Influence of Surgical Plugging on Horizontal Semicircular Canal Mechanics and Afferent Response Dynamics

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Rabbitt, R. D., R. Boyle, and S. M. Highstein. Influence of surgical plugging on horizontal semicircular canal mechanics and afferent response dynamics. J. Neurophysiol. 82: 1033–1053, 1999. Mechanical occlusion of one or more of the semicircular canals is a surgical procedure performed clinically to treat certain vestibular disorders and used experimentally to assess individual contributions of separate canals and/or otoliths to vestibular neural pathways. The present experiments were designed to determine if semicircular canal afferent nerve modulation to angular head acceleration is blocked by occlusion of the endolymphatic duct, and if not, what mechanism(s) might account for a persistent afferent response. The perilymphatic space was opened to gain acute access to the horizontal canal (HC) in the oyster toadfish, Opsanus tau. Firing rate responses of HC afferents to sinusoidal whole-body rotation were recorded in the unoccluded control condition, during the process of duct occlusion, and in the plugged condition. The results show that complete occlusion of the duct did not block horizontal canal sensitivity; individual afferents often exhibited a robust firing rate modulation in response to whole-body rotation in the plugged condition. At high stimulus frequencies (about >8 Hz) the average sensitivity (afferent gain; spikes/s per °/s of head velocity) in the plugged condition was nearly equal to that observed for unoccluded controls in the same animals. At low stimulus frequencies (about <0.1 Hz), the average sensitivity in the plugged condition was attenuated by more than two orders of magnitude relative to unoccluded controls. The peak afferent firing rate for sinusoidal stimuli was phase advanced ~90° in plugged canals relative to their control counterparts for stimulus frequencies ~0.1–2 Hz. Data indicate that afferents normally sensitive to angular velocity in the control condition became sensitive to angular acceleration in the plugged condition, whereas afferents sensitive to angular acceleration in the control condition became sensitive to the derivative of acceleration or angular jerk in the plugged condition. At higher frequencies (>8 Hz), the phase of afferents in the plugged condition became nearly equal, on average, to that observed in controls. A three-dimensional biomechanical model of the HC was developed to interpret the residual response in the plugged condition. Labyrinthine fluids were modeled as incompressible and Newtonian; the membranous duct, osseous canal and temporal bone were modeled as visco-elastic materials. The predicted attenuation and phase shift in cupular responses were in close agreement with the observed changes in afferent response dynamics after canal plugging. The model attributes the response of plugged canals to labyrinthine fluid pressure gradients that lead to membranous duct deformation, a spatial redistribution of labyrinthine fluids and cupular displacement. Validity of the model was established through its ability to predict: the relationship between plugged canal responses and unoccluded controls (present study), the relationship between afferent responses recorded during mechanotonic; the membranous duct, osseous canal and temporal bone were modeled as incompressible and Newtonian; the membranous duct, osseous canal and temporal bone were modeled as visco-elastic materials. The predicted attenuation and phase shift in cupular responses were in close agreement with the observed changes in afferent response dynamics after canal plugging. The model attributes the response of plugged canals to labyrinthine fluid pressure gradients that lead to membranous duct deformation, a spatial redistribution of labyrinthine fluids and cupular displacement. Validity of the model was established through its ability to predict: the relationship between plugged canal responses and unoccluded controls (present study), the relationship between afferent responses recorded during mechanical indentation of the membranous duct and physiological head rotation, the magnitude and phase of endolymphatic pressure generated during HC duct indentation, and previous model results for cupular gain and phase in the rigid-duct case. The same model was adjusted to conform to the morphology of the squirrel monkey and of the human to investigate the possible influence of canal plugging in primates. Membranous duct stiffness and perilymphatic cavity stiffness were identified as the most salient model parameters. Simulations indicate that canal plugging may be the most effective in relatively small species having small labyrinths, stiff round windows, and stiff bony perilymphatic enclosures.

INTRODUCTION

The role of the vestibular end organs in providing inputs to motion-control neural systems such as the vestibulococular reflex (VOR) first was demonstrated by Mach, Crum-Brown, and Ewald in the late 1800s (Camis 1930; Crum-Brown, 1894; Ewald, 1892). Ewald was among the first to use surgical modifications of the labyrinth to manipulate semicircular canal afferent inputs to the brain stem (Camis 1930; Ewald, 1892). He pioneered the approach termed “canal plugging,” where the slender duct of a semicircular canal is occluded surgically in an attempt to block sensitivity to angular head accelerations. Since the reintroduction of canal plugging by Money and Scott (1962), the approach has been employed for the treatment of certain vestibular disorders (Minor et al. 1998; Parnes 1992; Parnes and McClure 1990), to improve the exposure of the internal auditory canal fundus (Antonelli et al. 1995), and to study the influence of individual canals as well as otolithic inputs to vestibular reflex systems (e.g., Aw et al. 1996; Baker et al. 1982; Böhm et al. 1985; Correia and Money 1970; Minor and Goldberg 1990; Money 1967; Paige 1983, 1985; Parnes and McClure 1990; Schor and Miller 1982; Watt 1976). Results of these investigations are consistent with the hypothesis that the procedure selectively blocks the sensory capability of the occluded semicircular canal, at least under the conditions tested. It is unclear to what extent efficacy of the procedure might extend to other experimental conditions and/or motion stimuli.

Recent evidence indicates that the angular VOR recovers a high-frequency component even after the semicircular canals have been plugged completely (Angelaki and Hess 1996; Angelaki et al. 1996; Broussard and Bhatia 1996; Cohen et al. 1996; Davis et al. 1997; Lasker et al. 1997; Yakushin et al. 1997). If plugged canals indeed are rendered nonfunctional by the surgical manipulation, then one might conclude that the...
recovered responses reflect a form of neural adaptation that may use inputs from the otolith organs to sense angular movements. Another possibility is that canal plugging may not completely inactivate semicircular canal afferent responses and that the recovered VOR may draw, at least in part, from residual inputs originating from the plugged canals. Present results support the second hypothesis. Data recorded from single afferent nerves in the toadfish, *Opsanus tau*, show that the semicircular canals continue to respond, with modified dynamics, to angular accelerations after complete surgical occlusion of the endolymphatic duct. Pressure-induced deformation of the membranous labyrinth appears to be the key mechanism underlying the residual response. This is supported by the fact that a biomechanical model, accounting for membranous duct elasticity and labyrinthine fluid flows, quantitatively accounts for the observed changes in semicircular canal afferent responses in the toadfish after duct occlusion. When the same model was applied to the morphology of the human and of the squirrel monkey, it predicted a residual plugged-canal response in these species as well. Results may have important implications regarding the contributions of individual end organs to vestibular mediated neural systems in plugged-canal preparations.

**Experimental methods**

Preparation of toadfish and neural recording methods follow that previously described by Boyle and Highstein (1990) and Rabbitt et al. (1995). Briefly, fish of either sex and weighing ~500 g were provided by the Marine Biological Laboratory, Marine Resources facility (Woods Hole, MA). Anesthesia was induced by immersion in MS222 (25 mg/l sea water, 3-aminobenzoic acid ethyl ester, Sigma). Fish were immobilized partially by an intramuscular injection of pancuronium bromide (0.05 mg/kg); pancuronium bromide does not block opercular motion and allows for natural respiration by “gilling.” Each fish was placed in a plastic tank filled with fresh sea water covering all but the dorsal surface of the animal. The eyes and remainder of the body were kept covered with moist tissues.

The experimental set-up is illustrated in Fig. 1. A small craniotomy was made lateral to the dorsal course of the anterior canal, allowing direct access to the horizontal canal (HC) nerve, anterior canal (AC) ampulla, common crus (CC), and the utricle. The posterior canal (PC) and caudal section of the HC were not exposed. The craniotomy was elongated to expose the horizontal semicircular canal for a distance of ~0.8–1.3 cm posterior to the ampulla measured along the curved centerline of the horizontal canal. A bolus of fluorocarbon (FC75, 3M Corp.) was injected into the cranial opening to partially fill the dorsal region of the perilymphatic vestibule. The fluorocarbon improved optical access to the labyrinth and prevented evaporation of perilymph and dehydration of tissue. Fluorocarbon is not soluble in water or perilymph such that the bolus remained isolated to a constrained region of the surgical opening. A layer of normal perilymph remained on the surface of the labyrinth and a pool of perilymph continued to bathe the HC nerve. Glass microelectrodes filled with 2 M NaCl or LiCl were used for extracellular or intraxonal afferent recordings from the right horizontal canal nerve ~1 mm from the HC ampulla. Conventional amplification and spike discrimination were employed.

The HC of each fish was occluded mechanically by compressing the membranous endolymphatic duct firmly against the cartilaginous bone using a ~1.2-mm-diam glass rod tipped with a malleable substance (Takiwax; Cenco) that conforms to any surface irregularities. The location of the plug was measured in each fish using a pointer secured to a three-axis micromanipulator. Plugs were located 1 ± 0.2 cm from the crista as measured along the curved centerline of the canal. To ensure that the plugging procedure did not damage the transduction apparatus, afferent responses to sinusoidal stimulation (angular rotation and/or mechanical canal indentation) were monitored continuously while compressing the canal. After each experiment, a saturated solution of alcine green in artificial endolymph was injected into the HC ampulla to allow visual confirmation of the canal blockage. In several fish, the dye moved through the position of the blockage, indicating that only a partial plug was achieved. Data from partially plugged canals were excluded from the study.

![FIG. 1. Experimental set up. Schematic shows the location of the glass rod used to plug the membranous duct, the location of the horizontal canal (HC) mechanical stimulator and the location of the utricular stimulator (gray ellipse). Apparatus was secured to a rate table with the plane horizontal canal oriented perpendicular to the axis of rotation for controlled delivery of sinusoidal angular motion.](http://jn.physiology.org/content/112/4/1034.f1)
In five experiments, the fish tank was secured to a velocity servo-controlled rate table (10 ft-lb DC servo motor, Contraves) and oriented to provide maximal angular stimulation to the horizontal canal (Boyle and Highstein 1990). Micromanipulators (Fig. 1) were secured rigidly to the rate table and rotated with the animal. Control afferent responses to sinusoidal rotation were collected before canal plugging in each fish. The firing rate response of individual afferents to sinusoidal rotation of the animal (whole body = head velocity) were collected at angular velocities of ~0.1–20°/s over the frequency range ~0.1–20 Hz in the unoccluded and occluded conditions.

