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Measurement of ocular counterrolling (OCR) by polarized light

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Abstract

Ocular counterrolling (OCR) and torsion refer to twisting eye movements that rotate the eyes about the line of sight. OCR is useful in assessing the activation of the gravito-inertial sensors (otoliths) of the inner ear (vestibular system). Information on otolith function is used in medical and experimental work on human orientation. As valuable as this information is, continuous measurement of OCR is still not readily available. This paper describes a method of continuously measuring OCR using polarized light, a system of polarizers and a contact lens. A polarized hard contact lens is placed between two soft lenses and caused to adhere to the eye by soaking in deionized water. The phase difference between the incident rotating polarized light and the reflected light from this lens gives a measure of eye torsion with little contamination from other eye movement modes. Application and performance of this instrument will be discussed.

Introduction

Eye movement analysis is often used as a noninvasive technique to study vestibular sensory and motor coordination in humans. The vestibular system consists of the semicircular canals, which act primarily as AC-coupled angular velocity transducers, and the otolith organs, which act as DC-coupled linear acceleration transducers.[1] Stimulation of the posterior semicircular canal and the otolith organs produces torsional eye movements, which are twisting movements that rotate the eyes about the line of sight. These torsional movements occur during head tilt and rotate the eyes opposite to the direction of tilt.[1]

Stimulation of the horizontal and vertical semicircular canals produce horizontal and vertical eye movements, respectively, that rotate the eyes opposite to the stimulus.[1] Our understanding of how the horizontal and vertical semicircular canals function has been aided by the development of non-invasive, real time, high bandwidth instruments that measure these rotations.[2]

Investigations of vestibular function relfected by torsional movements has been hampered by the lack of a convenient, high bandwidth, measurement instrument. Ocular torsion is difficult to measure because the features of the eye are roughly symmetric about the axis of rotation (line of sight). Compounding this difficulty is the relatively small amplitude of ocular torsion: the maximum torsion of about 6° is reached at approximately 60° of head tilt.[3,4] Previous methods to measure torsion have used photographs of the eye taken at rates up to 30 frames per second;[3,4] hard contact lens systems have been used with embedded magnetic pick-up coils[5] and mirrors.[6] However, the photographic systems are costly, time-consuming and low bandwidth, while the contact lens systems can damage the cornea and cause serious increases in intraocular pressure.[2]

To overcome the drawbacks of these systems, we have developed a system that incorporates features from the soft contact lenses system developed by Edelman[3] and polarizer system by Kamada et al.[6] By constructing a contact lens polarizer and placing it between two soft contact lenses, our experimental system has potential as an inexpensive, simple, safe, high bandwidth, torsion measurement system.

Methods

Figure 1 shows the basic design of our experimental instrument which is a modification of Kamada's.[6] Collimated visible light (S1) is passed through a polarizer (P1) rotating at 9,900 rpm to produce light with a rotating plane of polarization. A beam splitter (B.S.) separates the light beam into a references (REF) and a signal (SIG). The reference (REF) is then focussed through a second, stationary polarizer (P2) onto a photo detector (D1) that records the resulting sinusoidal change in light intensity (330 Hz). The signal is reflected off the aluminized rear surface of the polarizer contact lens (P3). The sinusoidal changes in light intensity produced by the reflected beam are recorded by a second photo detector (D2). The angle between the lens (P3) and reference (P2) polarizers is manifested as a phase difference between the two sinusoidal signals. A lock-in amplifier (PAR JB-4) with a low pass cutoff at 60 Hz converts this phase difference to a signal that varies with eye torsion angle.

Construction of the polarizing lens was accomplished using 0.010" thick sheet of HN 32 cellulose acetate butyrate. The material was heated and curved to match the cornea. The rear surface was then coated with aluminum using a vacuum-deposition technique. The polarizer contact lens was then fitted to an artificial eye with a potentiometer to measure rotation. Changes in the angle between the reference and lens polarizers were accomplished by a motor connected to the artificial eye; a range of ±15° was tested. These signals were sampled by a PDP-11/34 (250 Hz sample rate) for display and analysis.

For human experiments, the polarizer lens (smaller diameter than the contact lens) is sandwiched between 2 soft contact lenses (B&L softlens plano). The top or outer lens is soaked in deionized water to make it hypotonic with respect to the physiological (0.09% saline) tonicity of the inner lens. The polarizer is placed between these two lenses and allowed to sit for several minutes so that the outer lens will shrink and form a tight bond to the inner lens. This sandwich is then placed on the eye and irrigated with deionized water. A "tightening" sensation, not unpleasant, can be felt by the subject which signals adherence to the eye.

Results

The ocular torsion instrument performed well under our controlled experimental conditions. A ramp change in torsional eye position, θ_T, versus time, using the motor driven artificial eye produced a smooth ramp change in measured eye torsion, θ_E (Fig. 2). Over the ±15° range tested, the instrument displayed a noise level of 0.5° with a low pass filter set at 60 Hz. This filter setting represents an acceptable compromise between noise and bandwidth. The linearity of the system is shown in Figure 3 where measured eye position is plotted against actual eye position. The virtually straight line shown by this graph attests to the close linear relationship of the instrument response over the range tested.

The affects of physiological solutions, i.e. saline, deionized water, and boric acid on the polarized materials were tested briefly. The saline and deionized water solutions caused the lens to lose its polarizing property. This process occurred over a period of 3 weeks. However better results were obtained when the lens was stored in boric acid. After 3 months the lens has yet to show any signs of breaking or deterioration. The use of boric acid solution would serve as a substitute for storage without any physiological complications since many eye washes contain boric acid solution.

