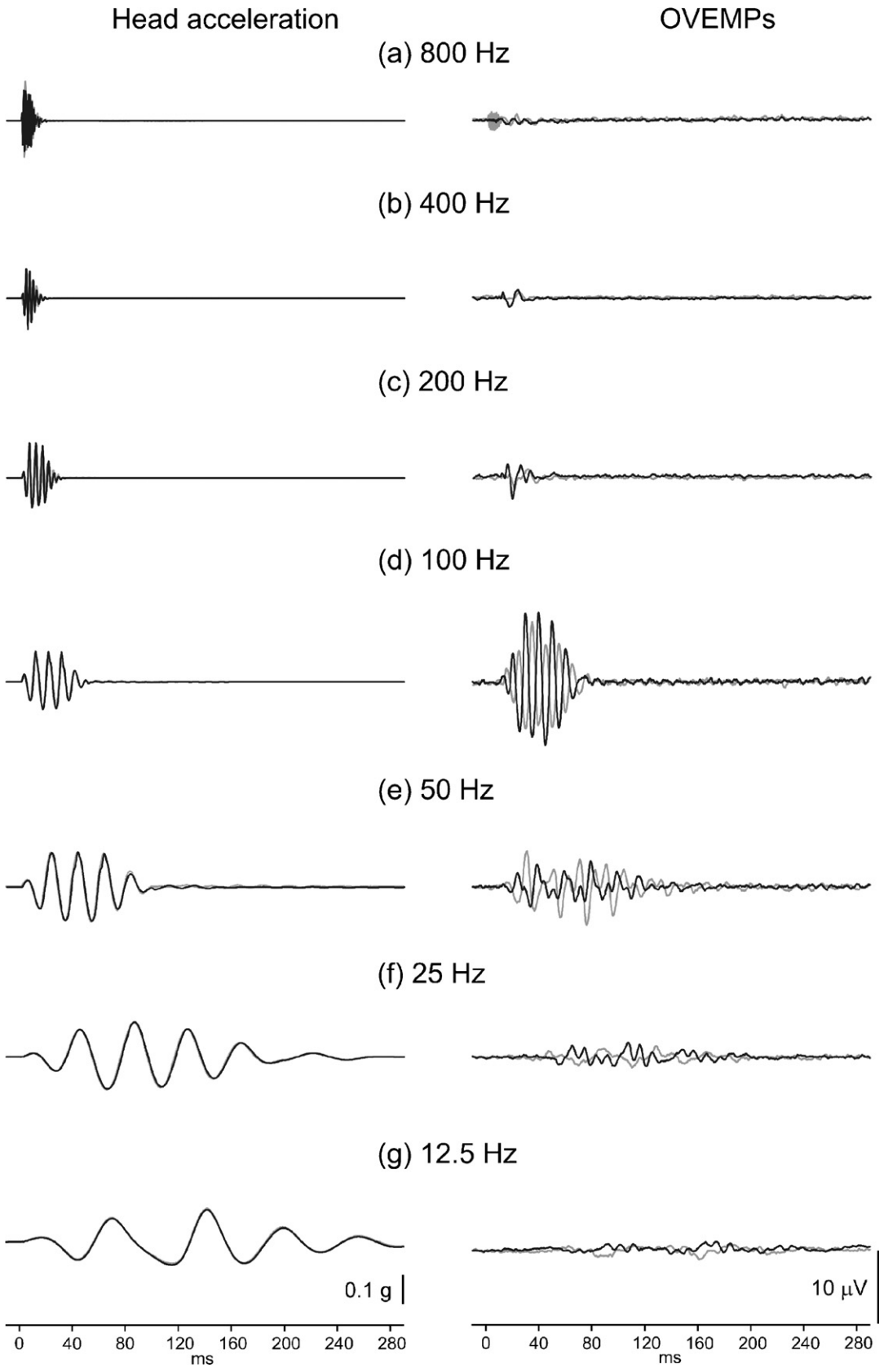

Fig. 2. Head acceleration measured from the left (black) and right (grey) mastoids (left column) and OVEMP responses from the left (black) and right (grey) eyes (right column) to sinusoidal vibrations at 800–12.5 Hz (parts a–g). The head moved as a whole in response to the applied forces and positive acceleration was always towards the left. Stimuli had lengths of 5 cycles (1 cycle rise, 2 hold and 2 fall), except at 800 Hz (duration 12.5 ms: 2.5 ms rise, 5 hold, 5 fall) and 12.5 Hz (duration 2 cycles: 1/2 rise, 1 hold and 1/2 fall).