In five separate animals, horizontal canal afferent responses to micromechanical indentation of the slender duct and utricle were recorded in the unoccluded control condition and in the occluded condition. In these experiments, the fish tank was placed on a vibration-isolation table. The micromechanical stimuli used microrodents fashioned from flat-end, ~0.12-cm diam glass rods, with one positioned perpendicular to the long-and-slim portion of the HC and a second glass rod over the utricle; the HC and utricular stimulators were lowered to static preload indentations of ~12.5 and ~25 μm, respectively (Boyle and Highstein 1991; Dickman and Correia 1989; Rabbitt et al. 1995). The HC indenter was located ~0.3 cm from the cupula, as measured along the curved centerline of the canal, between the canal block and the ampulla. The utricular indenter was located ~0.5 cm from the HC cupula as measured along the curved centerline coordinate. Each rod was secured to a piezoelectric microactuator (Burleigh PZL 060-11) mounted in-line with a linear variable differential transducer (LVDT; Schaevitz DEC-050) to obtain an analog measurement of instantaneous position of the glass rod stimulator.

Digital data acquisition (Cambridge Electronic Design 1401Plus, Apple Macintosh interface) was used to record externally discriminated spike times, angular velocity of the rate table, and the displacement of the indenters (LVDTs). The tachometer and LVDT signals were amplified externally to span the 12-bit range of the A/D. This places the A/D digital round off two orders of magnitude below the noise thus providing resolution of ~0.25°/s table velocity and <0.5 μm indentation displacement. These analog signals were filtered at 250 Hz and digitized at 500 samples per second. Stimulus trigger signals and externally discriminated spikes were recorded to a resolution of 0.08 ms.

Analysis was done off-line using a custom interactive analysis procedure (IgorPro, Wave Metrics). The constrained first-harmonic gain and phase of afferent modulation were determined using the data analysis described by Rabbitt et al. (1995, 1996). Briefly, afferent gain and phase were computed for five or more consecutive stimulus cycles by manually selecting portions of the record and averaging the results by cycle. Afferent spike times were averaged relative to the stimulus trigger using 100 bins per cycle phase histograms. Results were subsequently inverted to yield the firing rate (spikes/second) as a function of phase in the cycle. A discrete Fourier analysis was applied to the stimulus waveform and to the phase histogram neural response to compute the first-harmonic amplitude and phase of the stimulus and the afferent response. Empty bins were ignored in the afferent fitting procedure and the average rate was constrained to be ≥0 (Rabbitt et al. 1996).

Modeling methods

To understand how canal plugging might alter cupula deflections, we derived a three-dimensional mathematical model based on the morphology of the HC, membranous duct deformability, and labyrinthine fluid mechanics. The model includes cupular visco-elasticity, fluid viscosity, membranous duct visco-elasticity, osseous canal/temporal bone visco-elasticity, and, in primates, the connection to the middle ear via the cochlea. A detailed derivation of the model is provided in the APPENDIX. Plugging enters into the model by reducing the cross-sectional areas of the membranous duct and/or osseous canal to zero along a short segment of the canal (modeled as 0.13 cm in length). This occludes the flow of labyrinthine fluids at the location of the plug. The model was applied to the HC morphology of the fish, human, and squirrel monkey. The squirrel monkey model was based on morphological data from Igarashi and colleagues (Igarashi 1966; Igarashi et al. 1981), the infant human model was based on Curthoys and colleagues (Curthoys and Oman 1987; Curthoys et al. 1977), and the fish model was based on work by Ghanem et al. (1998). Measures of squirrel monkey membranous duct thickness were provided by C. Fernández (unpublished data). The curved centerlines of the primate HC models were assumed to fall within a single plane. The local cross-sectional area was simplified to a circular endolymphatic duct enveloped by an annular perilymphatic canal. For the fish and human models, the plug was located along the slender portion of the HC at a distance of 1 cm from the crista as measured along the curved centerline of the duct; for the squirrel monkey, this distance was reduced to 0.5 cm to account for the smaller size.

Short cylindrical fluid elements (100) were used to model the endolymph and 100 short annular fluid elements to model the perilymph. The elements were of equal length and oriented concentricity along the curved centerline of the HC. The geometry enters into the model by specification of a unique spatial position and cross-sectional area for each of the cylindrical and annular fluid elements. Continuity of fluid pressure and mass flow rate were enforced at the boundaries between adjacent fluid elements. Endolymph flow within the each cylindrical element was modeled as unsteady axisymmetric Stoke’s flow. Perilymphatic flow also was modeled as unsteady axisymmetric Stoke’s flow but was confined to an annular space rather than a cylindrical space.

Angular acceleration of the head induces inertial forces in both the perilymph and the endolymph; this lead to fluid flows and associated pressure gradients. The resulting transmembrane pressure gradients (local endolymph minus perilymph
pressures) are counteracted in the model by the distentional stiffness of the membranous duct. The elastic modulus and thickness of the membrane was assumed to be homogeneous in the present simulations. The volume compliance of each cylindrical segment of the membranous duct was therefore dependent only on position and local duct size. Larger cross-sections are more compliant due to Laplace’s law and Hooke’s law (Eq. A14). Proportional damping was used to model viscous properties of the membrane. The bone enclosing the perilymphatic space also was modeled as a visco-elastic using the same equations as for the membranous duct but adjusting the thickness and stiffness to that of the bone.

In the present experiments, a small volume of perilymph within the surgical opening was replaced with fluorocarbon. To account for this, the density of the fluid for the annular elements located in this region was set to that of fluorocarbon—normal perilymph density was used elsewhere. The same portion of the perilymphatic cavity was opened surgically to air. To account for this in the model simulations, the stiffness of the bony enclosure was set to zero in this region. For the squirrel monkey and the human, a short region of the perilymphatic vestibule was modeled as connected to the cochlea rather than lined by bone. A single lumped compliance was used to model the net volumetric stiffness of the cochlea and the middle ear.

The local streamwise endolymph volume displacement, streamwise perilymph volume displacement, endolymph pressure, perilymph pressure, membranous duct deformation, and bony enclosure deformation were allowed to vary as a function of position and were used in computations as the dependent variables. The model equations were cast in finite-difference form and solved using LU decomposition for sinusoidal forcing (Press et al. 1986) using custom software programmed in Igor Pro (Wave Metrics, Lake Oswego, OR).

Simulations were carried out independently for both the control and the plugged conditions. The response attenuation predicted to be caused by plugging was determined by dividing the complex-valued cupular volume displacement in the plugged condition by the cupular volume displacement in the control condition. Attenuation results are reported in Bode form (magnitude: cupular displacement in the plugged condition divided by the displacement in the control condition; and phase: cupular phase in the plugged condition minus the phase in the control condition). The model is linear, so the predicted attenuation did not change with stimulus amplitude. The global cupular attenuation and phase shift was predicted to extend to individual spatial locations and hair cells almost uniformly across the sensory epithelium (see Discussion). An attenuation of 0.1, for example, would therefore imply that the mechanical stimuli exciting each hair cell would be 10 times smaller in the occluded condition than in the control condition. For stimuli restricted to the range where hair cells and afferents respond linearly (Boyle and Highstein 1990; Highstein et al. 1996; Rabbitt et al. 1996), a 10-timess reduction in the mechanical activation of hair cells would further predict a 10-timess reduction in afferent gain.

**Experimental Results**

**Afferent response to rotation after canal occlusion**

Afferent responses to sinusoidal rotation (0.1–20 Hz) were collected before canal plugging in each fish to ensure that the organ was responding normally in the control condition. The HC duct then was plugged slowly, and a second set of afferent responses was recorded in the occluded condition. Control afferent nerves were typed into three groups, “low gain, velocity sensitive” (LG), “high gain, velocity sensitive” (HG), and “acceleration sensitive” (A) based on responses to sinusoidal whole-body rotation using the scheme developed by Boyle and Highstein (1990). The control population consisted of 261 records from 18 afferents (5 LG, n = 47; 4 HG, n = 46; 9 A, n = 168). An average of 15 (min 5, max 41) records were obtained for each control afferent at single frequencies over the range from 0.1–20 Hz. After occlusion, 440 records were obtained from 25 afferents in the same five fish. An average of 18 (min 5, max 50) records were obtained at discrete frequencies for each afferent in the plugged condition. Because the plugging procedure alters afferent response dynamics, it was not possible to use previously established methods to directly determine LG, HG, and A types in the plugged condition. It was possible, however, to estimate afferent types on the basis of data collected in the plugged condition using the model. This estimate is provided in the following text along with model results.

In each experiment, the firing rate of an individual afferent was monitored while slowly compressing the membranous duct against the cartilaginous substrate. Occlusion by compression of the duct is in itself a type of mechanical stimulus that induced large magnitude increases in afferent firing rate (Rabbitt et al. 1995). Figure 2A shows the response of an afferent recorded during compression of the duct; the afferent was LG type (0.28 spikes/s per °/s head velocity with a 2° phase lag before plugging at 2 Hz). The sharp increases in firing rate (indicated in A by arrows) correspond to the periods of duct compression made by manually lowering a glass rod using a fine micromanipulator drive. The afferent was allowed to recover to a firing rate near its resting value before continuing compression. Brief disturbances in the recovery, most evident from 700 to 930 s in this record, are fish movement artifacts that were unavoidable in some specimens. After the period of compression, recovery to the background firing rate was slowest for low-gain afferents, presumably due to the near absence of adaptation and rate sensitivity (Boyle and Highstein 1990; Fernández and Goldberg 1971; Goldberg and Fernández 1971a,b). Afferents always maintained a firing rate during plugging. However, the modulation of the firing rate to rotation was maintained only when the compression of the canal proceeded at an average rate of ≤4–5 μm/s. Rapid compression of the membranous duct always damaged the end organ and eliminated any detectable afferent modulation to rotational stimuli, followed up to ~6 h after compression. Injection of alcaine dye into the labyrinth provided evidence of partial or complete detachment of the cupula at its apex after rapid compression of the canal; data from these damaged canals were excluded from this report.

