INTRAOCULAR PRESSURE SENSING USING A MEMS PRESSURE SENSOR EMBEDDED IN A CONTACT LENS

Courtney Howell
courtney.howell@colorado.edu
Department of Electrical, Computer and Energy Engineering, University of Colorado Boulder, CO 80309

Abstract. This paper presents an improved, noninvasive method for 24-hour intraocular pressure measurement using a Platinum strain gage embedded in a soft, silicone contact lens. By using a circular strain gage that rests along the corneoscleral junction, changes in curvature of the eye are measured. Due to the relation between curvature and pressure, IOP is then calculated. The proposed design has a theoretical sensitivity of at least 0.5 mm Hg, outputting a voltage of 92 μV/mm Hg.

Background. Glaucoma is a widespread disease in which fluid builds up at the front of the eye, increasing intraocular pressure (IOP) to the point where it damages the optic nerve. It is estimated that Glaucoma is the second leading cause of blindness, and the leading cause of blindness for people over 60 [1]. While there is no cure for glaucoma, early diagnosis and treatment can significantly extend a patient's vision. Currently, optometrists most commonly use a tonometer to check intraocular pressure during yearly eye exams. Normal intraocular pressure is between 12-22 mm Hg; anything above this range indicates a possible glaucoma case and signifies a need for further investigation [2]. However, it is well documented that intraocular pressure fluctuates throughout the day, tending to be highest in the morning and lowest in the evening [3]. These fluctuations can range from 2 mm Hg up to 10 mm Hg. In order to avoid missing diagnosis of a possible glaucoma case due to appointment timing, a method of monitoring pressure over a 24-hour period is needed.

Many methods for 24-hour IOP monitoring have been proposed and researched, but many must be implanted in the eye, which makes using the device risky [6]. The current best solution to 24-hour monitoring while being minimally invasive is a contact lens with a pressure sensor in the lens. Two types of sensors have been studied: a capacitive pressure sensor with plates floating in the lens [4, 6], and a strain gage that measures changes in eye curvature and relates this to eye pressure [5, 6].

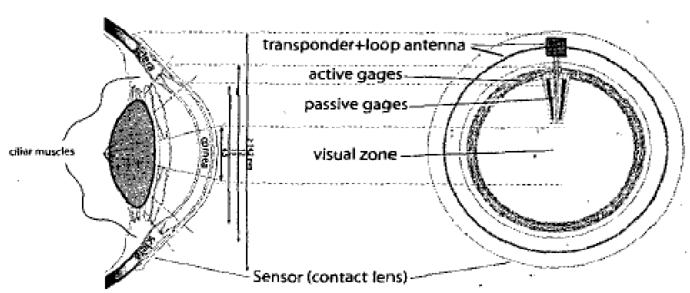

Figure 1: Overall design of the sensor. The device consists of two circular active gages placed along the corneoscleral junction to detect changes in curvature and two passive gages placed radially to compensate for temperature. The figure shows an area for the transponder for wireless communication with an external microprocessor, but no details of this circuitry are shown.

Of these two methods, the latter is more reliable and viable because it is less likely to obstruct vision or be affected by the material used for the contact lens itself. Previous research has produced a lens that is sensitive to within 2 mm Hg, but this sensitivity is still too low and could result in false or missed diagnoses of borderline cases. Our target sensitivity is about 0.5 mm Hg.

Design. My design is based on a previous design that uses circular strain gages inside a soft contact lens to measure the change in the radius of curvature of the eye [5]. Figure 1 shows the overall design of the device. I have modified the dimensions of the strain gages to increase sensitivity.

The device consists of two active strain gages and two passive strain gages in a Wheatstone bridge configuration as shown in Figure 2. The active gages are circular and should encompass the corneoscleral junction when worn, while the passive gages are straight and radially oriented so that they should not experience strain. Deformations in the curvature of the eye will change the resistance of the active gages, and the passive gages serve as temperature compensation. Thus, changes in curvature result in a change in output voltage. The device also includes a transponder and antenna to transmit the measured voltage wirelessly to an external microprocessor worn by the patient.

