A Technique to Measure Eyelid Pressure Using Piezoresistive Sensors

Alyra J. Shaw*, Brett A. Davis, Michael J. Collins, and Leo G. Carney

***Abstract*—In this paper, novel procedures were developed using a thin (0.17 mm) tactile piezoresistive pressure sensor mounted on a rigid contact lens to measure upper eyelid pressure. A hydrostatic calibration system was constructed, and the influence of conditioning (prestressing), drift (continued increasing response with a static load), and temperature variations on the response of the sensor were examined. To optimally position the sensor–contact lens combination under the upper eyelid margin, an *in vivo* measurement apparatus was constructed. Calibration gave a linear relationship between raw sensor output and actual pressure units for loads between 1 and 10 mmHg ($R^2 = 0.96$). Conditioning the sensor prior to use regulated the measurement response, and sensor output stabilized about 10 s after loading. While sensor output drifts slightly over several hours, it was not significant beyond the measurement time of 1 min used for eyelid pressure. The error associated with calibrating at room temperature but measuring at ocular surface temperature led to a very small overestimation of pressure. Eyelid pressure readings were observed when the upper eyelid was placed on the sensor, and removed during a recording. When the eyelid pressure was increased by pulling the lids tighter against the eye, the readings from the sensor significantly increased.**

***Index Terms*—Eyelid pressure, measurement, piezoresistive sensors.**

I. Introduction

THE EYELIDS act as an anterior physical barrier for the eye. Blinking of the eyelids is a protective mechanism that can occur in response to external stimuli, such as a sudden loud noise or a flash of light, with the closing mechanism taking less than 100 ms [1], [2]. The eyelids also maintain the health of the eye by replenishing the tear film over the cornea during normal involuntary blinking. In this role, the eyelids have been likened to windscreen wipers [3], with the inner edge of the eyelids serving to spread the tears with each blink.

Pressure from the eyelids can alter the corneal surface topography. The cornea is responsible for most of the focussing power of the eye; therefore, any changes to its surface shape can influence vision. Abnormal eyelids (due to disease) can increase or alter the pressure on the cornea [4]–[7]. However, pressure from healthy eyelids, when the upper and lower eyelids move closer to the center of the cornea during reading, can also cause temporary corneal distortion [8]–[13].

There have been attempts to measure eyelid pressure using modified contact lenses as manometers filled with either air or water. The apparatus designed by Miller [14] had a water-filled rubber balloon on the inner side of a hard contact lens and a catheter connected to a pressure transducer on the outer side. However, a second contact lens had to be worn to protect the cornea; therefore, the total thickness of the system was 2.5 mm. This thickness significantly alters the normal relationship between the eyelids and the eye. The use of water has also been criticized as there was no way to ensure that it was gas-free, since any air present would inflate the readings [15]. Lydon and Tait [15] used a hard contact lens with a silicone elastomer contact lens over the top to create a small chamber that was filled with gas. While the quantification of the eyelid pressure was not published, they concluded that these pressures were "small." An apparatus similar to Miller's system developed by Shikura *et al.* [16] recorded similar results for normal lid closure and tight eyelid squeezes. The measured eyelid pressure from these studies was between 1.7 and 51 mmHg [14]–[16]; however, the thickness and complexity of the systems makes the reliability of the results uncertain.

An effective system to measure eyelid pressure must be thin so that there is minimal alteration to the eyelid–cornea relationship, and should have high sensitivity to quantify small localized pressures. The system must be nontoxic and waterproof so that the tears do not influence measurements. It must also be able to conform to the surface of the eye and must not be affected by blinking or eye movement. To measure the pressure between the eyelids and the cornea, we used multiplexed array piezoresistive tactile pressure sensors (I-scan, Tekscan, Inc., Boston, MA). These sensors are relatively thin (170 μm), and are available in a suitable pressure range to measure eyelid pressure (rated at 5 lbf/in^2). Measurements can be taken up to 9.8 Hz and can be trimmed to suitable dimensions to be placed on the eye.