After duct occlusion, individual afferents routinely were observed to exhibit robust firing rate modulations in response to angular head accelerations. On the whole, however, the population averages showed consistent differences in the plugged and control conditions. The firing rate behavior of one example afferent to sinusoidal rotation recorded after complete plugging is shown in Fig. 2 C and D, at four stimulus frequencies as indicated (0.5, 1, 2, and 5 Hz). The sinusoidal
angular velocity of the rate table is given in B, the corresponding afferent firing rate in C, and phase histograms averaged over multiple cycles in D. To avoid firing rate nonlinearities (see Boyle and Highstein 1990) and improve recording stability, the peak angular velocity of the stimulus was reduced (in this case from 8.3 to 1°/s) as the frequency of rotation was raised (in this case from 0.5 to 5 Hz). Examine the middle two panels in B–D. At the same stimulus amplitude of 4.5°/s, a dramatic increase in the magnitude of firing rate modulation was observed when the stimulus frequency was increased from 1 to 2 Hz. To quantify the modulation, the response gain was defined as the constrained first harmonic of the afferent phase histogram (output; D, solid line) divided by the first harmonic of the stimulus (input; D, dashed line). For this particular afferent, the gain increased by two orders of magnitude as the frequency was increased from 0.5 to 5 Hz. This large increase in gain over this frequency range was not observed in control canals. Although the frequency response of this afferent is quite distinct from individual controls, it is notable that the response magnitude at any given frequency falls well within the physiological range observed for present control population and previously studied afferents in this species (Boyle and Highstein 1990; Rabbitt et al. 1995).

Bode plots were constructed for the response of each afferent using neural impulses collected during sinusoidal angular rotation at discrete frequencies (0.1–20 Hz). Figure 3 shows responses, plotted in the form of gain (spikes/s per °/s; A) and phase with respect to head velocity (°) of 2 representative afferents from one fish tested separately, one in the unoccluded control condition (+) and the other in the occluded condition (o). Because of the length of time required for initial afferent characterization and the irreversible nature of the plugging procedure, we were unsuccessful in obtaining a direct comparison of the responses of a single afferent in the plugged and unoccluded conditions.
The afferents presented in Fig. 3 were selected for illustration because their responses fell near the population averages in the control and occluded conditions. Simple polynomial curve fits (— and - - -) are provided to illustrate the trends. Note the large phase advance in B of the afferent in the occluded condition. The peak of the response modulation led peak ipsilaterally directed head velocity by $\sim 140^\circ$ at 1 Hz, for example. This large phase advance was not observed in the present uncoccluded controls or in previous studies (Boyle and Highstein 1990; Rabbitt et al. 1995). Also note the pronounced increase in the slope of the gain (A) that occurred in the plugged condition; again, a response property not observed in controls.

The trends in afferent gain and phase following canal occlusion are most clearly illustrated by the population results. Figure 4 provides the average gain (B) and phase (C) of responses obtained in the control condition (thin solid lines) and in the plugged condition (thick solid lines). The number of records at each frequency is provided in A. Boyle and Highstein (1990) demonstrated previously that afferents in the toadfish define a continuous distribution with large interafferent variability in gain and phase. This interafferent variability is responsible for the spread in the current population averages, indicated by the vertical bars at each frequency (1 SD).

On average, acute canal plugging caused $\sim 100$-fold attenuation in gain at 0.1 Hz, 4-fold attenuation at 1 Hz, but only a factor of $\sim 2$ at 10 Hz (Fig. 4B). The average slope of the plugged-canal gain was considerably steeper than the control population. In controls, afferent responses fell between angular velocity sensitive and angular acceleration sensitive (Fig. 4C) (also see Boyle and Highstein 1990). After canal occlusion, afferent responses were advanced an additional $\sim 90^\circ$ for stimulus frequencies less than $\sim 2$–5 Hz and fell between angular acceleration sensitive and angular jerk sensitive. Greater than 2–5 Hz the phase of plugged-canal population dropped and became nearly equal to the average of control afferents greater than $\sim 8$ Hz. These data show that canal plugging under the current experimental conditions does not eliminate afferent responses but rather changes their frequency-dependent response dynamics (gain and phase).

Afferent response to mechanical indentation after canal occlusion

The persistence of afferent modulation after HC duct occlusion indicated that cupular deflections also persisted after plugging. This led to the hypothesis that acceleration of the head caused transmembrane pressure gradients (local endolymph pressure minus perilymph pressure) sufficiently large to induce local distentions and contractions of the membranous duct, ampulla, and utricle accompanied by concomitant redistributions of endolymph and perilymph. Movement of endolymph in the ampulla during such a redistribution would lead to afferent modulations. To investigate feasibility of this idea, transmembrane pressure gradients were generated by mechanical indentation of the membranous duct in the absence of head acceleration. Mechanical stimuli were applied to the HC limb and to the utricle as described by Rabbitt et al. (1994, 1995). In these previous studies as well as in present control afferents ($n = 18$), indentation of the HC limb elicited an excitatory response, whereas indentation of the utricle elicited an inhibitory response of HC afferents. The excitatory response is associated with movement of endolymph out of the slender duct toward the utricle; the inhibitory response is associated with movement of endolymph out of the utricle toward the slender duct. To determine if this behavior persists after canal plugging, responses of individual afferents ($n = 12$) were recorded during HC and utricular indentation after complete occlusion of the duct. Mechanical stimulator locations were as indicated in Fig. 1. All afferents continued to be excited by HC indentation and continued to be inhibited by utricular indentation after plugging. Responses of a representative afferent after complete HC duct occlusion are shown in Fig. 5 for both HC (A) and utricular (B) indentation. Notice that the phase of the afferent response is reversed for utricular indentation relative to HC indentation indicating oppositely directed cupular deflection—the same type of relationship observed in uncoccluded controls. The afferent shown was typed as LG and responded nearly in phase with positive HC indentation; other afferents, particularly those having characteristics identified as HG or A type had high gains to indentational stimuli exceeding 10–20 times that of the LG afferent shown.

Excitatory responses of plugged canals to indentation of the
distinct experimental conditions were simulated: a surgically opened perilymphatic cavity corresponding to the present acute recording conditions in the toadfish (acute condition) and a sealed perilymphatic cavity with a completely ossified HC plug modeling a hypothetical chronic plugged-canal condition (chronic condition). The two conditions are discussed separately in the following text.

**Acute condition**

For stimuli between $-0.1$ and 2 Hz, data indicate that afferents sensitive to angular velocity in the control condition became sensitive to angular acceleration in the acute plugged condition, whereas afferents sensitive to angular acceleration in the control condition became sensitive to angular jerk. As an example, consider the gains of two afferents shown in Fig. 2A. The gain of the patent canal afferent (+) had a slope of $m \approx 1/2$ [$m = \Delta \log(\text{spikes}°/s)/\Delta \log(\text{Hz})$]. Because this slope is on a log-log scale, it provides the exponent of a power law. The afferent (o) shown after plugging had a gain exponent of $m \approx 2$. The exponent determines to what extent the afferent is sensitive to displacement ($m = -1$), velocity ($m = 0$), acceleration ($m = 1$), or jerk ($m = 2$). In previous studies (Boyle and Highstein 1990) and in the present population, afferents in the control condition had gain exponents primarily in the range $0 < m < 1$, indicating sensitivity falling between angular velocity and acceleration. Afferents from plugged canals had gain exponents primarily in the range $1 < m < 2$, indicating sensitivity falling between angular acceleration and jerk. The increased phases observed in the plugged condition are consistent with this change in exponent. The same trends are exhibited clearly in the population average shown in Fig. 3.

Afferent responses to angular movements were present immediately after the plugging procedure. Time was not sufficient for adaptation or remodeling of the end organ, so the increased gain exponent and advanced phase in plugged-canal responses relative to controls must reflect changes in canal mechanics. The model derived in the **Appendix** was applied to investigate if canal macromechanics could account for the observed changes in afferent response dynamics. Care was taken to reproduce the same conditions present during the experiments. To model the surgically opened condition, the pressure on the surface of the perilymph was set equal to atmospheric in the region of the surgical opening (0 gauge pressure). The plug was modeled by reducing the cross-sectional area of the endolymphatic duct to zero along a short segment of the canal. The perilymphatic duct was not occluded in the experiments nor was it occluded in the simulations for the acute condition.

Dashed curves in Fig. 4 show how the present control population (thin solid curves) would be predicted by the model to respond had it been possible to record the very same afferents in the plugged condition. This theoretical result was computed by dividing the model prediction for the cupular volume displacement in the plugged condition by the model prediction for the control condition to obtain a transfer function describing mechanical attenuation. The attenuation then was multiplied by the average first-harmonic afferent data collected in the control condition to project the data to the plugged condition (dashed curves, Fig. 4). There is reasonably good agreement between the gain and phase predicted by the model (dashed curves) and the average data collected in the plugged condition.

**Model results and discussion**

The mathematical model derived in the **Appendix** was used to interpret the present data and to address to what extent the results might extrapolate to other species and conditions. Two distinct experimental conditions were simulated: a surgically opened perilymphatic cavity corresponding to the present acute recording conditions in the toadfish (acute condition) and a sealed perilymphatic cavity with a completely ossified HC plug modeling a hypothetical chronic plugged-canal condition (chronic condition). The two conditions are discussed separately in the following text.
ATTENUATION IN THE ACUTE CONDITION. Figure 6 shows the plugged canal data and the model projection. The apparent difference between the control and occluded populations may indicate a sampling bias toward HG- and A-type afferents in the plugged condition—a bias that could explain the relatively small difference between the control and plugged conditions. As noted in the preceding text, the control population consisted of 47, 46, and 168 records for A-, B-, and C-type afferents, respectively. The phase and gain reduction in the plugged condition (thick solid curves) are slightly under predicted by this model projection. Part of the difference may be due to the fact that data in the control and plugged conditions were recorded from two distinct afferent populations. By using the model to project plugged data to the control condition, we estimate that the plugged population consisted of 13, 138, and 289 records obtained from 2 LG-, 9 HG-, and 14 A-type afferents, respectively. As noted in the preceding text, the control population consisted of 47, 46, and 168 records obtained from 5 LG-, 4 HG-, and 9 A-type afferents, respectively. The apparent difference between the control and occluded populations may indicate a sampling bias toward HG- and A-type afferents in the plugged condition—a bias that could explain the relatively small difference between the plugged canal data and the model projection.

