55.1: Feasibility Study on a Prototype of Vestibular Implant Using MEMS Gyroscopes

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Abstract
This paper presents a feasibility study on an MEMS-based implantable vestibular prosthesis. Our end-goal is to design a prosthesis that will replace the function of the damaged vestibular end-organ by providing an MEMS chip that will accurately sense, extract, and transmit 3-dimensional motion information for people who have permanently lost peripheral vestibular function. The prosthesis prototype includes orthogonal triads of accelerometers and gyroscopes on-a-single fingernail sized chip. Based on physilogical data on perceptual thresholds of linear acceleration and angular velocity in humans, we conclude that the MEMS technology is a viable candidate for such an implant. Functional architecture for the vestibular prosthesis is introduced.

Keywords
MEMS gyroscope, vestibular prosthesis, Inertial MEMS applications.

INTRODUCTION
The primary function of the vestibular system is to provide the brain with information about the body's motion and orientation. The absence of this information causes blurred vision (oscillopsia), balance difficulties, and spatial disorientation, vertigo, dizziness, imbalance, nausea, vomiting, and other symptoms often characterize dysfunction of the vestibular system. The symptoms may be quite mild, lasting minutes, or quite severe, resulting in total disability [1].

Sensory prostheses to artificially replace lost sensory function for a number of sensory systems are currently under investigation. For example, cochlear implants use electrical stimulation to restore hearing, providing some relief for patients suffering profound sensorineural hearing loss [2]. Using similar principles, a vestibular prosthesis could provide head orientation information to the nervous system for patients suffering from peripheral vestibular disorders. At least two categories of vestibular prosthesis might be considered. One approach is to provide the head movement information to the nervous system directly by electrically stimulating the vestibular neural pathways related to spatial orientation. Another approach is to provide the information via sensory substitution through other sensory systems (e.g., tactile, visual, auditory, etc.) [3]. This work falls in the first category. Our goal is to develop an implantable, vestibular neural prosthesis using electrical stimulation. To the best of our knowledge, there is only one group working on neural semicircular canal prosthesis [4, 5]. This group has already reported successful interface of the device with vestibular neurons. The group uses an off-the-shelf single axis piezoelectric vibrating gyroscope to measure the head rotation and to provide corresponding stimulus to the nervous system. The group reported experimental results on an animal model of guinea pigs, in which the gyroscope was mounted externally to the animal's skull using small bolts and miniature stainless-steel screws. Microcontroller was used to convert rotational information into electrical pulsatile stimulus.

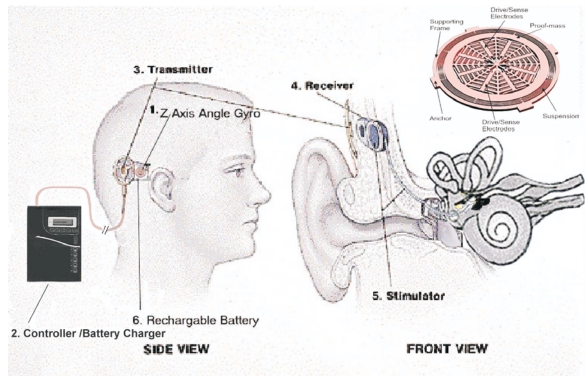

Figure 1. A conceptual model of a totally implantable vestibular prosthesis. The implant is based on 3-axes micro-size gyroscopes integrated alongside with signal conditioning electronics on the same silicon chip.

In contrast, our approach is based on a custom design of sensors using the MEMS technology. Using unique features of the MEMS technology it is possible to build extremely small sensors and potentially to integrate these sensors alongside with control electronics on the same silicon chip. We also combine two advanced technologies in micro-machined gyroscopes [6] and cochlear implants [2] to build the vestibular neural prosthesis. The objective of this research is to integrate on a silicon chip 3-dimensional gyroscopes, 3-

dimensional accelerometers, control electronics, signal processing unit, and electrical stimulator delivering the inertial information to the vestibular nerve. Figure 1 shows a conceptual model of this MEMS-based vestibular implant.