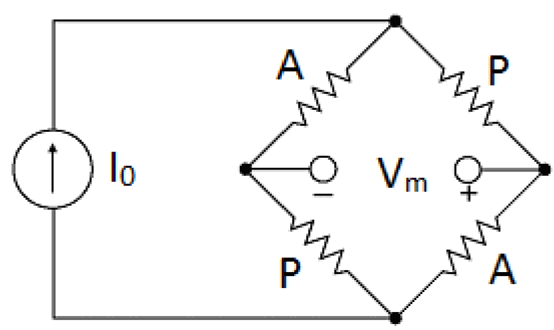

Figure 2: Wheatstone bridge configuration of the gages. A small DC current I0 of about 100 μA acts as a stimulus and the output voltage Vm is measured. The passive gages are marked with a P and active gages are marked with an A in the configuration.

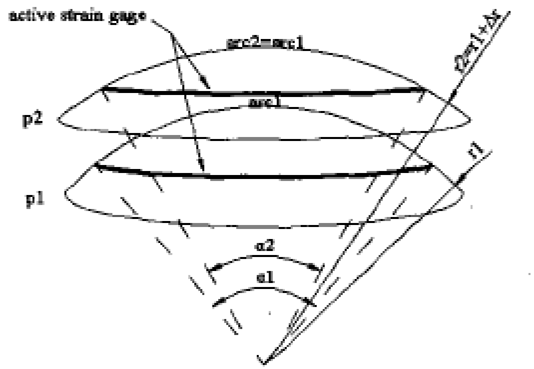

Figure 3: Contact lens deformation from an initial position p1 to p2. The lens follows the curvature of the cornea, and changes in intraocular pressure result in changes in corneal curvature. The curvature of the strain gage changes by Δr, but the arc length of the gage does not change. Strain on the gage is a nonlinear function of Δr, but can be estimated as linear because r >> Δr, so r1 ≈ r2.

The measured voltage is related to the input DC current I_0 by equation (1).

$$Vm = \frac{\varepsilon \cdot GF \cdot I_0 \cdot R_0}{2} \qquad (1)$$

where ε is the strain in the active gages, GF is the gage factor, and R_0 is the initial resistance of the gages.

The strain is a function of the change in curvature Δr by equation (2), which is also illustrated by Figure 3.

$$\varepsilon = \frac{(r + \Delta r) \cdot sin(\alpha \cdot \frac{r}{2(r + \Delta r)})}{r \cdot sin(\frac{\alpha}{2})} - 1 \qquad (2)$$

where r is the initial radius of curvature and α is the initial opening angle of the lens. Since we know that r >> Δr, we can simplify equation (2) further

$$\varepsilon' = \frac{1}{r} - \frac{\alpha}{2r} cot\left(\frac{\alpha}{2}\right) \qquad (3)$$

and substitute back into equation (1).

$$Vm = \frac{\varepsilon' \cdot \Delta r \cdot GF \cdot I_0 \cdot R_0}{2} \qquad (4)$$