ATTENUATION IN THE ACUTE CONDITION. Figure 6 shows the frequency dependent attenuation (A, nondimensional) and phase shift (B, ° re: control condition) predicted to be caused by canal plugging in the surgically opened acute condition for the toadfish (solid curves), squirrel monkey (thick short dashed curves) and the human (thick long dashed curves). Simulations for the chronic condition with a sealed perilymphatic cavity are discussed in the following text. Attenuation predictions for primates in the acute recording condition are remarkably similar to data for the fish even though there are relatively large interspecies differences in labyrinthine morphology and absolute cupular volume displacements. This similarity occurs because the differences in cupular volume displacement between the species are present in both the control and plugged conditions and hence cancel out to a large extent in computing the attenuation. Attenuation curves in the surgically opened acute condition depend primarily on the morphology and mechanical properties of the membranous duct; in the chronic condition discussed in the following text, the stiffness of the perilymphatic cavity, osseous canal, and connection to the middle ear are also important model parameters.

PRESSURE DISTRIBUTIONS. The model predicts that endolymph movement and cupular displacements arise in plugged canals owing to distention of some regions of the endolymphatic duct accompanied by concomitant contractions of other regions. Understanding the response of plugged canals therefore requires an understanding of transmembrane pressure gradients underlying the membranous duct deformation. Predicted transmembrane pressure gradients for the fish are shown in Fig. 7 for canal-centered rotation at five instants in time spanning one-half of the 5.7-Hz stimulus cycle (color scale gradients within each canal, see Fig. 8A for quantitative results). Time increases from top to bottom. Plugged-canal simulations are shown on the right and control simulations on the left. The top panels show the instant in time corresponding to peak clockwise (CW, ipsilateral) acceleration (A), the middle panels peak clockwise velocity (C), and the bottom panels peak counterclockwise (CCW) acceleration (or peak CW displacement, E).

The pressure in the control condition is predicted to be dominated by adverse gradients arising from local changes in the cross-sectional areas of the endolymphatic and perilymphatic ducts as well as by any linear acceleration component (the linear component is ~0 for the canal-centered case shown in Fig. 7). The peak magnitude transmembrane pressure gradient occurs in the long and slender duct during peak CW acceleration and in the utricle during peak CCW acceleration. This transmembrane pressure gradient is responsible for local membranous duct distention and contraction. Endolymph is predicted to flow from regions of contraction into regions of distention while preserving total endolymph volume. The distention-related flow is superimposed on a distention-independent flow that would be present even in the rigid-duct case. In the control condition (Fig. 7, left), endolymph flow is predicted to be dominated by the distention-independent component, but in the plugged condition (right), endolymph flow is predicted to be entirely due to distention and contraction of the membranous duct.

At any given point in the canal the pressure modulates sinusoidally with a fixed amplitude and phase. Consider, for example, a point just caudal to the occlusion (Fig. 7, right). At a time corresponding to peak acceleration (A, top), the pressure is positive, but 180° later in the cycle (E, bottom), the pressure at the same point in the canal is negative. The peak pressure at this location is in phase with peak acceleration of the head, or 90° advanced from peak angular velocity of the head. Specific results vary with position. Predicted pressure distributions for the toadfish are shown quantitatively in Fig. 8 at 5.7 Hz in the form of amplitude (dyne/cm²; A) and phase (° peak positive
pressure re: peak angular velocity; B) as functions of distance from the crista measured along the curved centerline coordinate (also see Fig. 13). The location of the canal plug is indicated by the vertical arrow. Results for the fish were computed for a surgically opened perilymphatic space, so the perilymphatic pressure is atmospheric (0 gauge) in the open region and relatively small in the closed osseous canal. This causes the transmembrane pressure (thick curves; A and B) to be dominated by the endolymph (thin solid curves) in both the control (thick dashed curves) and plugged (thick solid curves) conditions.
conditions. As discussed in the following text, this is not predicted for the case of a completely sealed chronic condition due the contribution of relatively large perilymphatic pressure modulations.

ENDOLYMPH DISPLACEMENT. For a perfectly rigid membranous duct, the position dependent magnitude of the endolymph displacement would be inversely proportional to the local duct cross-sectional area owing to conservation of mass. This was approximately true for simulations in the control condition where membranous duct distention was relatively small but was not the case for plugged canals where the endolymph flow was entirely dependent on duct distention. In control canal simulations, endolymph displacement was large in the slender duct relative to the ampulla. The opposite was predicted in plugged canals. The phase of peak endolymph displacement also was predicted to change between the two conditions. At 0.1 Hz, for example, the endolymph displacement in the control condition was predicted to be maximum at the time corresponding to peak head velocity, but in the plugged canal, the displacement was predicted to be maximum at the time corresponding to peak head acceleration. Hence consistent with the data, the integrating effect of fluid viscosity was lost in plugged canals.

Greater than ~2 Hz, endolymph flow was predicted to become unsteady resulting in more complex fluid displacement profiles. A balance between unsteady inertia and viscous shear stress was responsible for the complex flow patterns predicted in the large cross-sectional regions of the canal. Profiles in the slender region of the membranous duct were predicted to remain nearly Poiseuille at 5.7 Hz; but this did not extend greater than ~15 Hz, where the unsteady effect was predicted to extend to the slender duct as well. The unsteady effect depends on the local cross-sectional area and is identified most easily by the “M”-shape endolymph displacement patterns in Fig. 7, B–D. Similar unsteady profiles have been shown previously in simple one-dimensional flows by Stokes in the late 1800s (see Fung, 1990) and in arterial flow by Womersley (1955, 1958). As discussed by Damiano and Rabbitt (1996), the unsteady effect may influence afferent responses as a result of two factors. The first is that the viscous drag for unsteady flow would not be in phase with the bulk fluid flow rate, a factor that may alter the frequency-dependent phase and gain of cupular volume displacements. The second is that the gain and phase of unsteady cupular displacement would be spatially inhomogeneous. Spatial diversity may account for some of the differences in the mechanical activation of hair cells reported by Highstein et al. (1996). The same type of unsteady effects predicted in the control condition also were predicted in plugged canals. In addition to attenuation in the gain and phase of cupular volume displacements, the model predicted a small change in the displaced configuration of the cupula in control re: plugged canals due to the unsteady effects (see Fig. 7). As a result, canal plugging may influence the response dynamics of various classes of afferent nerves in slightly different ways depending on the morphological structure of dendritic projections in the crista. The change in the displaced shape of the cupula after canal plugging, however, was predicted to be small (~5%) relative to the large attenuation in macromechanical cupular volume displacement and shift in phase. Hence the model predicted that the attenuation caused by canal plugging would act almost equally on the activation of all hair cells such that all afferent types, if recorded within their linear response range, would be attenuated almost equally and phase shifted after canal plugging. Because it was not possible to record from the same identified afferents in both the control and plugged conditions, we unable to directly test this model prediction.

DEGREE OF BLOCKAGE. One of the most important model parameters influencing the effectiveness of canal plugging was the extent to which the lumen of the endolymphatic duct was blocked. Quantitative predictions are provided in Fig. 9 for the toadfish in the acute condition at seven levels of blockage ranging from 100% (thick solid curve) to 50% (thin solid curve). Complete endolymphatic plugs were predicted to cause large attenuations at low frequencies, whereas leaky plugs were predicted to be much less effective. For example, reducing the membranous duct area by 50% over a length of 1.3 mm was predicted to have little effect on the mechanical response (attenuation in Fig. 9 is ~1 and the phase shift is ~0). At high frequencies (more than ~8 Hz), canal plugging in the acute recording condition was predicted to be ineffective in attenuating the cupular gain regardless of the extent of endolymphatic duct occlusion.

PLUG LOCATION. Plugs positioned close to the ampulla were predicted by the model to be more effective than plugs located further from the ampulla. This was due to the increase in total volumetric compliance of the section of the duct located between the ampulla and the plug that accompanies the increased...
length. At 1 Hz in the fish, a plug located 1.2 cm from the crista, measured along the curved center line of the duct, was predicted to attenuate the cupular displacement by a factor of 0.27, whereas a plug located 0.4 cm from the crista was predicted to attenuate the cupular displacement by a factor of 0.17. The shift in phase was insensitive to the position of the plug and was predicted to vary by only 1° when the plug was located at 1.2 versus 0.4 cm from the crista. In the experiments, the plug was located 1 ± 0.2 cm from the crista. According to the model, the ±0.2 cm variation in the position of the plug may have introduced ~15% interanimal variation in the attenuation data and ~0.5% variation in the phase.

MEMBRANOUS DUCT STIFFNESS. The model also predicted that increasing the membrane stiffness would make canal plugging more effective in attenuating cupular responses, whereas reducing the stiffness would cause the opposite. Quantitative predictions for the fish are provided in Fig. 10 for the acute recording conditions. The phase predicted at high frequencies was particularly sensitive to membranous duct stiffness. Stiff labyrinths were predicted to show a phase advance of 90° at 10 Hz after plugging, whereas compliant labyrinths were predicted to respond with a phase similar to controls. It is not uncommon to have interanimal variations in soft tissue stiffness spanning more than an order of magnitude (Fung 1981, 1990). Soft-tissue stiffness also is known to vary systematically with age and sex. Hence some of the variability in high-frequency results between individual animals, between

### Table 1. Physical parameters used in model simulations

<table>
<thead>
<tr>
<th>Description</th>
<th>Symbol</th>
<th>Value</th>
<th>Units</th>
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</thead>
<tbody>
<tr>
<td>Endolymph density</td>
<td>( \rho^e )</td>
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</tr>
<tr>
<td>Endolymph viscosity</td>
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<td>dyne-s/cm²</td>
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<td>Endolymph mod. rigidity</td>
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<td>dyne/cm²</td>
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</tr>
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<td>cm</td>
</tr>
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<tr>
<td>Membranous duct density</td>
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<td>g/cm³</td>
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<tr>
<td>Membranous duct thickness</td>
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<td>cm</td>
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<tr>
<td>Membranous duct stiffness</td>
<td>( E_m )</td>
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<td>dyne/cm²</td>
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<tr>
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<td>s</td>
</tr>
<tr>
<td>Mass proportional damping</td>
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<td>1/s</td>
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<tr>
<td>Squirrel monkey middle-ear stiffness</td>
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<td>( K_{c} )</td>
<td>5 \times 10⁷</td>
<td>dyne/cm²</td>
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</table>

species, and between different laboratories may simply reflect natural variations in labyrinthine membrane stiffness. In the present simulations, we optimized the stiffness to fit the average data reported in Fig. 4 (see Table 1). The stiffness ascertainment by this method fell within the range reported by Yamashita et al. (1998) for the toadfish labyrinth based on pressure-volume data. It should be noted, however, that these pressure data span more than an order of magnitude, indicating large interanimal variability in membrane stiffness. This is consistent with the interanimal variability seen in the phase of plugged canal afferent responses in the present population at high frequencies of rotation. Results for the human and squirrel monkey discussed in the following text were computed using the same stiffness as for the fish. Predictions for these species therefore may change as experimental data become available.