Various properties of Tekscan sensors used in biomedical applications have been previously considered [17]–[20]. For eyelid pressure measurement, the influence of drift and temperature particularly need to be considered. Drift or creep is the change in sensor output with a static load, and is thought to be due to the piezoresistive ink [17]. To improve the drift response of a sensor, it is recommended that it be conditioned or prestressed prior to use. The sensor manufacturer states that I-scan sensors are temperature-sensitive, and typically, measurements (the measurements of I-scan sensors or the measurements performed by the sensors) vary by up to 0.45%/°C. The best method for

Manuscript received January 20, 2009; revised April 7, 2009. First published May 19, 2009; current version published September 16, 2009. *Asterisk indicates corresponding author.*

*A. J. Shaw is with the Contact Lens and Visual Optics Laboratory, School of Optometry, Queensland University of Technology (QUT), Brisbane, Qld. 4001, Australia (e-mail: aj.shaw@qut.edu.au).

B. A. Davis, M. J. Collins, and L. G. Carney are with the Contact Lens and Visual Optics Laboratory, School of Optometry, Queensland University of Technology (QUT), Brisbane, Qld. 4001, Australia (e-mail: b.davis@qut.edu.au; m.collins@qut.edu.au; l.carney@qut.edu.au).

Color versions of one or more of the figures in this paper are available online at http://ieeexplore.ieee.org.

Digital Object Identifier 10.1109/TBME.2009.2022550

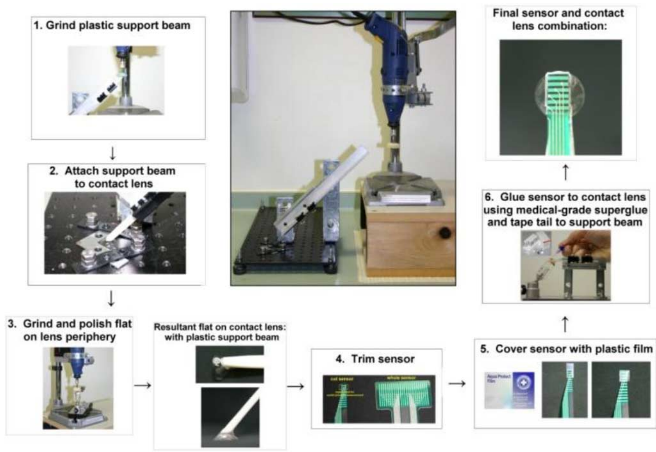

Fig. 1. Flowchart for constructing the sensor–contact lens combination. Process of: 1) grinding plastic support beam; 2) gluing support beam to contact lens using epoxy; 3) grinding and polishing a peripheral flat area on lens; 4) trimming the sensor; 5) covering the sensor with aquafilm tape; and 6) gluing the active part of the sensor to the peripheral flat lens area and taping the tail to the support beam.

temperature control is to calibrate the sensor at the same temperature as the measurement.

The quantification of eyelid pressure will provide a better understanding of its role in tear film spreading, along with corneal and contact lens biomechanics. A novel method to measure upper eyelid pressure using a thin (0.17 mm) tactile piezoresistive pressure sensor (I-scan, #4201, Tekscan Pty Ltd.) attached to a contact lens is described in this paper. To be able to use the sensor to measure eyelid pressure, it was necessary to design a contact lens to which the sensor could be attached. An apparatus was constructed to calibrate the sensor output. To understand the output response of the sensor, the drift and the influence of temperature were examined. A measurement apparatus was developed to accurately and safely place the sensor–contact lens combination in the eye to measure upper eyelid pressure.