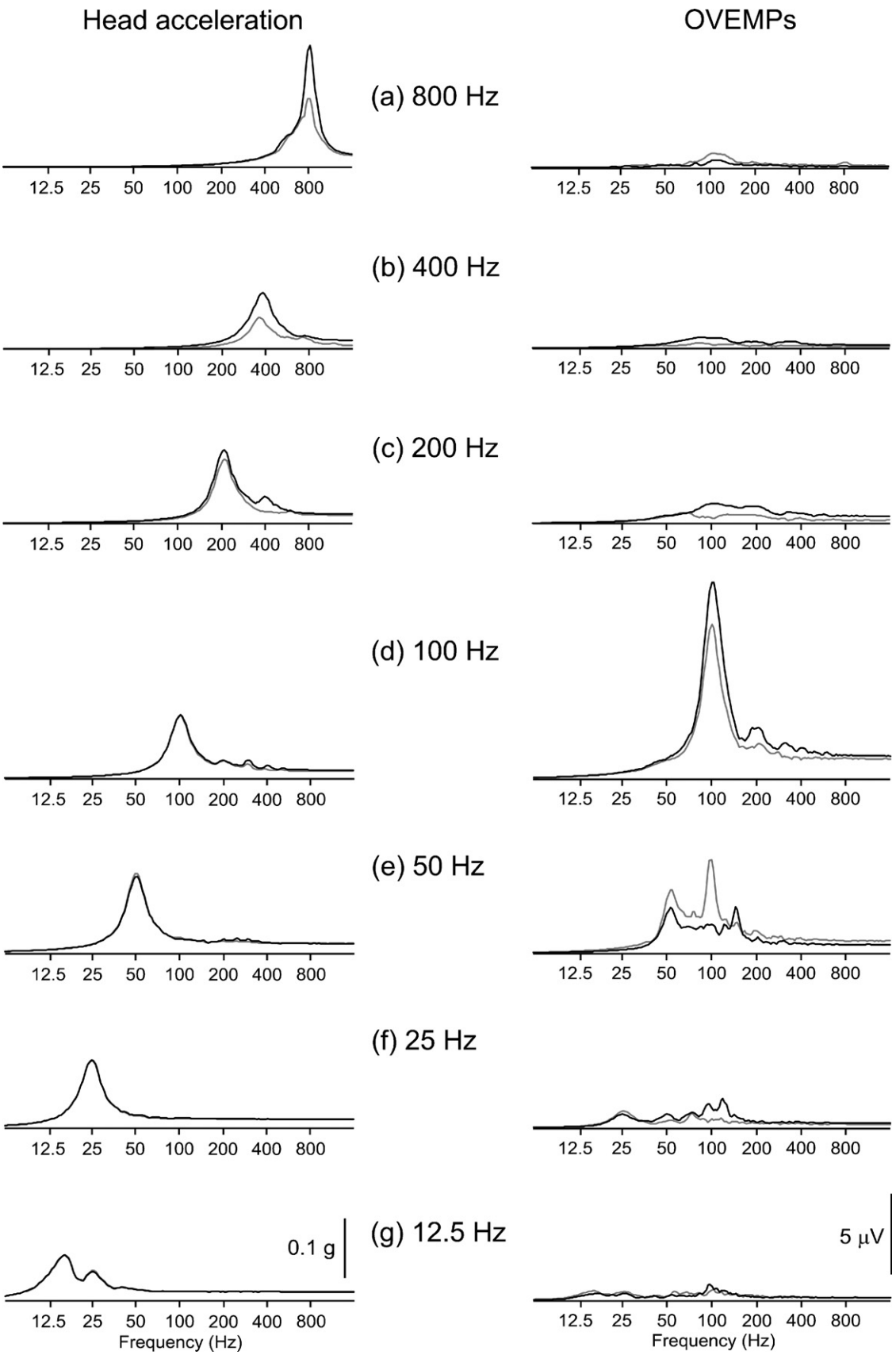

Fig. 3. The root power spectrum of the head accelerations (left column) and OVEMP responses (right column) to sinusoidal vibrations at 800–12.5 Hz (parts a–g) (black = left and grey = right).