PHYSIOLOGICAL REQUIREMENTS

The vestibular system consists of three semicircular canals and two other structures, called the utricle and the saccule. The semicircular canals detect rotational head movements, while the saccule and utricle detect linear movements of the head. All of these organs have small sensory hair cells that send impulses through the synapses and nerves to the brain, where information about head movement is combined and interpreted with information from the eyes, muscles, and joints.

The rotational perceptual threshold in humans was determined to be between 0.1 and $2^0/s$ [7]. For example, the minimum threshold for an angular rotation sensation in humans was reported to be around $0.44^0/s^2$, but lower, $0.11^0/s^2$, when the oculogyral illusion has been used as the threshold measure [8]. Other reports estimated the perceptual threshold of the whole body angular movement at $1.5^0/s$, [7].

It should be noticed, however, that perceptual thresholds are different for different rates of acceleration and vary from person to person. Montandon [9] determined that the threshold is $1^0/s^2$ in healthy individuals, but greater than $6 - 7^0/s^2$ in patients with vestibular dysfunction. Threshold of sensitivity to the linear acceleration is defined by the acceleration near the sensitivity limit of the subject's otolith organs, which is about $6cm/s^2$ in most cases [10].

In summary, the lower limit of the human head rotation is $0.11^0/s^2$ for the angular acceleration and $0.5^0/s$ for the angular velocity, while for the translation sensation (or linear acceleration) is $6cm/s^2$ ($\sim 6[mili - G]$). The reported sensation limits set the sensitivity requirements for the vestibular prosthesis.

Another critical physiological parameter is the firing rate of nerons and relation of the firing rate to the head rotation/translation. The average firing rate of regular vestibular units has been reported as 60 spikes/s in the guinea pig [11] and 90 spikes/s in the squirrel monkey [12]. The firing frequency increases when a semicircular canal responds to rotation in one direction, and decreases in another direction. In the guinea pig, the average sensitibity is roughly 0.3 spike/s per $^0/s$ for regular afferents and 0.7 spike/s per $^0/s$ for irregular afferents [11]. These experimental results set the requirement for the pulse generating unit determining the rate of electrical stimulation in the vestibular nerve.

FUNCTIONAL BLOCKS

Figure 2 shows the functional diagram of the vestibulat prosthesis. The device includes three main functional units - a sensing unit, a pulse generator, and a stimulator. The device also includes two supporting units: a power supply and an external controller and charging unit.

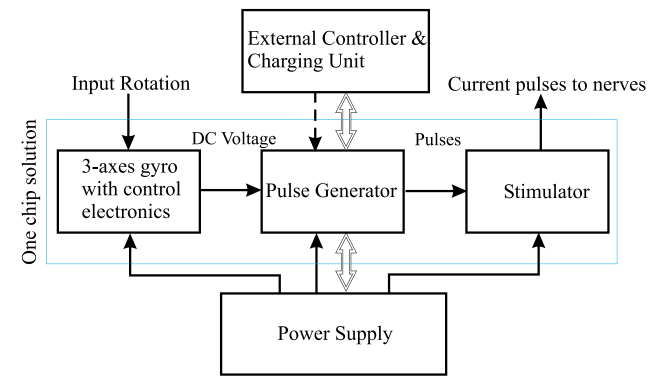

Figure 2. Functional diagram of the vestibular implant

The sensing unit includes 3-axis accelerometers and 3-axis gyroscopes. These devices sense linear and angular motion of the head and generate voltages proportional to the corresponding linear acceleration and angular velocity. Then, voltages are sent to the pulse generating unit where angular velocities (or linear accelerations) are translated into voltage pulses. In the stimulator, the voltage pulses are converted into current pulses and are delivered through specially designed electrodes to stimulate the corresponding vestibular nerve elements.

Each functional block of the chip consumes electrical power. For long term autonomous operation of the implantable device, it is required to supply power externally. Additionally, the implant should be able to communicate with an external controller and charging unit.

MEMS SENSORS

The purpose of the semicircular canal prosthesis is to restore balance function. Ideally, the prosthesis will be able to sense motion with sufficient precision and to deliver signals to the central neural system matching signals that the natural organ would generate.