II. Methods

A. Sensor–Contact Lens Combination

It has been reported that curving Tekscan sensors can cause an offset and decreased sensitivity of the sensor's output [19]. Therefore, for the output to result only from the applied pressure to the sensor and not curvature changes, the sensor must be fastened in a set shape on a nonflexible surface. It was, therefore, attached to a specially designed rigid contact lens (Capricornia Contact Lens Pty Ltd., Brisbane, Qld., Australia) (see Fig. 1). A generic back surface shape was used based on the average of 100 healthy young subjects with a radius of 7.8 mm (and prolate eccentricity $Q = -0.25$ [21]. A lens diameter of 15 mm was found to be large enough so that both eyelids maintained their position on the contact lens, thereby increasing the stability of the lens on the eye. As the contact lens diameter was larger than the average corneal size (11.7 mm [22]), peripheral curves were included in the contact lens design so that the lens back surface more closely matched the flatter sclera. The contact lens had a center thickness of 0.5 mm, which is well above the 0.13 mm critical minimum thickness to avoid flexure of a rigid (Perspex) contact lens [23]. The front surface of the contact lens was manufactured with a flat central area of 6 mm diameter, with a normal peripheral shape so that the eyelids could still easily slide over the contact lens surface.

To attach the sensor to the contact lens, it first needed to be trimmed to the appropriate size of nine cells (3×3). Trimming the sensor meant that it was no longer sealed from the tear film. So, a layer of very thin and flexible, "aquafilm" medical tape, which is both waterproof and bacteriaproof, was placed around the entire sensor. Additionally, this medical tape can be disinfected in the same way as other ocular instruments (using mediswabs, 70% isopropyl alcohol). Although applying the tape altered the sensitivity of the sensor, it was taken into account by calibrating with the tape in place.

So that the sensor and contact lens would remain stable on the eye (without rotating or flexing, which can cause false readings), a support beam was attached to the center of the contact lens using a nontoxic glue. A flat area was filed on the contact lens periphery so that the sensor could be mounted flat from the support beam onto the contact lens (see Fig. 1). The tail of the sensor was attached to the support beam with double-sided tape, while the active part of the sensor was adhered to the contact lens surface using Histoacryl, medical cyanoacrylate glue, which was approved by the Food and Drug Administration (FDA), and is commonly used to seal skin and corneal lacerations [24]. The active part of the sensor was positioned 3–6 mm from contact lens center, which is the approximate position of the upper eyelid and relative to the corneal center for primary horizontal gaze through to 40° downward eye gaze [25].

B. Calibration Apparatus

A novel calibration apparatus was designed since commercially available systems could not apply sufficiently low pressures, and did not allow the thicker sensor–contact lens combination to be inserted under the pressure applying plate. A hydrostatic pressure calibration system was designed using a column of water placed on the sensor (see Fig. 2). The sensor–contact lens combination was held with the sensor perpendicular to the water column and at the same height as the base tube. By lowering the water column onto the base tube, the water column acted on the sensor via the plastic membrane at the bottom end of the water column tube. Setting the density of water as 997.296 $\mathrm{kg/m^3}$ at 24 °C and acceleration to be 9.79 $\mathrm{m/s^2}$ at 27° latitude, a 7.04 cm column of water applied 5.17 mmHg to the underlying surface. This was confirmed by placing the water column tube on a balance and using the increase in mass from the water tube and the area loaded from the sensor software to independently determine the pressure applied. An increase in mass on the balance of 14 g loaded over 2.0 $\mathrm{cm^2}$ of the sensor is comparable 5.15 mmHg.