MECHANICAL INDENTATION. As described earlier, Fig. 5 shows the response of a single LG afferent to mechanical indentation of the HC duct (A) and the utricle (B) in the plugged condition. It is important to note that positive utricular indentation continues to cause inhibitory responses even after complete occlusion of the HC, thus indicating endolymph movement through the ampulla and a concomitant distention of the membranous duct. The ratio of the afferent gain during HC indentation to the response of a single LG afferent fibers during utricular indentation was 6.7 ± 2.6 (range 3.2–13.4; 0.5–5 Hz, \( n = 13 \)) in the control unoccluded condition and increased to 20.6 ± 9.2 (range 8.2–34.6; 0.5–5 Hz; \( n = 12 \)) in the occluded condition. The present model predicted a ratio of 9.4 for the control condition and 15.4 for the plugged condition; values well within the observed range. These model predictions are sensitive to the morphology of the duct, position of the plug, positions of the stimulators, and preloading of the stimulators, factors that varied between individual animals used in the experiments. Hence the relatively small differences between the model predictions and the average data are not surprising. What is most important to note is that a single model including membranous duct distensibility is sufficient to account for the observed changes in afferent responses after semicircular canal

![Image](http://jn.physiology.org/)
plugging for both physiological head rotation and for mechanical indentation stimuli.

MODEL VALIDATION IN THE ACUTE CONDITION. Both the attenuation and phase shift predicted for the toadfish in the acute, surgically opened, condition are in quantitative agreement with the present afferent data (see Figs. 4 and 6). It is important to note that the very same model (and same parameter set) predicts the endolymphatic pressure modulations recorded in the ampulla during mechanical indentation (Yamauchi et al. 1998). The present model also predicts the experimentally established relationship between afferent responses during mechanical duct indentation of the HC and during rotation stimuli in O. tau (Rabbitt et al. 1995). The same model also reproduces previous theoretical results when the stiffness is increased to simulate the rigid-duct case (Damiano and Rabbitt 1996; Oman et al. 1997). These validations suggest that the model may be sufficiently general to address questions outside the range of currently available experimental data.

Chronic condition

PREDICTED ATTENUATION IN CHRONIC CONDITIONS. To address how the present results might extrapolate to chronic canal plugs in primates, model parameters were adjusted to account for a completely sealed perilymphatic space and a stiff ossification completely precluding flow of endolymph and perilymph at the location of the plug. The connection to the middle ear, absent in the fish, was included as a lumped parameter volumetric stiffness located at the surface of the perilymphatic vestibule. Quantitative predictions for the chronic condition (II) are shown in Fig. 6 for the squirrel monkey (thin dotted lines) and the human (thin dashed lines). Canal plugging was predicted, on average (0.01–20 Hz), to generate ~10 times more cupular attenuation in the chronic condition (thin dotted and dashed curves) beyond that present in the surgically opened acute condition (thick dotted and dashed curves). In all simulations chronic canal plugging was predicted to attenuate cupular volume displacements by >100 × less than ~1 Hz and hence is expected to be highly effective at low frequencies. As the stimulus frequency was increased sufficiently high, canal plugging was predicted to become ineffective in attenuating cupular displacements even in the chronic, sealed, condition. The specific frequency at which plugging was predicted to become ineffective depended on the specific morphological structure and physical parameters. Parameters for the toadfish are relatively well known (Rabbitt et al. 1995; Yamauchi et al. 1998), but parameters for the human and the squirrel monkey are not as well established. Obtaining afferent recordings without compromising the perilymphatic space also has proven problematic, such that direct experimental testing of model predictions for the chronic condition, has not yet been possible. VOR data for the squirrel monkey appears to be consistent with the present model (Davis et al. 1997; Lasker et al. 1997), but such comparisons are indirect. Predictions for the chronic condition therefore should be viewed in light of sensitivities to model parameters discussed in the following text.

PRESSURE DISTRIBUTIONS. In contrast to the surgically opened acute condition, the endolymphatic and perilymphatic pressures in the sealed chronic condition were predicted to become nearly identical to each other (Fig. 8, C and D; thin dashed and solid curves). This pressure balance leads to relatively small transmembrane pressure gradients in both the patent (thick dashed curves) and plugged (thick solid curves) canals. This difference in the transmembrane pressure distributions between the acute condition (fish; Fig. 8, A and B) and the chronic condition (human; Fig. 8, C and D) underlies the reduced effectiveness of canal plugging predicted for surgically opened preparations (also see Fig. 6).

CONNECTION TO THE MIDDLE EAR. Computations for the human and the squirrel monkey used a stiff osseous canal that effectively eliminated any deformation of the bony perilymphatic space. Compliance of the vestibule therefore was dominated by the connection to the cochlea and the middle ear. This compliance has not been measured in primates. Lynch et al. (1987) report a round-window volumetric compliance in cat on the order of 10⁻⁸ cm³/dyne. It is not appropriate, however, to directly use this value in the present model due to the difference in size of the ears and the magnitudes of pressures involved. Treating the round window as an elastic plate would predict the compliance to be inversely proportional to the fourth power of the radius. On the basis of this, the compliance of the human round window may be nearly two orders of magnitude less than that of the cat. It is also important to note that the cat round window compliance was measured using sinusoidal stimuli generating pressures on the order of 100 dyn/cm²—a pressure two orders of magnitude higher than predicted during volitional angular head rotations. It is well known that, in the absence of pretension, soft tissues become more compliant for small strains (Fung 1981). On the basis of these considerations, volumetric stiffnesses of 5 × 10⁵ and 7.5 × 10⁶ dyn/cm³ were selected as baseline values for the human and squirrel monkey, respectively (see Table 1). The influence of changing the middle ear stiffness on the attenuation predicted for the human is shown in Fig. 11. Results for the baseline stiffness of 5 × 10⁵ dyn/cm³ are shown as thick solid curves. When the stiffness was reduced by a factor of 100 (dotted curves), plugging was predicted to be ineffective in reducing cupular deflections at frequencies greater than ~1 Hz. In contrast when the stiffness was increased by a factor of 10 (dashed curves), plugging was predicted to be highly effective reducing cupular deflections even at frequencies ~10 Hz. Increasing the stiffness by a factor of 100 (thick solid curves) produced very little additional attenuation. Therefore the thin solid curves provide an estimate of the largest attenuation that reasonably could be expected. VOR data in humans appears to be consistent with the baseline (thick solid) curves in Fig. 11 (Aw et al. 1996), but once again this comparison is indirect. Irrespective of uncertainties in model parameters, it seems clear that the effective stiffness and structural integrity of the perilymphatic space are important factors. At 1 Hz, for example, decreasing the compliance changed the attenuation factor nearly two orders of magnitude from ~0.01 to 1. A bone dehiscence or a surgical procedure, which increases the compliance of the perilymphatic cavity or middle ear, would be expected to decrease the effectiveness of canal plugging.

ROLE OF LINEAR ACCELERATION. Model results also indicate that the degree of cupular attenuation caused by canal plugging may be sensitive to linear accelerations of the head and/or eccentricity of the axis of rotation. Sensitivity of the semicircular canals to linear accelerations is a long standing question that has not yet been resolved (Benson 1974; Estes et al. 1975;
Goldberg and Fernández 1975; Ledoux 1949; Lowenstein 1970; Ross 1936). Part of the difficulty in experimentally testing linear sensitivity of semicircular canal afferents is due to the influence of opening the perilymphatic space—a significant factor in the present model predictions. To estimate the possible influence of linear accelerations on chronic plugged-canal responses, we applied the model for various axes of rotation located eccentric to the center of the canal in the human. Model results are shown in Fig. 12 for the right HC in response to sinusoidal rotations about five different axes (0–4). At 2 Hz, for example, plugged canal cupular responses were predicted to differ significantly depending on the location of the axis of rotation. Results indicate that linear acceleration may be an important parameter to be controlled in the measurement of plugged-canal responses.

Discussion of plugged-canal afferent responses

Acute damage to the sensory apparatus can be caused by occlusion

Another factor contributing to differences between afferent responses in acute versus chronic plugged-canal preparations may be the condition of the cupula. In the present study, individual afferents were monitored while slowly compressing the duct against the cartilaginous substrate using a series of discrete indentation steps. Afferent modulation was maintained only if compression of the canal proceeded slowly over the course of ~3–5 min (see Fig. 2A). Plugging at faster rates always resulted in loss of canal sensitivity to rotational stimuli. Following the approach of Hillman (1974), injection of alcaine dye into the endolymph revealed a detachment of the cupula at the apex in unresponsive canals. As Hillman suggested, cupular detachment appears to serve as a “relief valve” to accommodate excess transcupular pressure—in this case, relief for pressure generated by compression of the duct during the mechanical plugging procedure.

Compression of the endolymphatic duct is a stimulus that produces controllable semicircular canal afferent responses and has been used to mimic angular motion of the head (Dickman and Corriera 1989; Rabbitt et al. 1995). For sinusoidal stimuli at 1 Hz, a ±1-μm indentation mimics about ±4°/s head velocity in the toadfish (Rabbitt et al. 1995). The stimulus is nearly linear, such that doubling the magnitude of indentation doubles the magnitude of afferent responses. On the basis of this experimentally established relationship, rapid compression of the duct during a short time course, several seconds for example, would generate pressures and cupular displacements several orders of magnitude larger than those generated by natural physiological head movements. Consider, for example, the rapid increases in firing rate appearing at times 450 and 650 s in the record of Fig. 2A. This low-gain afferent had a measured gain of 0.3 spikes/s per °/s to rotation; therefore an increase of ~50 spikes/s corresponds to an equivalent angular velocity ~180°/s. Each step in the compression was carried out during

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**FIG. 11.** Influence of cochlea/middle-ear stiffness. Primate model predictions were sensitive to the overall stiffness of the vestibule as influenced by the connection to the middle ear and/or any compromise of the temporal bone associated with the surgical procedure. Predicted changes in the cupular attenuation (A) and phase (B) caused by altering the stiffness by 2 orders of magnitude up (thin solid curves) and 2 orders of magnitude down (dotted curves) are shown for the human model. Note that plugging was predicted to become much less effective in attenuating canal responses if the stiffness of the bony cavity is compromised.

