A Device for Diagnosing Vestibular Disorders Using Stochastic System Identification

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Abstract—This describes the development and testing of a first-line diagnostic tool that determines the transfer function of a patient's vestibulo-ocular reflex. The functional prototype consists of two components. Diagnostic glasses measure head acceleration and eye position, while a portable actuator noninvasively perturbs the vestibular system. Stochastic system identification is used to calculate the transfer function based on head acceleration and eye movement. We will demonstrate the viability of this device by using it to distinguish between functional and dysfunctional vestibular systems.

1 Introduction

Vestibular system disorders are some of the most prevalent and difficult to diagnose medical conditions in the United States. Although there are many distinct causes of vestibular dysfunction, most manifest as vertigo or dizziness, making diagnosis challenging. A full battery of tests performed at a specialized diagnostic facility is currently required to accurately diagnose some vestibular disorders. The first-line diagnostic tool detailed in this paper has the potential to drastically decrease the cost and increase the quality of vestibular diagnostics.

1.1 Problem

Vestibular disorders often go undiagnosed or unreported, and the following estimations demonstrate the prevalence of such conditions and the need for more effective diagnostics. As many as 35 percent of Americans aged 40 or older have experienced some form of vestibular dysfunction during their lives [1], and 3 million Americans consult a physician regarding disequilibrium each year. Furthermore, only a third of these patients have vestibular disorders [2]. Distinguishing vestibular disorders from each other and from other causes of disequilibrium represent significant medical challenges.

1.2 Current Diagnostics

There are a number of problems with the current state of vestibular diagnostics. The primary issue is economic. The cost of vestibular diagnostic equipment, such as a rotating chair and eye tracking system, is $100,000 [3]. This makes it economically infeasible for many medical centers to have the diagnostic equipment.

The testing is also inefficient; there is no single test that can be performed to investigate all possible causes of vestibular disorders. Furthermore, the results of this battery of tests are taken as a whole rather than as a sequence.

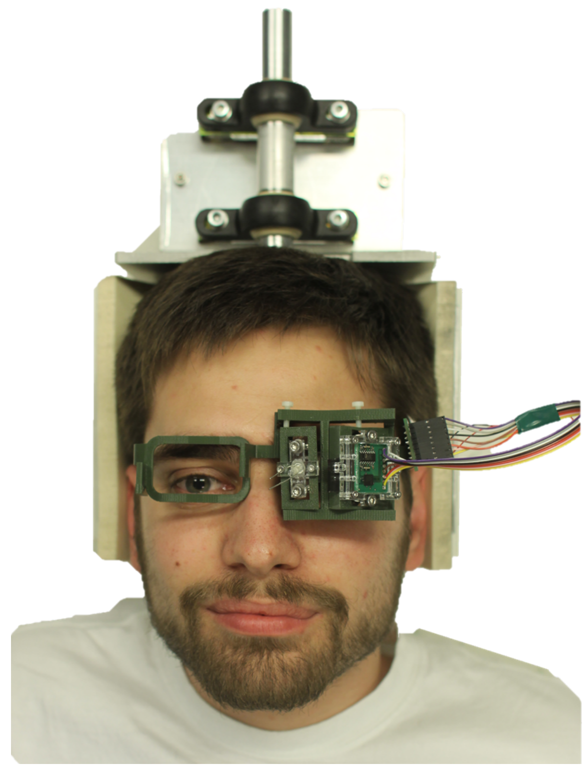

Fig. 1. Vestibular Diagnostic System. The completed system consists of a set of diagnostic glasses, here depicted on the face of a male subject, and a head actuator. The head actuator is a ”pillow” design that the patient can place their head in while lying down, as shown above.

1.2.1 Techniques

There are a number of tests currently used to assess vestibular function. The three most common techniques are the rotary chair test, the caloric test, and Electronystagmography.