The same hydrostatic pressure calibration system was used for conditioning or prestressing the sensor. The influence of conditioning was examined by applying a measurement load of 7.8 mmHg on three occasions. The sensor was then conditioned

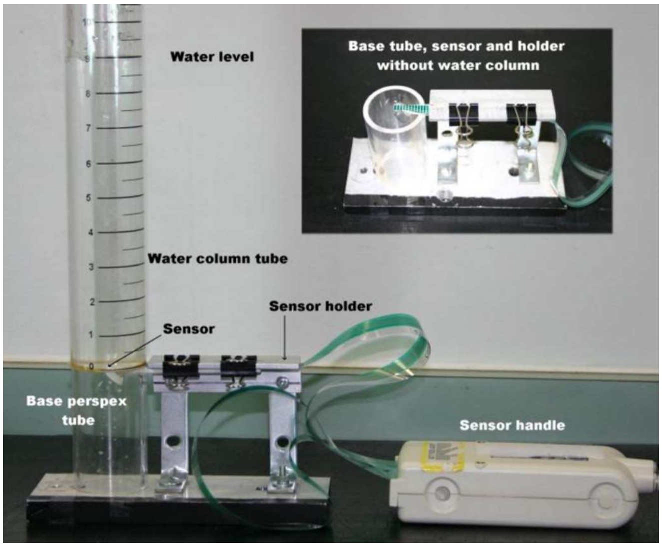

Fig. 2. Water column height calibration system. Inset shows system without water column in place.

with loads of 26 mmHg applied four times, each for 1 min with 30 s break between loads. The measurement load of 7.8 mmHg was then once again applied three times. The importance of the magnitude of the conditioning load and the length of time between conditioning and measurement were also examined with conditioning loads of 10.3, 25.9, and 51.7 mmHg and breaks of 10, 30, and 60 min between conditioning and measurement.

Each cell of the pressure sensor was individually calibrated as each varies significantly in both offset and sensitivity. This means that the raw score equivalent for a certain pressure can vary significantly from cell to cell. Raw scores can also vary from day to day, so calibration is required prior to every measurement. Custom calibrations for other Tekscan sensors typically have used three- or ten-point polynomial fits, which were shown to be more accurate than the linear or the power law options used in the Tekscan software [26]. The calibration process involved randomly applying loads of 1, 2, 2.5, 3, 3.5, 4, 5, 6, 8, and 10 mmHg, respectively, on two occasions. The raw score data were averaged for each load between 10 and 30 s after loading, and the best-fit polynomial calibration was calculated.

C. Likely Influence of Drift and Temperature on In Vivo Eyelid Pressure Measurements

To examine sensor drift, ten loads (1, 2, 2.5, 3, 3.5, 4, 5, 6, 8, and 10 mmHg) were each randomly loaded two times and the pressure recorded for at least 20 s. The mean raw score between 15 and 20 s after loading was calculated for each load, and the time taken for the output to remain within 10% of this average was determined.

The influence of temperature has been previously managed by keeping the room temperature close to the measurement temperature (for example, at the skin surface) [27]. By placing temperature and pressure sensors inside an incubator, the effect of changing temperature on the pressure output was investigated. The output was monitored as the temperature was increased from 22 °C to 39 °C, and then decreased back to 22 °C. Also, 7.8, 10.3, 12.9, and 15.5 mmHg loads were measured in the incubator for room temperature (23 °C) and at an average ocular surface temperature of 36 °C [28]–[31], so that the error associated with calibrating at 23 °C, but measuring at 36 °C, could be estimated.

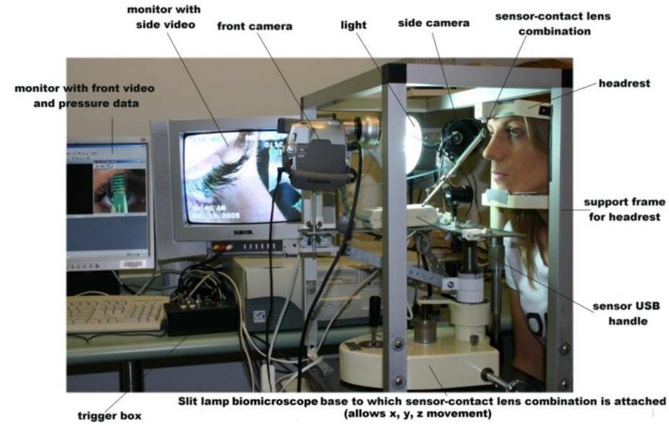

Fig. 3. Apparatus for upper eyelid pressure measurements.