While human experiments on torsion using this system were not available at publication, a successful contact lens sandwich using the polarizer has been made and placed in a human eye. The results from these preliminary studies indicate no harmful effects in short term use of this system (2-3 hours). Examination of the eye after application showed no cornea edema and no post application effects were noted in our one volunteer.

Discussion

The present application of a contact lens polarizer appears to be a promising technique to measure ocular torsion. The instrument's performance under bench test conditions compare favorably with those of Kamada despite our use of a curved polarizer. While Kamada's system showed a linear range of ±17° and a noise level of about 1.0° (measured from their Figure 3), our system had at least a linear range of ±15° and a noise level of less than 0.5°. These results are comparable to performance of the more conventional horizontal and vertical eye movement monitors available.[2] However, the range and noise level of our instrument is not as good as the photographic or hard lens methods. These methods have a noise level of 10 min. arc and range of >±30°. Nevertheless the drawback of these systems make this less precise method very attractive. For example, the photographic method costs about 20¢ per frame, takes at least 1 minute per frame to analyze, and the data is usually not available until several days after the experiment. The hard lens systems have the problems of clinical acceptability and short duration of subject tolerance.

Before widespread clinical and experimental use of this instrument can begin several problem areas must be investigated. Among these are miniaturization of the mechanical apparatus, controlling the curvature of the polarized contact lens, and the effects of long term wear of the polarized lens. Finally, possible problems associated with horizontal and vertical eye movements on the linearity and artifacts using this system need further attention.

Acknowledgments

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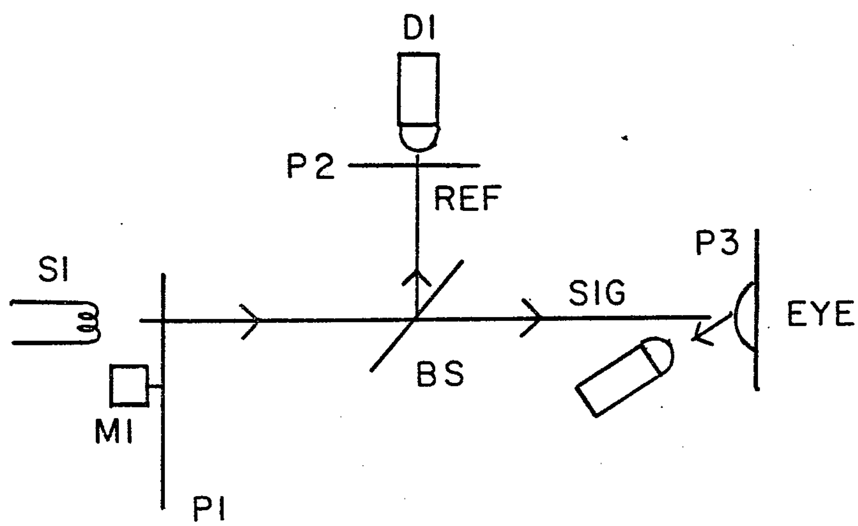

Figure 1. A schematic of the instrument adapted from Kamada et al (1976). The motor (M1) rotates the polarizer (P1) at 9,900 rpm to give a 330 Hz sinusoidal signal at the photo detectors (D1, D2). The phase differences between the reference (REF) and the signal (SIG) varies with the angle between the reference polarizer (P2) and the polarizer contact lens (P3).

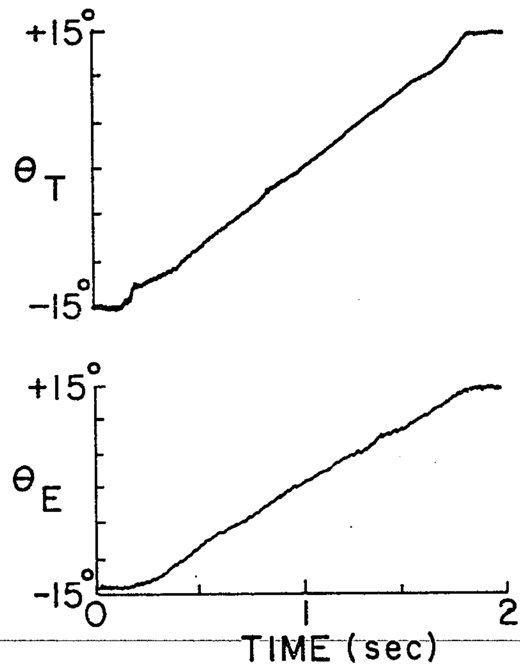

Figure 2. Shows the ramp position versus time of the motor driven artificial eye (θ_T) and the output of the instrument (θ_E) versus time showing the noise level of about 0.5°.

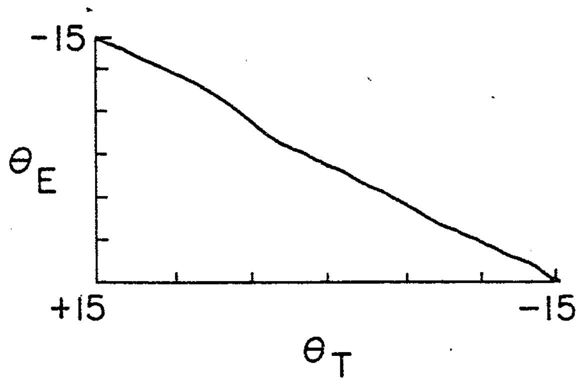

Figure 3. A graph of the output of the instrument (θ_E) versus the actual eye position (θ_T). The linearity is indicated by the straightness of the data.