For the rotary chair, the patient is placed on a chair that is rotated from side to side sinusoidally and the position of the eyes is recorded. For the caloric test, hot or cold water is injected into the patients inner ear thereby stimulating the vestibular system. By observing the vestibular system's response in the motion of the patient's eyes, the clinician can make a rough determination of the level of function of the vestibulo-ocular system and begin to identify what part of the vestibular system is affected. If there is an inner ear disorder, this test has the benefit of determining which side it is on. The procedure is, however, very uncomfortable for the patient. Electronystagmography relies on placing electrodes near the eye to pick up its electric dipole moment in order to measure small, involuntary eye motions.

1.2.2 Shortcomings

Although the individual tests provide valuable information, that information is often extracted in an expensive, inefficient, inaccurate, or invasive way. The rotary chair is the largest contributor to the cost of vestibular diagnostics. It costs over $100,000 and requires a full room [3]. The caloric test is inexpensive, but it is invasive and causes significant patient discomfort [4]. Nystagmography can also be expensive (over $20,000) and has a tradeoff between bandwidth and accuracy. Electronystagmography has high bandwidth, but is affected by electrode placement and interactions by nearby muscles. Video nystagmography is accurate, but usually slow, less than 100 frames per second.

2 System Overview

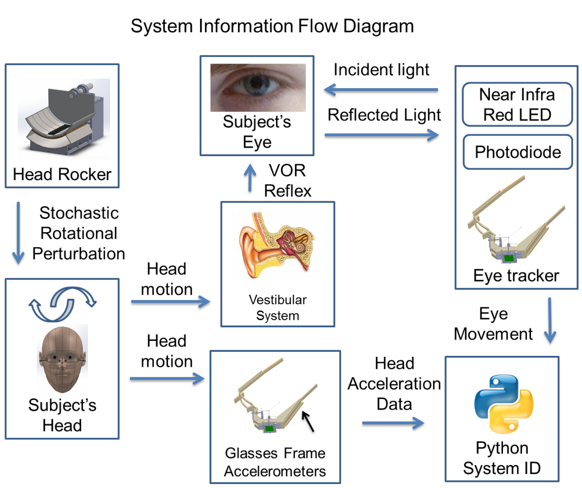

Fig. 2. System Information Flow Diagram

2.1 Design Concept

The goal of this project was to produce a first-line diagnostic tool for vestibular disorders. To that end, the primary objectives were to keep the device compact and inexpensive without sacrificing accuracy or bandwidth.

We thus decided to use a head-mounted glasses tool as the primary diagnostic device. It is light enough to be comfortably worn by the patient. The glasses contain accelerometers to measure head acceleration, an optical system for eye tracking, and all the associated electronics.

In order to measure the vestibulo-ocular reflex, the vestibular system must be perturbed. Our method of delivering this perturbation is to rotate the head. The actuator designed for this project utilizes a pillow design where the patient places their head in a cradle that rotates about the axis at the center of the head.

2.2 Testing Procedures

The goal of this project was to create a device simple enough to allow non-specialists to diagnose vestibular disorders. The final product consists of three parts: a set of diagnostic glasses, a portable head actuator, and the software suite that can take in the information from these devices and characterize vestibular system function.

For any test, the physician will put the diagnostic glasses on the patient. A calibration step will be necessary for each patient. During this step, the patient will focus on points to the left, right, top, bottom, and center of their field of vision. This will allow for the most accurate eye tracking and diagnostics. The software will guide the physician through the calibration, and no physician expertise in operating the device will be necessary or expected.

The patient will then place their head in the cradle component of the head actuator. The physician will select the desired program of head motion corresponding to the test they wish to perform. The test itself should take under a minute and the physician should be able to view the results of the test immediately.

2.3 Additional Applications

Although the primary goal of this project was to create a first-line diagnostic tool, there are several other intriguing applications of the device as well as other potential applications of the technology.

The most promising alternate use of the device is as a long-term diagnostic aid. Further development could lead to a version that patients could wear home during day-to-day activities. This could allow physicians to obtain information from subjects over a much longer time period. In particular, the diganostician could examine vestibular data from any point during the day when a patient experienced an episode of dizziness.

Several aspects of the technology utilized in this device are novel. The technique used for eye tracking is novel and substantially more accurate than existing technology. Thus, it could have applications in any industry where measuring eye motion is of interest. Such areas could include computer interfaces, video gaming, and consumer studies.