We propose to implement the vestibular prosthesis using MEMS technology. An ensemble of six inertial MEMS sensors is required to measure six-degrees of freedom of the head motion on a single silicon chip and packaged in a volume smaller than 1 cubic centimeter. Micromachining can shrink the sensors size by orders of magnitude, reduce the fabrication cost significantly, and allow the electronics to be integrated on the same silicon chip [13].

Our first prototype of the vestibular prosthesis is implemented using polysilicon surface micromachining technology. In the prosthesis, the three semicircular canals are replaced by 3-axis MEMS gyroscopes, while the two otolith organs are replaced by 3-axis acceleromters, Fig-

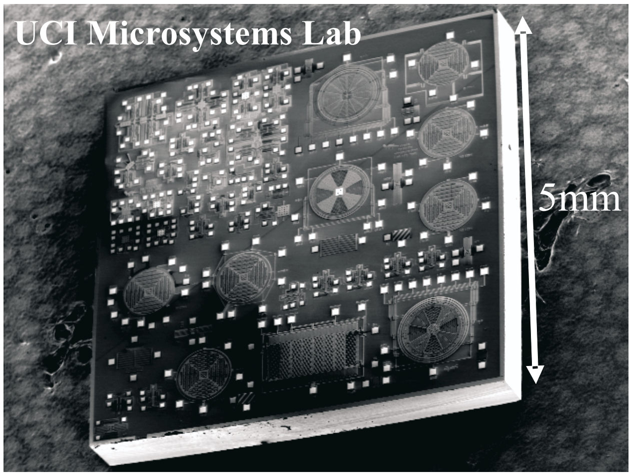

Figure 3. A prototype multi-sensor unit including accelerometers and gyroscopes. The experimental unit does not include electronics on the chip. The experimental chip is fabricated using JDS/Uniphase's MUMPs technology

ure 3.

The MEMS accelerometer consists of a proof mass suspended by compliant beams anchored to a fixed frame. External acceleration due to motion of the object to which the sensor's frame is attached, displaces the support frame relative to the proof mass, which in turn, changes the initial stress in the suspension spring. Both this relative displacement and the suspension-beam stress can be used as a measure of the external acceleration.

In the most general case, the proof-mass motion can have six degrees of freedom. But typically in a unidirectional accelerometer, the geometrical design of the suspension is such that one of these axes has low stiffness while high stiffness along other axes. For example, in case of the Z-axis accelerometer, the proof mass of the device will displace in out-of-plane of the chip only if there is an acceleration component along the z-axis.

Our micromachined gyroscopes use vibrating element to measure rotational velocity based on the Coriolis principle [14]. Just like a linear accelerometer, the micromachined vibrating angular rate sensor consists of a mass suspended on elastic supporting flexures anchored to the substrate, Figure 4.

In the basis of operation, the proof-mass, which constitute the active portion of the sensor, is driven by an oscillator circuit at a precise amplitude and high frequency, Fig. 5. When subjected to a rotation, the proof-mass will be subjected to the Coriolis force:

$$F = 2m\Omega \times V_c$$

where m - mass, V_c - instantaneous radial velocity of the center of mass, Ω - input rate.

As an illustrative example, consider a Z-axis gyroscope. The behavior of the gyroscope is naturally described

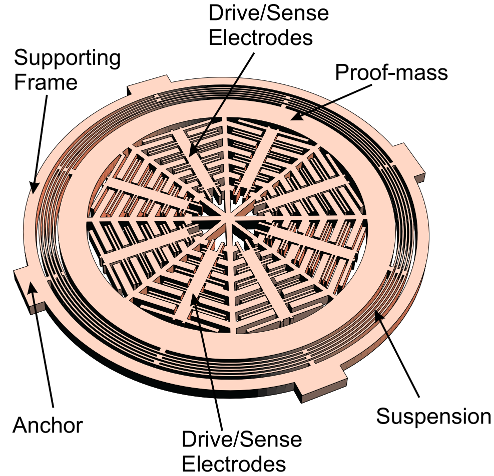

Figure 4. Schematic illustration of a MEMS implementation of the z-axis rate integrating gyroscope

with respect to the non-inertial coordinate frame $\{x, y, z\}$, Fig. 5. In this case, the governing equations in Cartesian coordinates $\{x, y, z\}$ are given by

$$\begin{aligned} \ddot{x} + \omega_n^2 x - 2\Omega\dot{y} = 0 \\ \ddot{y} + \omega_n^2 y + 2\Omega\dot{x} = 0 \end{aligned} \quad (1)$$

The essential feature of these equations is the presence of the Coriolis acceleration terms $-2\Omega\dot{y}$ and $2\Omega\dot{x}$. It is the Coriolis acceleration that causes a transfer of energy between the two gyroscope modes of operation.

