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Low-Power Sensing for Vestibular Prostheses

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***Abstract*—This paper describes a novel sensing approach for reducing power requirements of implantable vestibular prostheses. A passive, microfabricated polymeric inertial sensor for detecting angular head rotations based on the biomechanics of the human semicircular canal is described. Angular head motion is coded by deflection of a highly compliant capacitor plate placed in parallel with a rigid reference electrode. This capacitance change serves to detect instantaneous angular velocity along a given axis of rotation. Designed for integration with a microelectromechanical systems-based fully implantable vestibular prosthesis, this sensing method can provide substantial power savings when compared with contemporary gyroscopes.**

I. INTRODUCTION

THE emerging area of vestibular prostheses provides a need to drastically reduce system power consumption. In this class of next generation sensory replacement implants, otherwise absent head angular and linear motion cues are replaced by direct and selective activation of nerve elements in the peripheral vestibular system. Vestibular prostheses are an essential therapeutic intervention for some individuals suffering from bilateral vestibular hypofunction. These individuals do not benefit from vestibular rehabilitation exercises designed to promote the central nervous system to adapt to the vestibular deficit; thus they currently have no therapeutic alternatives. Current development of vestibular prostheses relies upon seminal work of Goldberg and Fernandez revealing the coding of instantaneous head angular velocity by the semicircular canals (SCCs) as a discharge rate of afferent neural fibers [1],[2].Early studies by Suzuki, Cohen *et al.* illustrated that electrical stimulation of canal-specific nerves results in the coordinated activation of extraocular muscles to produce compensatory eye movement [3],[4]. Building upon these fundamental studies, Gong and Merfeld developed the first one-dimensional unilateral system relying upon a conventional gyroscope to sense horizontal head rotations and demonstrated partially compensatory eye movement in animals when stimulated with charge-balanced electrical pulses [5],[6]. Extending the ability to a three-dimensional (3-D) system Della Santina *et al.* further validated the efficacy of electrical stimulation of canal-specific vestibular nerves in mediating the vestibuloocular reflex about the axis of interest [7],[8].

Although these results are very promising and rely upon similar system-level implementations (Fig. 1) there remain significant challenges to making a fully implantable vestibular prosthesis a reality. First, the angular rotation sensors (gyroscopes) consume nearly 90 percent of the system power [9]. Second, current spread during excitation is both inefficient and results in spurious excitation of nearby branches of distinct vestibular nerves [7]. Others have attempted to improve performance through current steering to reduce spurious excitation [10], as well as implementing signal processing methods to preemptively correct for potentially erroneous stimulation resulting from poor specificity in excitation [11]. One group has attempted to reduce size by integrating the sensor and signal processing [12], Our approach focuses on directly on reducing the power consumed by the inertial sensors. By modeling our sensor after the human SCCs, we are pursuing a novel approach whereby the sensor is inspired by the

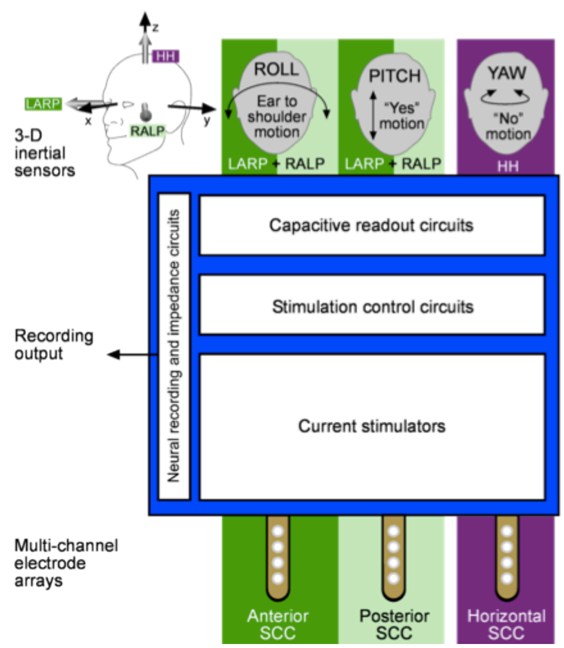

Fig. 1. Vestibular Prosthesis Block Diagram. Three custom angular rotation sensors capture head motion in 3-D. Due to the orientation of the canals, a unilateral vestibular system should sense rotations about three main axes: (1) horizontal *HH*, (2) left anterior and right posterior *LARP*, and (3) right anterior and left posterior *RALP*. Based on the detected angular velocity, frequency modulated, biphasic, charged-balanced current pulses would be generated. The stimulation control circuits will determine the current stimulators activated to realize the target current and the target nerve fibers.