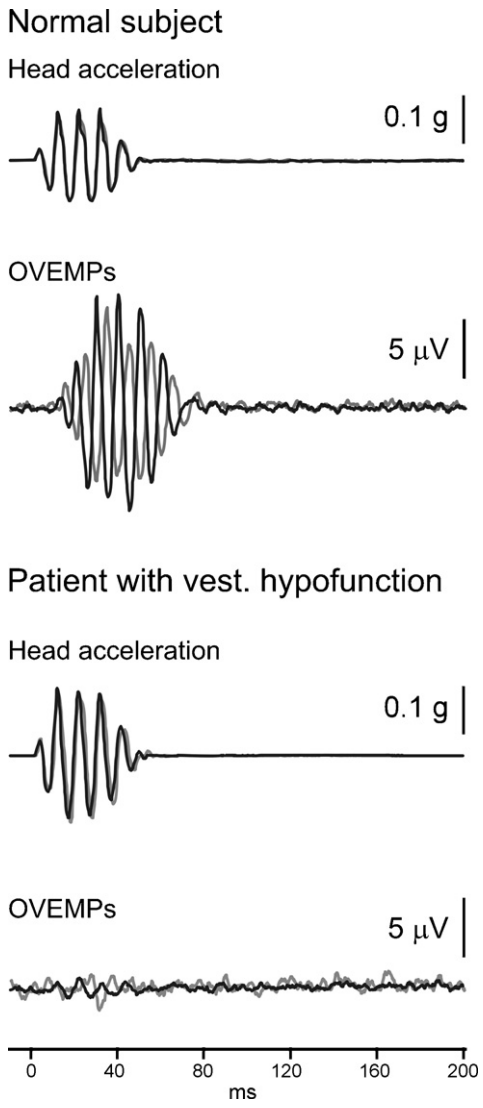

Fig. 4. Head acceleration and OVEMPs recorded from a normal subject (upper traces) and a patient with bilateral vestibular hypofunction (lower traces), who were best matched for age. Head acceleration, measured from the left (black) and right (grey) sides of the head, was slightly greater in the patient. Despite this, the OVEMPs, measured from the left (black) and right (grey) eyes, were very small in the patient compared to the control subject.