**FIG. 12.** Sensitivity to eccentricity of rotation. The cupular attenuation (A) and phase shift (B) caused by canal plugging were predicted to depend on the axis of rotation owing to linear acceleration sensitivity. Predictions are shown for rotations about the neck (thick solid curves), ipsilateral canal (thin solid curves), nasal offset (dotted curves), occipital offset (dashed curves), and contralateral canal (dot-dashed curves) in the human model.
an \(\sim 10\)-s period, which provides an equivalent angular acceleration of \(\sim 18^\circ/s^2\) and a compression rate of \(\sim 4-5\) \(\mu m/s\). Had the same partial compression of the canal been completed in 1 s, the equivalent angular acceleration would approach \(\sim 8000^\circ/s^2\). Complete compression of the canal in 1 s would increase this value by more than an order of magnitude. Angular head accelerations and velocities of this magnitude generally result in trauma (Klinich et al. 1996). This is probably why rapid plugging caused cupular detachment.

The present data indicate that cupular damage probably occurs in most conventional plugging procedures and is inevitable during rapid compression of the canal. Therefore the question of cupular regeneration/repair becomes important in tracking recovery and adaptation after plugging procedures. Although the present study does not address the cupular regeneration process, on several occasions, we noted an absence of rotational response in the canal nerve \(\leq 6\) h after mechanical labyrinthine trauma. On the basis of the extracellular-mucopolysaccharide structure of the cupula and variable extent of damage, complete regeneration may be on the order of several days to weeks (Béranger 1961; Dohlman 1960; Hillman 1974; Silver et al. 1998). The slow compression approach used here to occlude the canal maintains integrity of the cupula and thereby allowed for acute study of afferent response dynamics without a functional recovery of the cupula being an issue. When using a more conventional, rapid surgical approach, a reasonable expectation would be for afferents to reproduce the present data after cupular recovery.

**Afferent response dynamics in the occluded condition**

Present experimental results show that HC afferents recorded in the acute condition continue to modulate their firing rate in response to sinusoidal head rotations even after the endolymphatic duct has been blocked completely. Because of the isolation of the HC nerve at the location of axon recordings, there is no doubt that the reported afferents supply the hair cells of the HC crista (Boyle et al. 1991). Injection of alcine dye into the HC ampulla after the experiment left no doubt that the endolymphatic duct was completely plugged. Firm compression of the duct against the cartilaginous substrate by the glass rod, and the use of rigid fixtures, served to minimize any possible stimulus due to mechanical movement of the rod relative to the fish. Plugged-canal responses also were observed for mechanical indentation of the utricle in the complete absence of rotational stimuli. As a separate control, at the end of several experiments an individual afferent was recorded while the glass rod plugging the canal was raised, thus removing the contact between the rod and the canal, and no detectable difference in the response was observed for the examined 5–10 cycles of rotation; the canal was inspected quickly and a 1.3-mm segment of the limb was observed to remain completely occluded. Therefore the possibility of artifact is remote.

Afferent responses observed in the toadfish HC after plugging raise the question concerning why the procedure appears to have reasonable efficacy in humans and in some experimental studies of the VOR. At least four factors may contribute to this. The first and most obvious factor is the condition of the preparation. In the present experiments, a portion of the perilymphatic space was opened surgically to allow for access to the HC nerve and endolymphatic duct. This contrasts the chronic plugged-canal case where, presumably, the perilymphatic space is sealed in rigid bone the entire plugged region becomes ossified. The model indicates that this difference could account for about a 10-fold reduction in cupular displacements in the chronic condition over and above the reduction present in the acute condition, at least in the low- to midfrequency range (see Fig. 6). A second contributing factor may be the power spectrum of the stimulus. Occlusion of the duct was shown here to be highly effective for low-frequency stimuli even after opening the perilymphatic space—it is predicted to become even more effective for an uncompromised perilymphatic space. This does not extend to high frequencies where the model predicts robust canal afferent responses in both acute and chronic conditions. Therefore plugged-canal responses to stimuli containing high-frequency components should not a priori be attributed to the emergence of an otolith afferent input or central adaptive mechanism. When comparing results from different species or attempting to predict afferent response after the plugging procedure, it is important to note that canal plugging was predicted by the model to be more effective in species having short interlabyrinth distances, stiff membranous ducts, and relatively stiff perilymphatic space and/or connection to the middle ear (such as the squirrel monkey). A third contributing factor may be differences in sensitivity of various afferent types to angular motion stimuli—high-threshold units may be the most susceptible to attenuation caused by plugging. For the toadfish, they are the LG afferents, and their counterparts are present in other vertebrates, e.g., the regularly discharging canal afferents in monkeys (Goldberg and Fernández 1971b). This may cause a change in the population of responding afferents participating in a particular vestibular reflex favoring the more sensitive high-gain units after plugging. Given differential projections of various classes of afferents, plugging may not act uniformly across all systems. Also plugged-canal responses become phase advanced by \(\sim 90^\circ\) in the midband relative to controls. The additional phase advance and the skewing of the entire population toward low-threshold units would both serve to reduce the angular velocity-sensitive inputs to the brain stem in favor of angular acceleration and jerk-sensitive inputs. It is unclear what influence this might have on the central processing and performance of vestibular reflex systems. In the squirrel monkey, for example, adaptation of the VOR after plugging apparently would require an additional central integration of canal inputs and/or the use of inputs from other organs to be effective. Without central adaptation, a significant phase lead and frequency sensitivity in the slow-phase VOR would be expected, at least at low frequencies (see Figs. 3 and 6). There is some experimental evidence supporting these possibilities, in that phase leads and high-frequency recovery have been observed in animals with plugged canals (Angelaki et al. 1996; Baker et al. 1982; Broussard and Bhatia 1996; Lasker et al. 1997; Yakushin et al. 1997). Two additional factors may contribute to the long-term dynamics of afferent responses in plugged canals. First, it is not known to what extent adaptation of the end organ itself might remodel and further modify afferent responses in chronic preparations. Second, because most plugging procedures damage/dislodge the cupula, recovery also must include the influence and time course of cupular regeneration.

It is important to reiterate that the residual afferent responses
observed in plugged semicircular canals are significant only at high stimulus frequencies. The specific frequency above which canal plugging becomes ineffective depends on several factors specific to the particular species and experimental/surgical approach. To provide some general guidelines—present results indicate that endolymphatic duct plugging reduces cupular displacement by \( \geq 100 \)-fold in the surgically opened acute preparation only at stimulus frequencies less than \( \sim 0.2 \) Hz. Results further indicate that complete canal plugging reduces cupular displacement by \( \geq 100 \)-fold in the ossified and sealed chronic preparation only at stimulus frequencies less than \( \sim 1 \) Hz. Given sensitivities to species-dependent morphological structure, surgical approach, and experimental design (see Figs. 6 and 9–12), it would not be surprising to see relatively large variability in the efficacy of canal plugging particularly for high stimulus frequencies.

**APPENDIX: ELASTO-HYDRODYNAMIC MODEL**

A finite difference approach was employed by dividing the membranous canal and perilymphatic space into \( N \) short discrete segments. Each segment \( n \) was assigned a different cross-sectional area and three-dimensional spatial location. Schematics of the perilymphatic and endolymphatic segments are illustrated in Fig. A1 along with model notation. The momentum equation providing the relationship between the endolymphatic volume displacement \( Q^e_n \) (cm\(^3\)) at cross-section \( n \) to the streamwise endolymph pressure gradient can be written (Damiano and Rabbitt 1996)

\[
L^e_n[Q^e_n] = P^e_{n+1} - P^e_n + f^e
\]

In this, the operator \( L^e_n \) written in Fourier domain is

\[
L^e_n = \frac{\Delta t}{\bar{\Lambda}^n} \left( i \gamma_n \bar{A}_n^e - \omega \rho_n^e \bar{A}_n^e \right) + i \alpha m_n \bar{\Lambda}_n^e \tag{A2}
\]

where \( \rho_n^e \) (g/cm\(^3\)) is the density, \( \mu_n^e \) (dyne-s/cm\(^2\)) is the absolute viscosity, \( \gamma_n^e \) (dyne/cm\(^2\)) is the modulus of rigidity of the material/ fluid in segment \( n \), \( i \) is \( \sqrt{-1} \), \( \omega \) (rad/s) is the excitation frequency, \( \bar{\Lambda}_n^e \) is the local cross-sectional area of the membranous duct, \( P^e_n \) (dyne/cm\(^2\)) is the local perturbation in the endolymphatic pressure, and \( f^e \) (dyne/cm\(^2\)) is the endolymphatic inertial forcing. The superscript “\( e \)” denotes the endolymph and the subscript “\( n \)” denotes the finite-difference element number. Cross-sectional areas are provided in Fig. A2.

The coefficient \( \lambda_n \) accounts for the frequency-dependent shape of the velocity profile across the cross-section. For Poiseuille flow in the endolymphatic duct, which is valid for low-stimulus frequencies, \( \lambda_n \sim 8 \pi \), which is the exact value for steady flow in a circular tube. The functional dependence of \( \lambda_n \) on the local Stokes (\( S_l \)) number is provided in the following text, and its dependence canal ellipticity by Oman et al. (1987). In the present model, the parameter is \( \lambda_n \) computed as a function of local diameter, excitation frequency, and material properties. Cross-sectional areas are assumed.

**Equation** A2 describes the volume displacement of a Kelvin-Voigt viscoelastic material, which reduces to the linearized Newtonian fluid for the case of zero stiffness and to a simple linear-elastic solid when the viscosity is set to zero (Fung 1990, 1981). We take this model to apply to both the endolymph and the cupula and adjust the modulus of rigidity and the viscosity appropriately to account for the differing properties of the materials (see Table 1).