TABLE 1
Simulation of Lorentz Forces on head actuator

Model: Head Actuation Lorentz Force Simulation 3.mph Version: COMSOL 4.3.1.115		
Torque 1, z component	15.04313	Nm
Torque 2, z component	-15.2486	Nm
Electromagnetic Force 1, y component	38.299	N
Electromagnetic Force 1, y component	35.2307	N

3 Hardware Design

3.1 Actuator

In order to perform system identification on the vestibulo-ocular system, a binary stochastic torque input to the head was desired. To this end, a head actuator was designed and built, utilizing a Lorentz Force actuator.

3.1.1 Model

The actuator was designed to generate torques capable of rotating a human head. The CAD model of the actuator can be seen in Figure 3. The moving component of the actuator (the cradle) is a steel sheet with a line of magnets embedded at the middle (shown in black). Just beneath the moving cradle, there is a coil of wire (orange) attached to the base of the actuator. Extending below the coil, an additional sheet of steel completes the magnetic circuit. Current is passed through the coil to generate a Lorentz force along the z-axis in Figure 3, and reversing the current simply reverses the direction of the force. Any desired program of head rotation can be achieved by passing the appropriate currents through the coil.

3.1.2 Lorentz Force Simulation

To optimize the Lorentz force generated by the pillow actuator, we performed several simulations based on the CAD model. To optimize the dimensions and materials used for the actuator, we have decided to use the COMSOL Finite Element Analysis (FEA) software.

The magnetic field in the device and electrical current through the coil were simulated and used to calculate the resulting Lorentz Force, as shown in Figure 4. Varying parameters and re-simulating allowed optimization of some parameters and assisted in the design process. The final simulation predicted a Lorentz Force of 30N, which translates to a torque of 15Nm on the cradle. An analytic model confirms that 15Nm is enough torque to rotate the head through pi/16 radians at 5 Hz.

3.2 Glasses

The final prototype of the diagnostic glasses can be seen in Figure 5. The frame for the glasses was 3D printed. Grooves in the sides of the glasses house the printed circuit board. The optics responsible (green and blue components) are only shown for the left eye in this prototype.