The resultant Coriolis force is perpendicular to both the input rate and the instantaneous radial velocity in the drive direction. This produces a motion of the proof-mass in direction perpendicular to its initial oscillation. To measure rotation rate, the proof-mass is driven to a fixed amplitude along the x-axis by applying an electrostatic drive force to the proof-mass along the x-axis. In the absence of rotation there will be no motion of the proof-mass along the y-axis, Fig. 5(a). Under rotation, however, the Coriolis acceleration will cause energy to be transferred from the x-axis (primary mode) to the y-axis (secondary mode) building up a vibration amplitude along the y-axis. The ratio of the amplitude in the secondary mode of vibration to the amplitude of the primary mode of vibration can be shown to be proportional to the rotation rate and is given by

$$\frac{y}{x} = 2Q\frac{\Omega}{\omega_n} \quad (2)$$

Notice, that the gyroscope response is proportional to quality factor Q of the device. Thus, MEMS gyroscopes have to be vacuum packaged to achieve have amplitude of response in the sense direction.

In case of the x-axis gyroscope, the proof-mass is driven to a fixed amplitude along the y-axis by applying an electrostatic drive force to the proof-mass along the y-

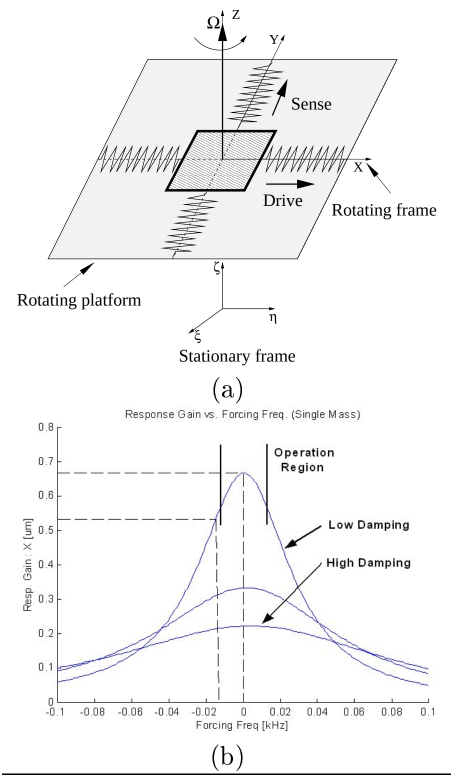

Figure 5. (a) Mass-spring model of the vibratory micromachined gyroscope; (b) The response of the vibratory gyroscope to the Coriolis force.

axis. Under rotation with respect to the x-axis, the Coriolis acceleration will cause energy to be transferred from the y-axis (primary mode) to the z-axis (secondary mode) building up a vibration amplitude perpendicular to the surface of the chip. The amplitude is measured and related to the input angular velocity along the x-axis. Similarly, the y-axis gyroscope measures rotation with respect to the y-axis by driving the device in resonance along the x-axis and measuring the response in out of plane direction relative to the chip. The described principle of operation is not the only option for implementation of MEMS gyroscopes. Multi-degree of freedom gyroscopes and vibratory gyroscopes with rate integrating properties are also under investigation [6].

Thus, the sensing unit of the proposed vestibular prosthesis should include three axes accelerometers and three axes gyroscopes implemented on the same silicon chip. This configuration of micro-devices allows to measure all six degrees of freedom motion of the human head. The output from the accelerometers and gyroscopes are voltages proportional to linear accelerations and angular velocities, respectively.

PULSE GENERATOR

The sensors sense the motion of the head, and send analog voltage signal to control electronics, i.e. pulse generator. The pulse generator includes a low-pass filter, an A/D converter, and the programmable microprocessor with externally adjustable gains. The analog-to-digital (A/D) converter converts the filtered analog signal to digital signal. The microcontroller reads in values from the A/D converter, and adjusts the parameters of the prosthesis.