The inertial forcing term, \( f^e_n \) in Eq. A2, is due to angular head acceleration and is computed for each element using

\[
f^e_n = \rho_n^e \frac{d}{dt} \left( \frac{d}{dt} (\hat{\Omega} - \hat{\Omega}^n) \times \vec{R}_n \right) \cdot \Delta \vec{r}_n
\]

The independent variable \( \hat{\Omega} \) is the angular position of the head and the dependent \( \hat{\Omega}^n \) variable is the angular deflection of the membranous duct relative to the head. Note that \( \hat{\Omega}^n \) would be zero if the membranous duct moved rigidly with the head. \( \vec{R}_n \) is the vector from the center of rotation to the center of the endolymph fluid segment \( n \), and \( \Delta \vec{r}_n \) is along the fluid segment in the direction tangent to the local canal centerline. This expression was obtained following the Galilean transformation described by Damiano and Rabbitt (1996) that attaches a moving coordinate frame to the head and allows fluid and cupular displacements to be computed relative to the head. Because the present model allows for some angular movement of the membranous labyrinth relative to the head, the transformation was carried out for the endolymph relative to the angular position of the moving endolymphatic duct.

**FIG. A1.** Model schematic and notation. Membranous duct was divided into \( N \) short cylindrical elements having local cross-sectional area \( A_n^e \) and the osseous duct into \( N \) short annular elements of local cross-sectional area \( A_n^o \) (areas provided in Fig. A2). Membranous duct was tied to the bony duct by elastic connective filaments. Streamwise flow through each cross-sectional area is denoted \( Q^e_n \) for the endolymph and \( Q^o_n \) for the perilymph. Corresponding pressures are denoted \( P^e_n \) and \( P^o_n \). Local volumetric expansion of the membranous duct is denoted \( q^e_n \) and any expansion of the osseous canal is denoted \( q^o_n \). Inward volume flows caused by the indentation of the perilymph and endolymph are denoted \( v^e_n \) and \( v^o_n \), respectively. Angular acceleration of the head \( \hat{\Omega} \) is shown in the positive, HC excitatory, direction perpendicular to the canal plane. Vector \( \vec{R}_n \) runs from the center of rotation to the local centerline of the duct s.
valid because the motion of the head is much larger than the motion of the endolymph and/or perilymph—including these terms does not significantly change $m_d$ (<10\%).

The fine filaments (trabeculae) connecting the membranous duct to the osseous canal are modeled as an array of elastic fibers through which the perilymph is allowed to flow. Stiffness of the filaments is lumped into an effective shear modulus $\gamma$ occupying the perilymphatic space. This shear modulus is distinct from that of the perilymph in that it does not restrict perilymph movement—it only restricts movement of the membranous duct relative to the head. The effective stiffness is obtained by integrating the filament-derived elastic shear stress around the surface of the membranous duct. The result is

$$k^d = \int_{\theta_{\text{max}}}^{\theta_{\text{min}}} 2\pi \rho^p \left( \frac{\sigma'}{\sigma'' + \sigma'} \right) \left( \hat{\Omega} \times \hat{R}^p \right) \cdot d\hat{s}$$

The viscous drag caused by simple shear of the perilymph provides

$$c^d = \int_{\theta_{\text{max}}}^{\theta_{\text{min}}} 2\pi \mu\left( \frac{\sigma'}{\sigma'' + \sigma'} \right) \left( \hat{\Omega} \times \hat{R}^d \right) \cdot d\hat{s}$$

where we have neglected unsteady fluid effects in this term. The inertial forcing for the duct is

$$f^d = -\int_{\theta_{\text{max}}}^{\theta_{\text{min}}} \rho^d A^d \frac{d}{dt} \left( \hat{\Omega} \times \hat{R}^d \right) \cdot d\hat{s}$$

In addition to streamwise movement of the membranous duct, the difference between the pressure in the endolymph and that in the perilymph induces local distention of the membranous duct. To address this coupling, the duct wall was modeled as a viscoelastic membrane, immersed in a viscous fluid. For this model, the effective steric volume displacement, $q^d$ (cm$^3$), through a fixed control surface defining the undeformed duct segment is determined by

$$J^d[q^d] = P^d_n - P^d_o$$

where $P^d_n$ and $P^d_o$ is the local transmembrane pressure gradient acting across the membranous duct. These pressures are the same values that appear in the streamwise momentum equations for the endolymph and the perilymph (Eq. A1 with superscripts "e" and "p" respectively). The Fourier operator $J^d_e$ is defined as

$$J^d_e = \{[\kappa^e_s - \omega^d \eta^e_s] + i\omega^e \xi_s\}$$

In this expression, $\eta^e_s$ is the effective mass, $\kappa^e_s$ is the effective stiffness, and $\xi_s$ is the effective viscosity resisting distention of the membranous duct. The values of these terms come from the physical properties of the membrane, the local geometry, as well as the entrainment of endolymph and perilymph caused by the motion of the duct as described in the following text.

The incompressible Navier-Stokes equations were used to estimate the effect of fluid entrainment by the moving membrane normal to its surface. This was done by assuming each segment acts as a point source generating a pure radial flow and integrating the linearized fluid equations in the direction normal to the membranous duct surface. The resulting effective mass is

$$\eta^d_e = \frac{\rho^d}{8\pi\Delta s_e} \left( 1 + \left( \frac{\sigma'_{\text{eff}}}{\sigma''_{\text{eff}}} \right)^2 \right)$$

where $\rho^d$ is the density of the fluid in the perilymphatic space. The effective stiffness $\kappa^d_{\text{eff}}$ was found by modeling labyrinth wall as a thin viscoelastic material with elastic modulus $E^d_{\text{eff}}$ and stiffness proportional damping. Assuming uniform tangential hoop stress through the thickness (Laplace’s law) $h^d_{\text{eff}}$ provides the effective stiffness.
Finally, the effective damping $\zeta^e$ was set equal to the proportional damping in the membrane plus the effect due to perilymph fluid entrainment. The entrainment term was determined again from the Navier-Stokes equations by assuming axisymmetric, incompressible, viscous flow outside of the membranous duct. Integrating in the radial direction, the pressure generated at the surface of the canal due to viscosity of the deforming perilymph fluid was determined. Combining terms gives the effective damping as

$$\zeta^e = \frac{\mu^p}{4\pi \Delta h} \left( \frac{1}{a_n^e} - \frac{1}{a_n^p} \right) + \beta_1 \kappa^e + \beta_2 \eta^e$$  \hspace{1cm} (A15)$$

where $\mu^p$ is the absolute viscosity of the perilymph, $\beta_1$ is the stiffness proportional damping coefficient and $\beta_2$ is the mass proportional damping coefficient.

Using a similar approach as applied for the membranous duct, the osseous perilymphatic canal was modeled using the Fourier operator

$$J_n^0 q_n = P_n^e$$  \hspace{1cm} (A16)$$

where $P_n^e$ is the pressure in the annular perilymphatic space and $q_n^e$ is the effective flow caused by any deformation of the perilymph duct. The bony duct also was modeled using Eqs. A12–A15 by replacing the superscript “e” with “p” and $\sigma_n^p$ with the effective outside radius of the perilymphatic space. Because the stiffness of the bony duct is very high, Eq. A16 serves to conserve the geometry and limit the effective radial flux of perilymph $q_n^e$. Conservation of mass requires the total volume of endolymph to remain constant and the total volume of perilymph to remain constant. Because of this, the model equations describe a dynamic redistribution of the perilymphatic and endolymphatic volumes—not a change in their volumes. The equations are equally valid for application to the normal case for a closed temporal bone or for plugged canals, which is not adequate for plugged canals, Eq. A1 would reduce to the outer region model of Damiano and Rabbitt (1996).

For the perilymph, conservation of mass provides

$$Q_n^p = Q_n^e + \sum_{k=1}^{n} (q_k^e - q_k^e)$$  \hspace{1cm} (A17)$$

where the term $v_n^e$ is the inward volume flow caused by mechanical indentation of endolymphatic segment $k$, which is zero for all segments that are not acted on by mechanical indentation and is zero during normal head rotation. Note that if the duct was perfectly rigid, then the endolymph volume displacement at each cross-section $n$ would be equal to the volume displacement of the cupula $Q_n^c = Q_n^e$. In this special case, which is not adequate for plugged canals, Eq. A1 would reduce to the outer region model of Damiano and Rabbitt (1996).

For the perilymph, conservation of mass provides

$$Q_n^p = Q_n^e + \sum_{k=1}^{n} (q_k^e - q_k^e)$$  \hspace{1cm} (A18)$$

where the term $v_n^e$ is the inward volume displacement caused by any mechanical indentation of perilymphatic segment $k$. During physiological conditions, $v_n^e$ is zero, but the term is included here to study the additional influence of indentation of the perilymph duct in future work. Because both the perilymphatic and endolymphatic ducts form closed fluid spaces, conservation of mass also requires that the sum around each loop is zero. Hence

$$\sum_{k=1}^{N} (q_k^e - q_k^e) = 0$$  \hspace{1cm} (A19)$$

and

$$\sum_{k=1}^{N} (q_k^e - q_k^e) = 0$$  \hspace{1cm} (A20)$$

The preceding equations (A1–A20) were solved for the streamwise volume flow, the local distention of the duct, and the pressure as a function of position within the horizontal canal. In the analysis, we were particularly interested in the spatial distribution of field variables within the horizontal canal. The contributions of the AC and PC were included by lumping their respective distentional impedances at HC bifurcation points. These point impedances were estimated to be equal to that of the slender HC duct plus one half of the utricular region. The same was done for the perilymphatic connection to the scala vestibuli. Half of the impedance for each endolymphatic canal was lumped at the common crus and the other half at the bifurcation from the HC.