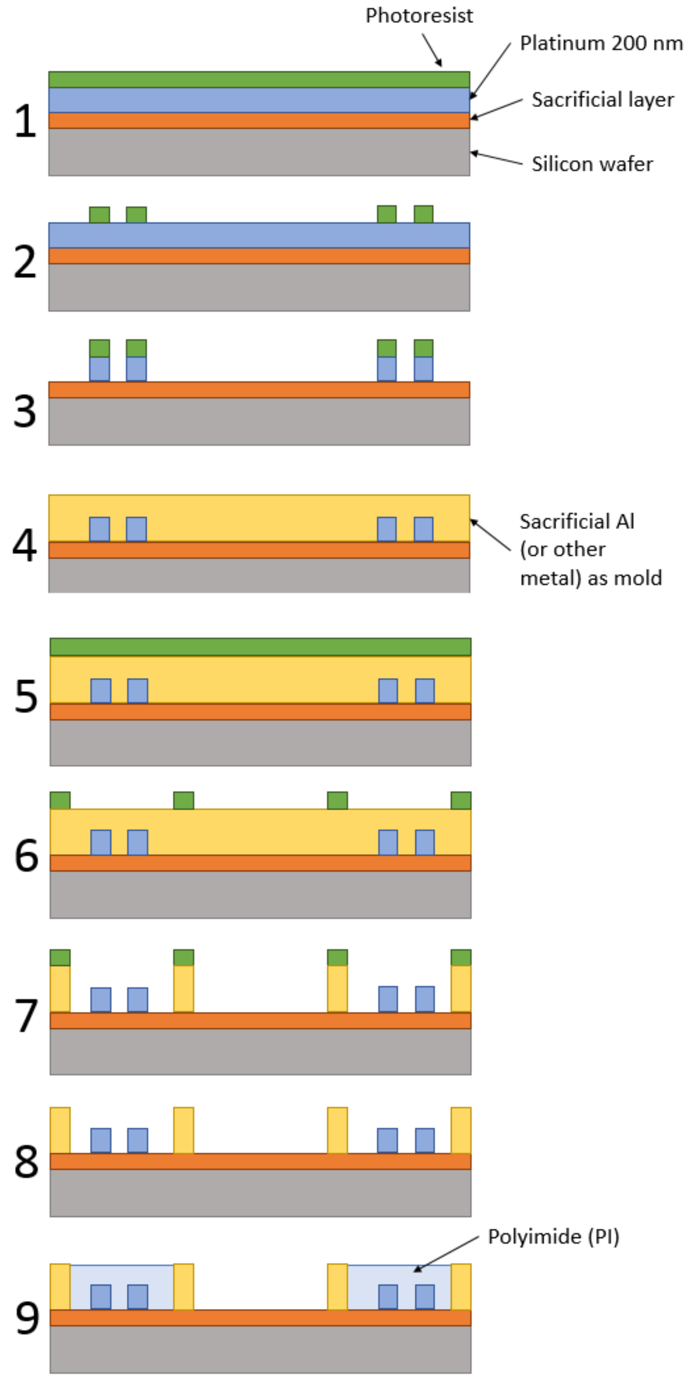

Figure 4.1: Fabrication process for the strain gage. Showing photolithography of platinum strain gages (steps 1-3), creation of aluminum mold through photolithography (steps 4-8), and pouring of polyimide into aluminum mold (step 9).

Studies have shown that radius of curvature is proportionally related to IOP such that a change of 1 mm Hg in pressure results in a 3 μm deformation in curvature [5], so we can relate measured voltage to change in IOP more directly.

$$\Delta p = \frac{6 \cdot Vm}{\varepsilon' \cdot GF \cdot I_0 \cdot R_0} \tag{5}$$

The final result is a proportional relationship between output voltage and change in pressure.

Fabrication. For the strain gages, 200 nm thick platinum is used. Platinum was chosen because it has a high resistivity and gage factor compared to other conductive metals, which means