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biomechanics of the human peripheral vestibular apparatus. This has the potential to drastically reduce power since energy is input to the system by human motion and the sensor itself is a purely passive structure.

This paper describes our current efforts in developing the low-power angular rotation sensor and is organized as follows. In Section II we describe the design and fabrication of a prototype biomimetic angular rotation sensor. In Section III we compare the sensor's expected power consumption with commercial gyroscopes. In addition, we address our plans for future work and describe additional system components that we are developing to further reduce power consumption.

II. Biomimetic Angular Rotation Sensor

A. Semicircular Canal Biomechanics

Located in the peripheral vestibular system, three nearly mutually orthogonal SCCs serve as our angular rotation sensors. The fluid within each SCC acts as an inertial load and although membranous, each canal behaves as a rigid structure during intentional head motion. As the head experiences an angular acceleration, the fluid lags the canal and serves to deform a gelatinous membrane (cupula) housed within an enlargement within the canal (ampulla) that occludes each canal (Fig. 2). Effectively the cupula senses a pressure differential induced by angular acceleration. Over frequencies experienced during normal head motion, the canals integrate angular acceleration and provide an angular velocity signal expressed by a proportional cupular deflection [5], [14]. This signal is directly coded to a neural firing rate applied to the ampullary nerve afferent fibers innervating the canal. Used in an important reflex, the angular vestibuloocular reflex (aVOR), this signal plays an important role in stabilizing gaze during head rotations.

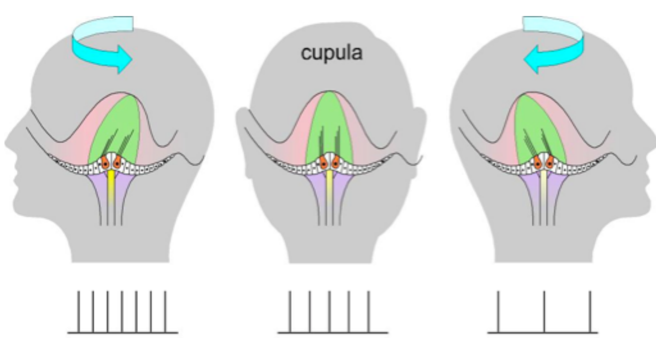


Fig. 2. Angular velocity coded by the left horizontal SCC as a neural firing rate. We estimate a baseline rate of 80 action potentials per second is exhibited by vestibular nerves in the absence of any angular rotation. The rate may vary between 0-200 Hz due to head rotations [1],[2],[4,][5],[7] .

Our design approach is to mimic the mechanical behavior of the natural SCC in detecting angular acceleration. We are developing a polymeric microfabricated membrane that mimicks the dynamics of the human cupula and SCC, a *MEMS-cupula*. Relying upon a second-order, heavily damped, torsional-pendulum model of the SCC [13],[14] the cupula and endolymph are treated as a linear system. The angular displacement of the cupula $\theta(t)$ is related to the input angular acceleration $\alpha(t)$ by the differential equation

$$I\frac{d^2\theta(t)}{dt^2} + B\frac{d\theta(t)}{dt} + K\theta(t) = I\alpha(t) \qquad (1)$$

where I is the effective movement of inertia, B is the viscous-damping couple and K is the elastic restoring couple (Fig. 3). Over the frequency range produced by normal head movement, the system exhibits a bandpass behavior with cupular displacement coding instantaneous angular velocity with essentially zero phase lag from approximately 0.1 Hz to 5 Hz.

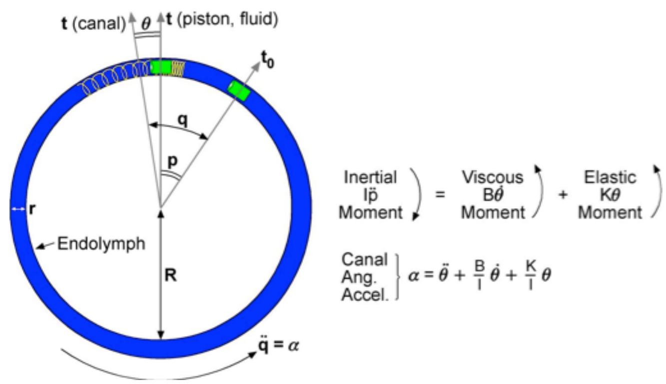


Fig. 3. Hydrodynamic SCC Model. Although the SCCs are sensitive to angular acceleration, the integrating nature of the canal provides an instantaneous angular velocity signal to the central vestibular system.