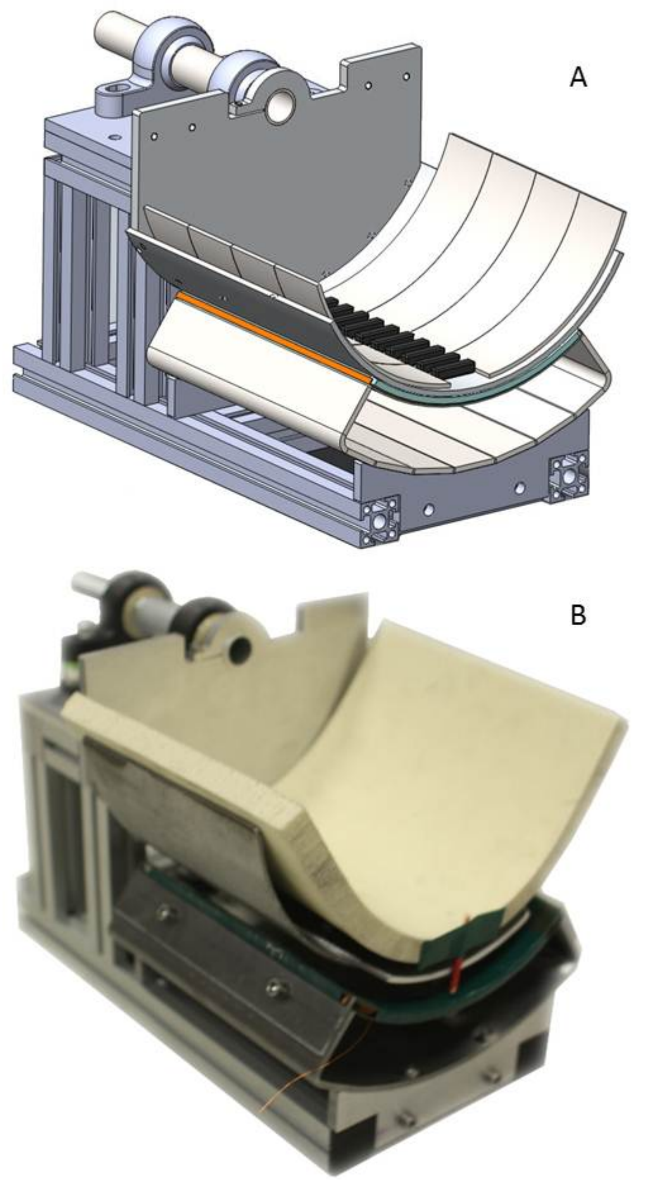

Fig. 3. Head Actuator. (A) Final CAD model and (B) constructed prototype of Lorentz force actuator.

3.2.1 Purkinje Measurement

The most novel technological component of this device is the method of tracking eye movement. Purkinje images, or reflections from structures in the eye, were utilized to obtain an exceptionally accurate measurement of eye position. There are four different Purkinje images, the first of which (P1) is the reflection from the outer surface of the cornea.

This novel method of eye tracking makes use of the non-spherical nature of the eye and the first Purkinje image. Our optical device focuses collimated light on the cornea, which is reflected and focused again on a lateral effect photodiode. Slight changes in the angle of the P1 reflection caused by eye movement can be measured as

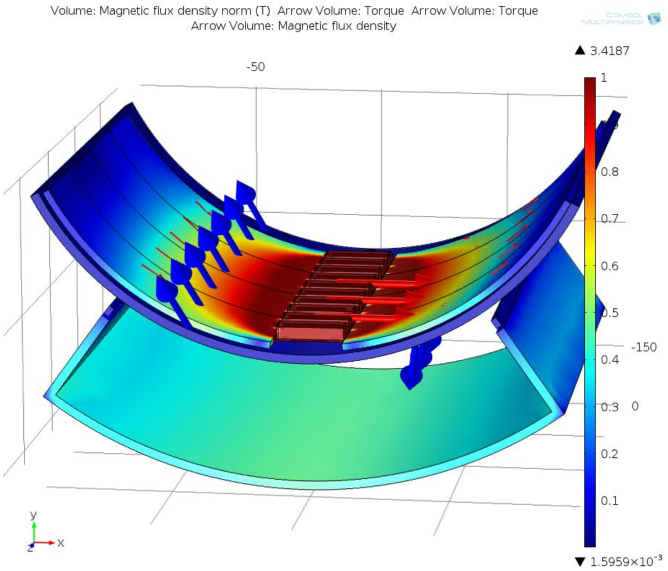

Fig. 4. COMSOL FEA model of Lorentz force head actuator. This model demonstrates that the actuator is capable of delivering 15Nm of torque to the head, which is sufficient for the desired angular acceleration.

changes in the position of the focused light on the lateral effect photodiode.

3.2.2 Optics

Once the head is perturbed, the vestibular-ocular system is activated and the stability of the eye is measured by utilizing the optical system (Figure 5a). This portable system is retrofitted into the standard design for eyeglasses. Several key aspects considered when designing this device include: (1) Mounting optical components to the custom glasses structure, (2) optical alignment from the LED to the sensor, and (3) direct integration of the electronics.

The optical components are directly mounted to the glasses, such that the optical paths are theoretically aligned (Figure 5b). Set screws connecting the elongated arms on lens mounts allows for adjustment of the pitch, yaw, and in plane z-depth on the photodiode op0ical component. Additionally, slotted fixtures allow for up to 1mm of adjustment of the entire fixture (Figure 5c and d). This motion accounts for a total 4 degrees of freedom from the photodiode optical component. An additional 3 degrees of freedom are integrated into the LED optical component design. The custom PCB board is integrated into a slot on the side of glasses.

3.2.3 Electronics

Our electronics consist of a main board, an accelerometer module, and a lateral effect photodiode module (Figure 6). The main board itself has an accelerometer to complement the accelerometer module which is mounted to the glasses on the opposite side of the head.