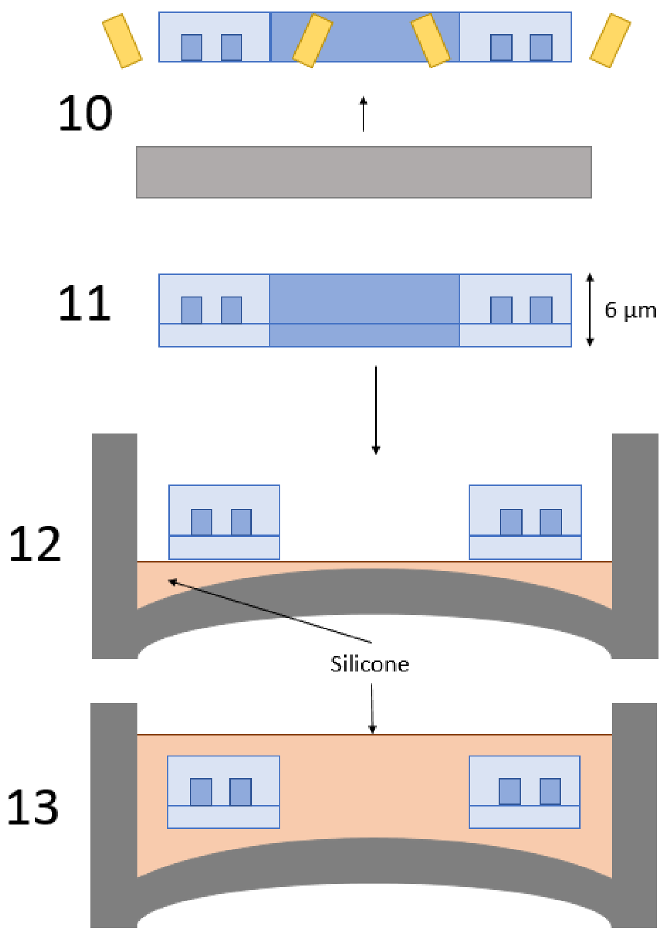

Figure 4.2: Fabrication process for the strain gage continued. Showing release of strain gage from aluminum sacrificial layer and mold (step 10), deposition of polyimide on uncovered side (step 11), and embedding of strain gage in silicone contact lens (steps 12-13). Note that the contact lens mold in steps 12 and 13 is highly simplified in the figure. Also note that in steps 10 and 11 the darker blue in the center is meant to portray depth of the polyimide ring due to only showing a side view of the device.

Table 1: Device dimensions.

Parameter		Value
Initial Radius of Curvature	r	8 mm
Initial Opening Angle	α	90 °
Platinum Thickness	t	200 nm
Platinum Width	w	5 µm

smaller changes in pressure will result in larger output voltages, effectively providing the best sensitivity for any conductive metal. The strain gages reside between polyimide layers. Polyimide serves as a flexible, yet strong, thermally stable, and hydrophobic protective layer. While polyimide is biocompatible, it is also yellowish in color, so to avoid discoloration of the wearer's vision, it only surrounds the gages and the bulk of the contact lens itself is made from silicone. Silicone is biocompatible as well and has the benefits of being more hydrophilic than polyimide (after chemical treatment) and clear in color, which makes the lens more comfortable and breathable to the eye [7].

Fabrication of the device is based on the process described in [5, 6] and begins with photolithography of the 200 nm thick p layer. Once the platinum is etched, aluminum is again used with repeated photolithography to make a channel for the polyimide. Polyimide is poured into this channel and allowed to set, then the sacrificial aluminum is removed. The other side of the device is covered in polyimide, producing a ring of polyimide approximately 6 µm thick with the strain gage embedded inside. A thin layer of silicone is poured into a contact lens mold and partially cured before the sensor is inserted into the mold. Once the sensor is in place, more silicone is poured to fill the mold. This fabrication process is summarized in Figure 4.1 and 4.2.

Analysis of Performance. The final dimensions chosen for the gages are summarized in Table 1. Dimensions were chosen with the following criteria in mind

- Maximum corneal deformation occurs at the corneoscleral junction position
- The average corneoscleral junction position is about 11.5 mm in diameter
- Due to the natural curvature of the eye, the initial opening angle cannot be less than about 70 degrees
- Initial radius of curvature should be in the range of 5-11 mm

From these dimensions, theoretical initial resistance was calculated using equation (6).

$$R_0 = \frac{\rho \cdot 2\pi \cdot r \cdot \sin(\frac{\alpha}{2})}{t \cdot w} \tag{6}$$

where ρ is the resistivity of Platinum, t is the thickness of the

Table 2: Constants and Calculated Variables

Constant		**Value**
Input current	I_0	100 μA
Gage factor of platinum	GF	6.1 [8]
Change in pressure	Δp	0.5 mm Hg
Resistivity of platinum	ρ	1.06*10^-7 [9]
Calculated Parameter		
Initial resistance	R_0	3767.57 Ohms
Strain	ε'	26.83
Strain gage arc length	l	35.54 mm
Strain gage diameter	d	11.31 mm
Output voltage	V_m	46.24 μV

gage, and w is the width of the gage.