B. MEMS-cupula Design

To employ our sensor in a physiologically relevant domain and meet the requirements for the aVOR, we constrain our system's time constants to $\tau_2 = B/K = 0.32$ sec and $\tau_1 = I/B = 3.2$ msec [15]. When considering the hydrodynamic response of the SCCs, we can relate the two time constants to morphological and material parameters. For the shorter (faster) time constant:

$$\tau_1 = \frac{I}{B} = \frac{\rho r^2}{8\eta} \qquad r = \sqrt{\frac{8\eta\tau_1}{\rho}} = 160\,\mu m \qquad (2)$$

where r is the radius of the canal lumen shaped as a torus, ρ is the density of the fluid (1000 kg/m^3) in the canal, and η is the viscosity of the fluid (10^{-3} Ns/m^2). Endolymph is very similar to de-ionized water and we therefore select deionized water as the fluid in the lumen.

To obtain the longer time constant while preserving the differential pressure sensor role of the cupula, we use a microfabricated diaphragm that spans the canal lumen. It is modeled as a uniformly loaded circular clamped plate with a radius r. By designing the sensor to operate within the linear regime between pressure, P and deflection, w_o [16] we developed a relationship between τ_2 and the elastic resting couple of the membrane [17]:

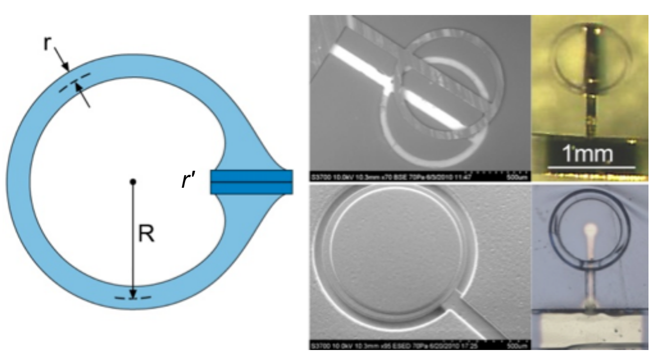
$$\tau_2 = \frac{B}{K} = \frac{3r^2\eta\pi R}{Eh^3} \qquad R = \frac{\tau_2 E h^3}{3r^2\eta\pi} = 2.6mm \qquad (3)$$

where h is the membrane thickness, E is Young's modulus, r is radius of the lumen, and R is the radius of curvature of the torus.

C. MEMS-cupula Fabrication

We selected PMMA (polymethyl-methacrylate) as the membrane material due to its desirable flexibility (E = 2 GPa). We then balanced the goal of achieving a small R, to remain within the space constraints for implantation [18], with the challenge of fabricating an ultra thin membrane (1 μm thick). Furthermore, to retain the flexibility of the diaphragm we selected a capacitive sensing method that requires the fabrication of both a sense and a reference electrode.

Understanding that a highly flexible membrane is a critical feature of our approach, we developed a scaled proof-of-concept MEMS-cupula using a PMMA substrate. Figure 4 illustrates the fabrication process. Utilizing a spun-cast SU-8 based micromolding technique for etchless micropatterning [19] a thin (1μm-thick) round (1 mm-diameter) membrane with a plated gold disk sense electrode was fabricated. Similar to the natural cupula, the sense electrode completely occludes the canal. The diameter was bounded by the constraint of employing easily available glass capillaries to cap the sensor and provide fluidic ports. A thick (30 μm) capacitor plated with a gold strip, sectioned to enable the flow of fluid through the electrode, serves as the reference electrode (Fig. 4). A tall "lip" (60 μm) on the sense electrode serves to mate the two plates, and a spacer (6 μm) on the sense electrode provides a gap between the two plates.