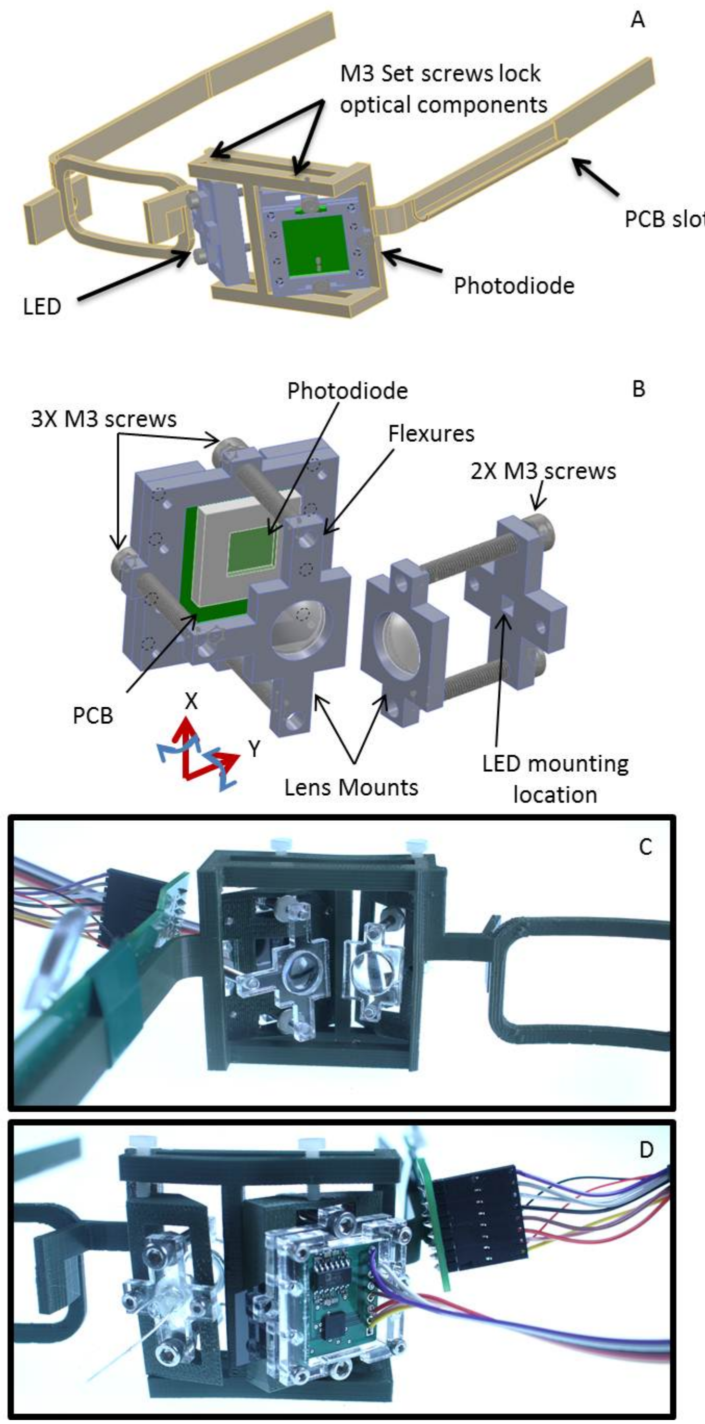

Fig. 5. Diagnostic Glasses. (A) Final CAD model showing the location of various optical components. The glasses base (tan components in A) were 3D printed and the optical and electrical components in B were attached. C and D show the final glasses prototype viewed from the back and front, respectively.

The main board charges a 45mAh lithium polymer battery via micro-usb. The battery powers three buck-boost converters at -5V, 3.3V, and 5V. The 3.3V line is used to power most of the digital circuitry. The 5V and -5V are used to bias the lateral effect photodiode, as well as to power the lateral effect photodiode modules op-amp and ADC. All sensors report to the Cortex M3 microcontroller on the main board. Each of the four currents from the lateral effect photodiode pass through a 4-channel op-amp in a current-to-voltage configuration. These currents are passed through an AD7195 ADC, which has a resolution of up to 24 bits and a bandwidth of 4.8kHz.