An input current of 100 μA was chosen since current on this magnitude is still distinguishable to a sensitive microcontroller while keeping the power of the system low. Because the strain gage acts as a resistor, we don't want much heat being dissipated since the device rests on the eye.

The output voltage for 0.5 mm Hg was calculated using equation (5). Table 2 summarizes all constants and calculated values. The calculations showed that this design would theoretically output about 92 V/mm Hg, meaning 0.5 mm Hg would output a voltage of 46.24 μV. This voltage is distinguishable from 0V by the external microprocessor, meaning our target sensitivity was met.

Conclusions. The use of a disposable contact lens to measure intraocular pressure shows great promise for revolutionizing the way we diagnose glaucoma. The theoretical results obtained suggest that by changing the dimensions of the strain gages, we can significantly increase sensitivity. These results do not, however, account for any noise in the system, so simulation and actual fabrication would be highly beneficial to ensure that any noise in the system does not significantly impact results or decrease effective sensitivity.

My analysis also did not address any effects the wireless transmission component might have on the final output voltage. Further investigation into this part of the device is necessary to determine if wireless transmission is even viable for this application and what (if any) errors it might introduce into the system.

Future work should also focus on making sure the dimensions of the contact lens still allow it to fit comfortably on a person's eye, since the final purpose is to make a device that can be regularly and widely used instead of a tonometer. For example, prescription contact lenses have a range of curvatures so that they can fit comfortably on eyes with different curvatures, so to make this contact usable for any patient, a range curvatures will be needed. The width or radius of the strain gages will then need to vary to maintain the same sensitivity.

References.

[1] Boyd, Kierstan. "What Is Glaucoma?" American Academy of Ophthalmology, American Academy of Ophthalmology, 1 Mar. 2017, www.aao.org/eye-health/diseases/what-is-glaucoma.

[2] Tsai, James C. "High Eye Pressure and Glaucoma." Glaucoma Research Foundation, Glaucoma Research Foundation, 29 Oct. 2017, www.glaucoma.org/gleams/high-eye-pressure-and-glaucoma.php.

[3] Mosaed, Sameh. "Maximizing 24-Hour IOP Control." Review of Ophthalmology, Jobson Medical Information LLC, 16 Mar. 2006, www.reviewofophthalmology.com/article/maximizing-24-hour-iop-control.

[4] Juliet, A Vimala, et al. "Design and Analysis of Noninvasive Intraocular Pressure Sensor." ARPN Journal of Engineering and Applied Sciences, vol. 10, no. 9, May 2015, www.arpnjournals.com/jeas/research_papers/rp_2015/jeas_0515_2092.pdf.

[5] Leonardi, M, et al. "A Soft Contact Lens With a MEMS Strain Gage Embedded For Intraocular Pressure Monitoring." The 12h International Conference on Solid State Sensors, Actuators, and Microsystems, June 2003, pp. 1043–1046. IEEE Xplore, IEEE, ieeexplore.ieee.org/stamp/stamp.jsp?arnumber=1216947.

[6] Piso, Daniel, et al. "Modern Monitoring Intraocular Pressure Sensing Devices Based on Application Specific Integrated Circuits." Journal of Biomaterials and Nanobiotechnology, 20 Mar. 2012, pp. 301–309., doi:10.4236/jbnb.2012.322037. SciRes.

[7] Sensimed AG. SENSIMED Triggerfish White Paper. Sensimed AG, 2003.

[8] Craig, J. "Bridge Measure." Georgia Tech, 2000.

[9] Collaboration for NDT Education. "Conductivity and Resistivity Values for Misc. Materials." Collaboration for NDT Education, Mar. 2002.