The pulse generator has six input and six output channels. Each input channel provides information from the sensing unit in the form of voltages proportional to either angular velocity or angular acceleration. The output of the pulse generator is a pulse train with specific amplitude, rate, and duration encoding the inertial information, which, ideally, should be identical to those produced by the natural vestibular system.

The transfer function relating input angular velocity to the stimulating pulses is different from one subject to another. Fernandez and Goldberg [15] proposed a transfer function to model primary afferent responses of the vestibular system in a squirrel monkey model:

$$H(s) = \frac{\tau_A s}{1+\tau_A s} \cdot \frac{1+\tau_L s}{(1+\tau_1 s)(1+\tau_2 s)} \tag{3}$$

where τ_A is related to the level of adaptation to the constantly acting stimulus (i.e., time interval after which there will be no neural response to the constantly acting stimulus) and τ_L is related to hair cell response to the velocity of cupula (a motion transducer from endolymph to vestibular nerve, providing sensation of movement to other parts of the body). Two other parameters, τ_1 and τ_2 define dynamic parameters of the natural vestibular organ. τ_1 is the ratio of damping (combined, endolymph and cupula) over the spring constant of cupula, and τ_2 is the ratio of the effective momentum of inertia of cupula to damping. A portion of the transfer function

$$\frac{1}{(1+\tau_1 s)(1+\tau_2 s)} \tag{4}$$

approximates the system endolymph-cupula as a simple damped oscillator. While the component

$$\frac{\tau_A s}{1+\tau_A s} \tag{5}$$

is added to account for the system's adaptation in response to constant stimuli. In subjects appearing to have little adaptation, this pole of the transfer function (5) disappears. Zero of the transfer function (3), $1+\tau_L s$, indicates that the pulse rate is a function of cupula velocity as well as displacement.

Testing conducted in squirrel monkeys [12] led to the time constants resulting in the following average transfer function:

$$H(s) = \frac{80s}{1+80s} \cdot \frac{1+0.049s}{(1+5.7s)(1+0.003s)} \tag{6}$$

For example, the effects captured by this transfer function were observed in all mammals to some degree (see, for example, [11]).

Our pulse generating block will produce an impulse train at the frequency determined by the transfer function, similar to the one described by (6).

Each prosthesis has to be programmed and individually adjusted after implantation. This requires reprogrammable features integrated right on the chip, such as setting and adjusting time constants of the transfer function, adjusting the time constant of the digital high-pass filter and the input channel A/D converter. First prototype of the pulse generator will be using a programmable micro-controller, e.g. an 8-bit CMOS micro-controller PIC16F84-04/SO with flash memory and serial in-system programming capabilities, or a general purpose DSP for the proof of principle purpose.

STIMULATOR

It has been shown that charge transfer, or current, is a greater factor than voltage in electrical stimulation of the nerves [16]. Hence, a current output, as opposed to a voltage output, more desirable for the electrical stimulator of the vestibular implant.

The stimulation will be in the form of an impulse train, the frequency of which will be determined by the direction and magnitude of angular motion.

The output of this stimulator will be a biphasic current pulse. A biphasic pulse is used to maintain an overall charge output of zero. This extends electrode life and reduces tissue damage due to electrical stimulation [17]. In our initial prototypes, the inputs to the device can be from a microcontroller or a DSP that determines the pulse frequency, pulse amplitude, pulse rate, and also the lag between positive and negative phases of the biphasic pulse. In our final all-on-one-chip prosthesis, all signal processing functions will be implemented using on-chip mixed digital/analog integrated circuit, mechanical frequency references, and mechanical filters. The stimulator delivers current signals to the vestibular nerve through a flexible electrode array, which would be similar in many respects to a cochlear implant electrode array.