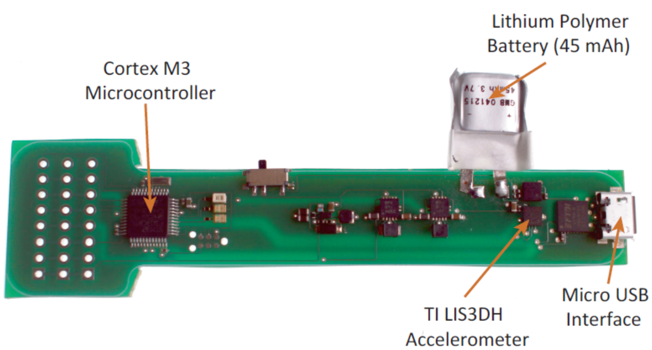

Fig. 6. Electronics for diagnostic glasses

4 System Identification

The input to our system was the head acceleration (rotation) and the output was the eye position. Based on these two parameters, we were able to characterize the vestibular system by calculating the transfer function.

4.1 Mathematical Framework

To calculate the impulse response, we have deconvolved the eye motion from the head acceleration. We have made use of the convolution theorem as follows:

For a transfer function h, we have the relation between input x and output y:

$$y = x * h$$

The convolution theorem states:

$$\mathcal{F}\{x * h\} = \mathcal{F}\{x\} \cdot \mathcal{F}\{h\}$$

where the dot represents pointwise multiplication. Thus

$$\mathcal{F}\{y\} = \mathcal{F}\{x\} \cdot \mathcal{F}\{h\}$$

or

$$\mathcal{F}\{h\} = \frac{\mathcal{F}\{y\}}{\mathcal{F}\{x\}}$$

yielding

$$h = \mathcal{F}^{-1}\left(\frac{\mathcal{F}\{y\}}{\mathcal{F}\{x\}}\right)$$

For this project, we used this simple form of system identification to correlate a single axis of acceleration with a single direction eye motion. However, fundamentally, our system takes six inputs (accelerations) and produces four outputs (eye positions). If we wished to perform system identification on this more complex setup, we could use a generalization of the above procedure. To describe an n-input, m-output system, the impulse response should be viewed as set of n-tensors, one for each of the m outputs. Then for each output, a generalized version of convolution theorem gives:

$$\mathcal{F}\{y_a\} = \mathcal{F}\{x_1\} \cdot ... \cdot \mathcal{F}\{x_n\} \cdot \mathcal{F}\{h_a\}$$

where each term is viewed as a n-tensor of the same shape as h_1 and is constant across all indices but its own. Then the impulse response can be extracted by a n-dimensional inverse fourier transform:

$$h_a = \mathcal{F}_n^{-1} \frac{\mathcal{F}\{y_a\}}{\mathcal{F}\{x_1\} \cdot ... \cdot \mathcal{F}\{x_n\}}$$

4.2 Advantages

This device utilizes binary stochastic system identification to calculate the vestibular systems transfer function, making it the most efficient possible method of data collection.

Existing diagnostics (most notably the rotary chair) utilize sinusoidal system identification. This requires a separate measurement to be made at each frequency. This antiquated mathematical technique is fundamentally less efficient than stochastic system identification. Our actuator moves the head back and forth randomly, allowing information to be collected about all frequencies simultaneously.

4.3 Results

We have analyzed 200 seconds of accelerometer and eye data taken at a rate of 10ms/sample. Figure 7 shows a typical 4 second slice of this data.

In order to reduce numerical stability issues often caused by using the method from 4.1, we have calculated the transfer function directly as

$$H = \frac{\mathcal{F}\{c_{xy}\}}{\mathcal{F}\{c_{xx}\}}$$

where cxy denotes the cross correlation of acceleration and eye position, and cxx denotes the autocorrelation of acceleration. Further, we have used Bartlett's method to reduce the variance of the spectogram at the cost of frequency resolution. In particular, we found the frequency response for 64 non-overlapping windows of our data, and averaged the resulting spectograms. Our resulting frequency response appears in Figure 8. We are still working to refine and interpret this.