Left Center Right

Fig. 4. Biomimetic Sensor Prototype. Left: Illustration of the MEMS-cupula and the SCC-torus. The sensor's axis of rotation is out of the page. The MEMS-cupula bisects the SCC-torus and deflects in response to angular acceleration induced fluid motion. The radius of curvature (R) and the lumen radius (r) are indicated. Center: Scanning electron micrographs of the SU-molds used to define the respective reference (top) and sense (bottom) electrodes of the MEMS-cupula. Right: Photographs of the fabricated MEMS-cupula electrodes. A released reference electrode (top) is shown. The bottom picture illustrates an unreleased sense electrode. The metallization on the bottom of each structure enables signal transfer. The total capacitive sensor is formed by mating the reference with the sense electrode. This electrode has a "lip" and a groove for aligning and eventually sealing it with the reference electrode.

III. Discussion

A. Comparison with Contemporary Gyroscopes

We have presented a strategy for developing a passive angular rotation sensor for a vestibular prosthesis. Our motivation is to produce a low-power alternative to contemporary microfabricated gyroscopes. Today's micro-machined Coriolis vibratory gyroscopes sense angular rotation through a transfer of energy between drive and sense modes. Although such sensors follow an aggressive power-scaling trajectory [20], an inherent requirement is an input of power into the sensor to set up the vibratory mode. Moreover an even greater amount of power, on the order of *mW* is required to readout the signal [21].

Our passive, biomimetic angular rotation sensor is a radical, high-risk departure from traditional gyroscopes. Our sensing paradigm relies upon human motion to input power into the system via movement of an inertial fluid mass. Our sensor is purely a passive, rather than an active structure, and therefore the only power consumed is that of the readout circuitry. Based on our initial estimates of the necessary sensor interface circuitry, we expect power consumption to remain below 1 mW, even for a 3-D system. Another important advantage of our approach is that the bandpass behavior of the sensor is achieved in the mechanical domain. This is in contrast to the commercial gyroscopes that do not exhibit a low frequency roll-off and therefore require additional filtering in the electrical domain to achieve this characteristic.

B. Future Plans for Biomimetic Sensor Fabrication and Integration

Given the large dynamic range of the natural human system, it is difficult to achieve the desired bandpass operation with a single capacitive structure. Furthermore, the sense electrode moves into the non-linear regime P and deflection, w_o. Upon review of our initial model of the sensor, we note that the size of the MEMS-cupula is bounded to match to the size of canal. Our simulations suggest a larger sensor is desirable and will relax the constraints on the sense electrode thickness for large deflections. One way to achieve this goal is to maintain the inner radius of the canal while increasing the size of the MEMS-cupula as indicated by r' in Fig. 4. Such an approach is also employed in the natural human system and in other species. As another potential avenue to match the necessary dynamic range, we will explore filling the canal with a higher viscosity fluid as an additional control parameter.

Integration of the sensor with a model of the semicircular canal (SCC-torus) remains. We envision a shaped glass capillary serving this role. To provide fluidic access, attaching and sealing a glass capillary will introduce a port on the underside of each plate of the capacitive sensor. The ultimate goal is a completely micromachined structure consisting of the MEMS-cupula and SCC-torus thereby reducing the need for assembling sensor components.

C. Low-Power Signal-Processing and Stimulation Circuitry

To complement our work on low power inertial sensing we have also been exploring an ultra low power field programmable analog array (FPAA) for signal processing and current generation. Based on a floating-gate architecture, the FPAA circuitry provides biphasic currents from 10 μA to 530 μA in two percent increments on a single channel [22]. Using the FPAA is advantageous since it has demonstrated reduction in power consumption by over four orders of magnitude when compared with contemporary digital signal processing microprocessor implementations. Furthermore the FPAA is reconfigurable and contains the necessary amplifier circuitry to measure the compound action potentials (CAPs) of vestibular nerve branches, and measure electrode impedances. CAP measurements will become increasingly important for a class of vestibular prostheses that operate closed loop by monitoring CAPs and provide electrical stimulation to stabilize sporadic vestibular nerve activity experienced by patients suffering from debilitating Meniere's disease [23].

At the tissue interface, we are exploring a micromachined multichannel high-density array for stimulating ampullary nerve fibers. Based on prior experience with high-density cochlear electrode arrays [24],[25], we have developed polymeric arrays with stimulating sites of 180 μm diameter placed 250 μm center-to-center. Our goal is to reduce spurious cross-canal excitation through focused and selective neural excitation.

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