EXPERIMENT

To verify feasibility of using MEMS sensors as a candidate for the vestibular prosthesis, we tested a prototype of MEMS Z-axes gyroscope from Analog Devices Inc. This prototype is developed using integrated polysilicon surface micromachining technology, with electronics integrated on the same chip with mechanical part of the sensor. The device is vacuum packaged. The sensor was placed on an inertial grade rate table. Leads from the package were connected to the off-table terminals via slip rings. The device was rotated both

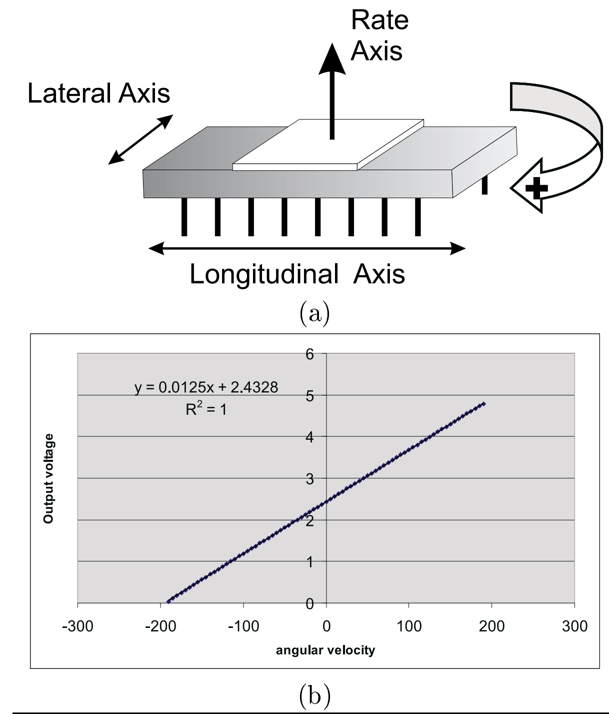

Figure 6. (a) A z-axis MEMS gyroscope was tested on an inertial grade rate table; (b) Experimental results of the rate table testing (input angular velocity vs. output voltage) of experimental ADI's z-axis MEMS gyroscope

in clock-wise and contro-clockwise direction, Fig. 6(a). The z-axis MEMS sensor demonstrated better than 1 deg/sec sensitivity and good linearity over the range of $\pm 150^0/sec$, Fig. 6(b). Voltage sensitivity of the device was $0.6^0/sec/Volt$ and rate noise density $0.05^0/s/\sqrt{Hz}$. This example of the sensor prototype demonstrated feasibility of using MEMS technology for implementation of the prosthesis. An orthogonal triad of MEMS single axis gyroscopes was also assembled and tested in a three-axes sensing unit.

We are currently characterizing our first prototype of multi-sensor unit, Figure 3, and also implementing the pulse generator, stimulator, and micro-electrodes for the stimulator. All these components will be designed with the same goal that the complete system integration on the same chip. During initial experiments with discrete sensors we will attempt to build the most flexible test-bed system. External controller and externally programmable pulse generator will be used during the first set of experiments.

Animal experiments will be conducted to first adjust parameters of the transfer function and verify functionality of the prosthesis. The vestibulat function will be assessed by measuring the vestibular-ocular reflex (VOR) [8], which is a reflexive movement of the eyes induced by measurements of motion made by the vestibular system. The VOR is generally accepted as one of the best objective measures of vestibular function.

CONCLUSIONS

The goal of this paper was to demonstrate that MEMS technology is a viable candidate for implementing a completely implantable vestibular prosthesis. Based on the available physiological data, we have identified performance requirements for the vestibular prosthesis and concluded that MEMS sensors can provide better performance than the human's sensation thresholds. Experimental data for a prototype of ADI's surface micromachined rate gyroscope supports the claim. We also presented an architecture for the vestibular prosthesis and fabricated the first prototype of the sensing unit of the prosthesis. Development of the completely implantable vestibular prosthesis is in its initial phase of exploration. Many technical issues need to be addressed, including the design of a robust low-drift sensing unit on the same chip, design of control architecture suppressing the drift over time in the unit, integration of pulse generator and stimulator on the same chip, capability providing wireless programming and power beaming to the chip, design of bio-compatible package for the prosthesis, and interface of the prosthesis with neurons.

Benefiting from the unique capabilities of the MEMS technology, this vestibular implant will be small and consume little power, and can be potentially manufactured in large quantities at low cost.

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