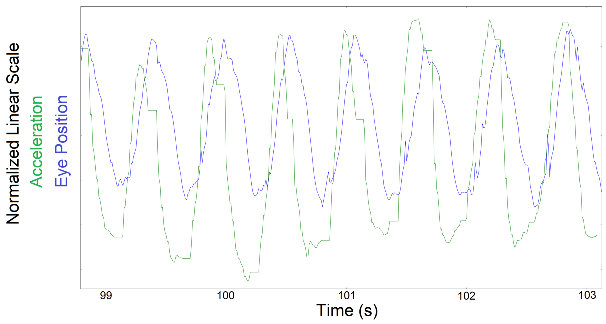

Fig. 7. Typical 4 seconds of accelerometer and eye data. The green line (head acceleration) typically lags behind the blue line (eye position) slightly.

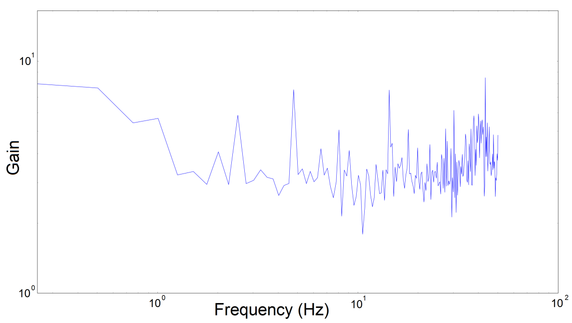

Fig. 8. Frequency response corresponding head acceleration to eye position.

4.4 Software

A software suite is in development. The goal is to make as simple a user interface as possible so that non-specialists can perform at least basic characterizations of vestibular system function.

5 Conclusion

5.1 System Verification

The purpose of this device is to characterize the transfer function of the vestibular-ocular reflex (VOR) of an individual to provide information useful in diagnosing vestibular system disorders. Unfortunately, no subjects with vestibular disorders were readily available to test with our device. However, it has been found that alcohol affects the function of the vestibular system, including the intensity of nystagmus [5]. Visual stabilization has been found to inadequately compensate for ethanol induced ataxia [6]. Gaze evoked nystagmus, and asymmetries of smooth pursuit can also be caused by intoxication [7]. For these reasons, we believe that using testing a subject before and after intoxication would be the best method for testing our system's ability to detect changes in the VOR transfer function.

A common clinical test for vestibular disorders is the Halmagy and Curthoys head-impulse test. This test determines dynamic visual acuity. The subject or clinician moves the subjects head at about 1 Hz and the eye movement is observed. As the head moves one direction, the eyes should track in the opposite direction smoothly. In a subject with bilateral VOR deficits, the subject's eyes will move too slowly to compensate for the head motion so their eyes will perform a corrective (catch up) saccade to regain fixation on the target. See figure INSERT below.

We chose to use a Halmagy and Curthoys head impulse test with a stochastic input with frequencies primarily around 1 Hz. Traditionally, the clinician is the both the actuator and the observer. This means that the data collected is qualitative and subjective. From our system we can quantitatively collect traditional data from the test as well as collect information about the transfer function of the VOR system. The team will measure the change in the VOR transfer function of one male subject, 23 years of age, before and after slight alcoholic intoxication. The stochastic perturbation lasts about 30 seconds, and the eye movement is tracked with our eye tracking system.

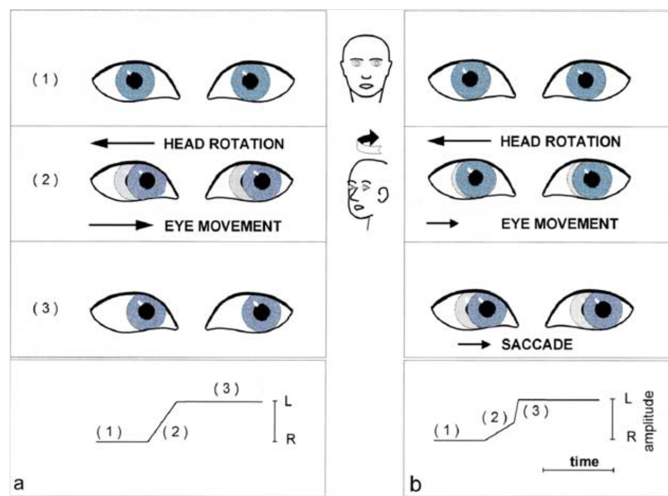

Fig. 9. Eye movements from a normal VOR (a) and a bilaterally deficient VOR (b)