In the limiting case when the membranous canal is infinitely stiff, the preceding equations become singular and the hydrostatic pressure cannot be determined. Through a fortuitous selection of dependent variables, we were able to avoid numerical difficulties arising in the nearly singular case. This was done by recasting the equations in the following matrix form

$$M \dot{\psi} = \ddot{\iota}$$  \hspace{1cm} (A21)$$

where the matrix $M$ is noted in the following text and contains model coefficients computed from the morphology and physical parameters using the preceding equations. When written in this form, there are $2^n N + 5$ equations, where $N$ denotes the number of discrete endolymphatic and perilymphatic canal segments used in the model ($n = 100$ for the present results). The dependent variables are

$$\psi = \left\{ \begin{array}{ll}
Q_n^p & \text{if } n = 1 \\
Q_n^e & \text{if } n = 2 \\
P_n^e & \text{if } n = 3 \\
P_n^p & \text{if } n = 4 \\
q_n^e & \text{if } 5 > n \geq 4 + N \\
q_n^e & \text{if } 4 + N > n \geq 4 + 2N \\
\Omega^d & \text{if } n = 5 + 2N \\
\end{array} \right.$$  \hspace{1cm} (A22)$$

The first dependent variable, $Q_n^e$, is the cupular volume displacement; the second, $Q_n^p$, is the streamwise perilymphatic volume displacement at the ampulla; the third, $P_n^e$, is the absolute endolymphatic pressure in the ampulla on the HC lumen side of the cupula; and the fourth, $P_n^p$, is the absolute perilymphatic pressure just outside of the HC ampulla. The last dependent variable, $\Omega^d$, is the angular displacement of the endolymphatic duct relative to the moving skull. The remaining dependent variables are intermediate sums of endolymphatic and
perilymphatic duct distentional volumes. The forcing vector appearing in Eq. A21 is

\[
\sigma^f = \begin{cases} 
\sum_{k=1}^{N} (f^i_k + L_1^i [\nu_k]) & \text{if } n = 1 \\
\sum_{k=1}^{N} (f^i_k + L_2^i [\nu_k]) & \text{if } n = 2 \\
0 & \text{if } 3 \leq n \leq 4 \\
\sum_{k=1}^{N} (L_{k-4}^i [\nu_k] - L_{k-5}^i [\nu_k] + f^i_k - f^i_0) - f^i_0 & \text{if } 5 \leq n \geq 3 + N \\
\sum_{k=1}^{N} (L_{k-2}^i [\nu_k] + f^i_k) & \text{if } n = 4 + N \\
\sum_{k=1}^{N} (L_{k-2}^i [\nu_k] + f^i_k) & \text{if } n = 5 + 2N 
\end{cases}
\] (A23)

where \( f^i_k \) is the inertial force associated with the endolymphatic segment \( k \), \( f^i_0 \) is the inertial force associated with the membranous duct. The remaining forcing terms are due to mechanical indentation of the membranous duct (\( r \xi \)), and the perilymphatic space (\( v_\nu^i \)), respectively.

Elements of the matrix \( M \) and forcing vector \( \xi \) were determined for each segment using the physical parameters provided in Table 1. The matrix system then was solved using LU decomposition at 40 discrete frequencies equally spaced on a log scale from 0.003 to 30 Hz. The endolymphatic and perilymphatic pressures are included implicitly as functions of the centerline coordinate “s” in these equations. Once results were computed by inversion of Eq. A21, the spatial distribution of pressure then was computed.

Most of the parameters appearing in the model are relatively well known in terms of the morphology of the different labyrinths and the physical properties of the fluids (see Table 1). The density and viscosity of the endolymph and perilymph are nearly equal to water (Steer et al. 1967). Because the cupula has neutral buoyancy when suspended in endolymph, its effective density was set equal to the density of the endolymph. The modulus of elasticity for the cupula was estimated by matching the mechanical lower-corner frequency reported previously for the toadfish (Highstein et al. 1996; Rabbitt et al. 1996), which yields a stiffness on the same order as other mucopolysaccharides (Fung 1981; Lutz et al. 1973; Philipoff 1966). Cupular viscosity was estimated from Yamauchi et al. (1998).

The physical properties of the membranous duct were estimated by measuring the pressure in the HC ampulla during mechanical indentation of the canal limb (Yamauchi et al. 1998). The present model assumes circular cross-sections and hence does not include bending stiffness. The magnitude of structural damping within the membranous duct has not been measured to date. Proportional damping coefficients have been estimated for similar biological structures, and the present estimates are based on these data (eardrum: Funnell et al. 1987; skin: Wilkes et al. 1973). Present results are relatively insensitive to changes in these structural damping parameters.

The stiffness of the osseous canal was estimated using the elastic modulus of bone, which is very high relative to the stiffness of the membranous duct (Fung 1981). Because of this, the perilymphatic duct was essentially rigid in the present model and prevented any flow of perilymph through the surface. In such, model predictions are insensitive to variations in the perilymphatic duct wall properties.

**Cross-sectional velocity profiles**

Following Damiano and Rabbitt (1996), the local particle displacement of the fluid \( u(s, r, \theta) \) is governed by

\[
\rho \frac{\partial^2 u}{\partial t^2} - \nabla \cdot \left( \mu \frac{\partial u}{\partial t} + \gamma u \right) = f(s)e^{it\omega} \tag{A24}
\]

where \( r \) is the fluid density, \( m \) is the absolute viscosity and \( f(s) \) is the local pressure gradient. This is based on the asymptotic analysis of Damiano and Rabbitt (1996). We have added the additional influence of material shear stiffness \( \gamma \). The values of density, viscosity, and stiffness are adjusted as a function of position around the HC duct that match the actual material properties in the endolymph, perilymph, or cupula.

Viscous drag and inertial effects in the endolymph are approximated using circular cross-sections. In this case, the dependence of the velocity on the local radial cross-sectional coordinate \( r \) is easily found by a Fourier-Bessel expansion to be

\[
\frac{\partial u}{\partial t} = B(s)e^{it\omega} \sum_{n=1}^{\infty} \frac{J_n(1, J_0)}{(\beta_n^2 - i\beta_n^2 K_0 + i\beta_n^2 J_0)(\beta_n J_0)} \tag{A25}
\]

where the nondimensional Stokes number, \( S_t = \sigma^2 \rho \alpha^2 \mu \), is a measure of the unsteady inertia relative to the viscous drag, the nondimensional parameter, \( K = \gamma / \omega \mu \), is a measure of the elastic restoring force relative to the viscous drag, and \( B(s) \) is a complex-valued function of position. In this expression, \( J_n = J_n(\beta_n r/\sigma) \) are zero-order Bessel functions of the first kind, \( \beta_n \) are the eigenvalues required to match the no-slip boundary condition at the duct wall, \( r \) is the local radial distance from the centerline of the duct, and \( s \) is the local cross-sectional radius. The inner product \( \langle f, g \rangle \) is defined as the integral over the local cross-sectional area.

The same approach is applied for the perilymph but modified to account for the annular space. Viscous drag and inertial effects in the perilymph are approximated by replacing \( J_0 \) in Eq. A25 with \( J_{0,\nu} \) or \( J_{0,\nu} = \alpha_\nu Y_{0,\nu} \), where \( Y_{0,\nu} \) are zero-order Bessel functions of the second kind. The eigenparameter, \( \alpha_\nu \), is determined to meet the no-slip boundary condition on the outside surface of the membranous duct and on the inside surface of the osseous canal.

Once fluid velocity distribution is known, the flow rate \( Q_n \) at cross-section \( n \) is

\[
Q_n = \int \int_{A_n} \frac{\partial u}{\partial t} dA \tag{A26}
\]

The magnitude and phase of the viscous shear stress acting on the wall relative to the volume flow rate is determined by the parameter \( \lambda \):

\[
\lambda = \frac{\pi}{\int_{A_n} \frac{\partial u}{\partial \eta} d\xi} \int_{A_n} \mu \frac{\partial u}{\partial \eta} dA \tag{A27}
\]

where \( \Lambda_n \) is the cross-sectional area, \( \partial A_n \) is the periphery, \( \xi \) is a coordinate tangent to the periphery of the cross section, and \( \eta \) is a coordinate normal to the surface.
**M matrix**

$$\mathbf{M} = \begin{bmatrix}
\sum_{k=1}^{N} L_k^2 & 0 & 0 & 0 & L_k^1 & L_k^3 & \cdots & L_k^N & 0 & 0 & \cdots & 0 & f_k^1 \\
0 & \sum_{k=1}^{N} L_k^2 & 0 & 0 & -L_k^1 & -L_k^3 & \cdots & -L_k^N & L_k^1 & L_k^3 & \cdots & L_k^N & f_k^1 \\
0 & 0 & -1 & 1 & 0 & J_k^1 & \cdots & 0 & 0 & 0 & \cdots & 0 & 0 \\
0 & 0 & 0 & -1 & 0 & 0 & \cdots & 0 & 0 & 0 & \cdots & 0 & 0 \\
-\sum_{k=1}^{N} L_k^2 & \sum_{k=1}^{N} L_k^2 & -1 & 1 & -L_k^1 - L_k^3 & -L_k^1 - L_k^3 & \cdots & -L_k^1 - L_k^3 & L_k^1 & L_k^3 & \cdots & 0 & 0 \\
\vdots & \vdots & \vdots & \vdots & \vdots & \vdots & \cdots & \vdots & \vdots & \vdots & \cdots & \vdots & \vdots \\
0 & 0 & 0 & 0 & 0 & 0 & \cdots & 1 & 0 & 0 & \cdots & 0 & 0 \\
0 & -L_k^1 & 0 & -1 & L_k^1 & L_k^3 & \cdots & 0 & -L_k^1 & -L_k^3 & -L_k^3 & \cdots & \vdots \\
\vdots & \vdots & \vdots & \vdots & \vdots & \vdots & \cdots & \vdots & \vdots & \vdots & \cdots & \vdots & \vdots \\
0 & 0 & 0 & 0 & 0 & 0 & \cdots & 0 & 0 & 0 & 1 & 0 & \cdots \end{bmatrix}$$

**Pressure distribution**

Once the matrix equations are inverted, the pressure distribution around the endolymphatic duct is computed using

$$P_n^e = P_n^e + \sum_{k=0}^{N} (L_k^1 Q_k^e + q_k^e - \nu_k^e - f_k^e)$$  \hspace{1cm} (A29)

Similarly, the pressure distribution in the perilymph is computed using

$$P_n^p = P_n^p + \sum_{k=0}^{N} (L_k^1 Q_k^p + q_k^p - \nu_k^p - f_k^p)$$  \hspace{1cm} (A30)

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