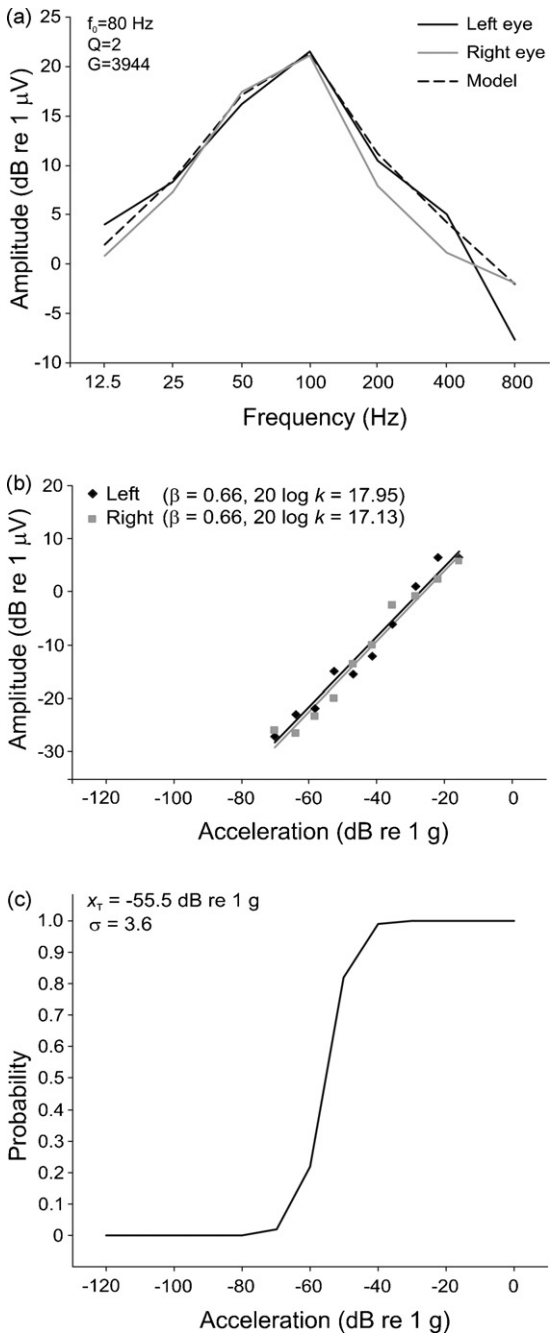

Fig. 5. (a) OVEMP amplitude in the average of four subjects as a function of stimulus frequency (Eq. (1))(in all sections left eye = black, right eye = grey), (b) growth in mean OVEMP amplitude as a function of mean head acceleration (Eq. (2))and (c) probability of hearing the stimulus as a function of head acceleration (Eq. (3)).

At the lowest intensity the subjects could not hear or feel the stimulus on the mastoid. The estimated auditory threshold at 100 Hz (Fig. 5c, Eq. (3)) was approximately 55 dB below 1 *g*, which compares well with previous estimates [26]. The mean perceptual threshold was some 15 dB higher than the lowest intensity at which we could measure an OVEMP in the averaged response.

Our results show very clearly that OVEMPs are highly tuned with a best frequency of about 100 Hz and with a band-pass characteristic between about 25 and 200 Hz, thus showing a strong similarity with data from afferent fibres of the anamniote otoliths. The lowest intensity at which we were able to record a reliable OVEMP in the averaged response was −70.2 dB re 1 *g*, which is comparable with Graybiel and Patterson's [7] estimate of 3.44×10^{-4} *g* (−69.2 dB re 1 *g*) for the threshold for stimulation of the human otolith organ using acceleration. This value is 20–50 dB above estimated thresholds for the 8th nerve afferent data in anamniotes. However, as in the case of estimating auditory thresholds from the auditory evoked potentials (AEP), which can overestimate thresholds by as much as 20 dB or more [11], especially for low-frequency sound, it is almost certainly the case that our OVEMP thresholds are 20 dB or more above the primary afferent threshold. There will be some signal loss during synaptic transmission in the VOR pathway and also the signal-to-noise characteristics will be reduced from surface EMG recordings, with many other muscles contributing noise. Allowing for this additional factor, our estimated threshold lies between 0 and 30 dB of the anamniote values. We may postulate therefore that human vestibular afferents in response to audio-frequency vibration do indeed fall within the sensitivity range of fibres innervating the frog saccule. With regards the growth in response, the power law slopes we obtained, 0.6–0.8, were in the range found

both physiologically and psychophysically for the auditory system [20].

The very low thresholds we found are remarkable as they suggest that humans possess a frog- or fish-like sensory mechanism which appears to exceed the cochlea for detection of substrate-borne low-frequency vibration and which until now has not been properly recognised. Previous reports have shown that vibration around 100 Hz is an effective means of activating the vestibular system in humans but signs of such activation are usually not evident in normal subjects, only becoming apparent in the presence of a unilateral lesion [8,10,12]. By measuring OVEMP responses to low-frequency vibration, not only is vestibular activation evident in normal volunteers, but very low stimulus intensities can be shown to be effective.

These observations raise some fundamental questions regarding the mechanisms that may contribute to the tuning and sensitivity properties. Hair-cells are known to exhibit electrical resonance in the low-frequency range due to the interaction of transduction and basolateral currents [9,19]. The otolith organs are known to have a mechanical tuning due to their elastic and inertial properties, the band-width of their mechanical response extending to 500 Hz [5]. It is possible also that the neural properties of the VOR circuit contribute to the tuning. In addition, a fundamental question is also raised as to the possible behavioral consequences, if any, such a mechanism may have. Several, not necessarily mutually exclusive, hypotheses have been proposed. It has been suggested that vestibular acoustic sensitivity could contribute to the quality of acoustic perception and acoustic affect and may account for the compulsion to exposure to loud low-frequency sound [22]. As early as the 1930s Tait [21] suggested that the otoliths in humans could contribute to the perception of one's own voice. Given the threshold values we have obtained, there can be no doubt that the voice will itself activate the vestibular system. Our findings should therefore stimulate further investigations into the questions raised by the remarkable human vestibular sensitivity to low-frequency seismic energy.

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