Acknowledgments

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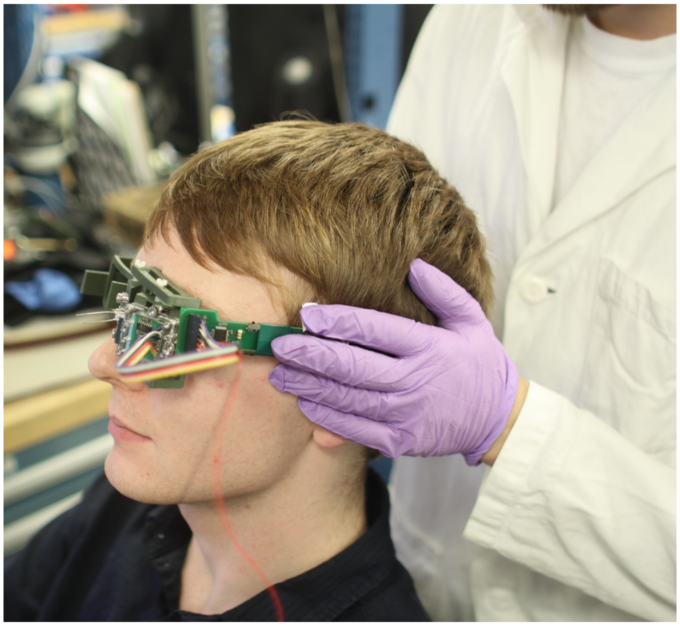


Fig. 10. "Physician" performing basic diagnostic measurement on male subject.

REFERENCES

[1] Agrawal Y, Carey JP, Della Santina CC, Schubert MC, Minor LB. Disorders of balance and vestibular function in US adults. Arch Intern Med. 2009;169(10): 938-944.

[2] Neuhauser HK, Radtke A, von Brevern M et al. Burden of dizziness and vertigo in the community. Arch Intern Med. 2008;168(19):21182124.

[3] Hain, Timothy, M.D. "Rotary Chair Testing." Dizziness-and-Balance. N.p., 18 2012. Web. 18 Dec 2012. ¡http://www.dizziness-and-balance.com/testing/ENG/rchair.html¿.

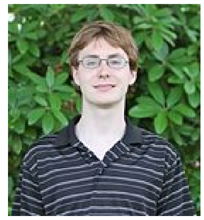
[4] Fife TD, Tusa RJ, Furman JM, Zee DS, Frohman E, Baloh RW, Hain T, Goebel J, Demer J, Eviatar L, Assessment. vestibular testing techniques in adults and children: report of the Therapeutics and Technology Assessment Subcommittee of the American Academy of Neurology. 2000; 55: 14311441.

[5] Kai Savolainen, Vesa Riihimki, Eero Vaheri and Markku Linnoila Scandinavian Journal of Work, Environment and Health Vol. 6, No. 2 (June 1980), pp. 94-103

[6] Nieschalk, M., Ortmann, C., West, a, Schml, F., Stoll, W., and Fechner, G. (1999). Effects of alcohol on body-sway patterns in human subjects. International journal of legal medicine, 112(4), 25360. Retrieved from http://www.ncbi.nlm.nih.gov/pubmed/10433036

[7] Brandt, T., and Strupp, M. (2005). General vestibular testing. Clinical neurophysiology?: official journal of the International Federation of Clinical Neurophysiology, 116(2), 40626. doi:10.1016/j.clinph.2004.08.009

[8] Cummings, Charles. Cum Product Details Cummings Otolaryngology: Head and Neck Surgerymings O. 2004. Ch. 144. Web. ¡http://famona.tripod.com/ent/cummings/¿.



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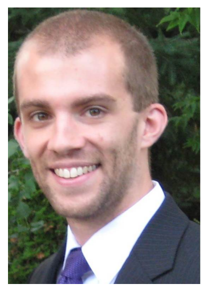

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