D. Eyelid Pressure In Vivo Measurement Apparatus

For measurement on the eye, the sensor–contact lens combination needed to be stabilized so that it could not translate or rotate with respect to the eyelid position. The plastic support beam (attached to the sensor–contact lens combination) was fastened to a ball joint so that its orientation could be altered easily (see Fig. 3). Two video cameras provided front and side recording of the sensor–contact lens being placed onto the eye (see Fig. 3). The front (*en face*) recording was saved directly to the computer that controlled the pressure measurement system, so that it was synchronized with the pressure data. The side camera was connected to a monitor so that the image could be viewed as the contact lens was placed onto the eye. The sensor–contact lens and video cameras were mounted on a platform attached to a slit-lamp biomicroscope base, which allowed their position and height (x, y, and z planes) to be adjusted simultaneously, and the sensor to be accurately positioned under the upper eyelid margin (see Fig. 3). This research was approved by the University Human Research Ethics Committee and all subjects gave informed consent before participation.

III. Results

A. Calibration

The average raw score output for a pressure cell for loads of 1, 2, 2.5, 3, 3.5, 4, 5, 6, 8, and 10 mmHg (each applied twice, average between 10 and 30 s after the load is applied) is shown in Fig. 4. For this pressure range of the sensor, linear regression provided a good fit to the data (for this example, the coefficient of determination $R^2 = 0.96$).

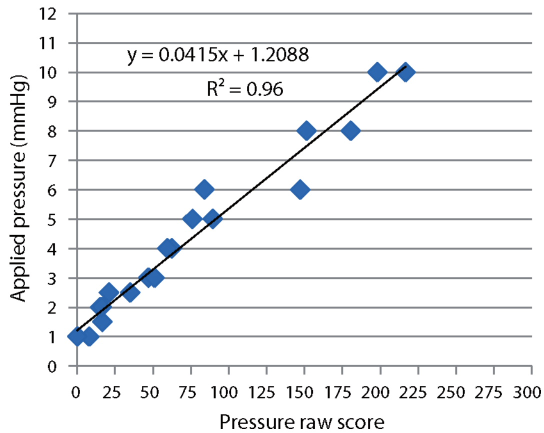

Fig. 4. Calibration data for one cell with applied pressure between 1 and 10 mmHg.

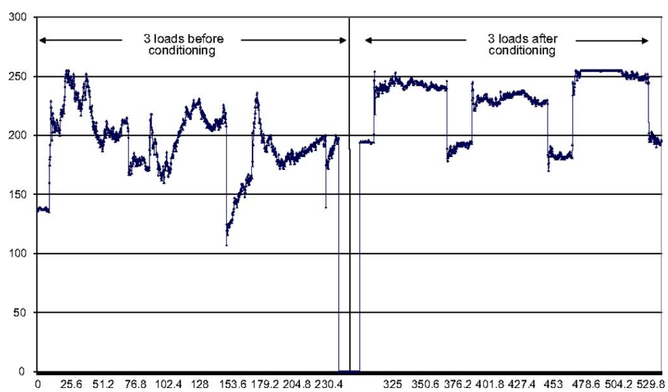

Fig. 5. Six loads of 7.8 mmHg: three applied before conditioning and three applied after conditioning.

B. Conditioning

Preconditioning the sensor showed evidence of regulating the measurement response. The response of the sensor varied prior to conditioning, whereas, after conditioning, the application and removal of loads is obvious with a more consistent response for each of the three loads (see Fig. 5). When assessing the magnitude of the conditioning load as a variable, it was found that after conditioning with 10.3 mmHg, the output for the 7.8 mmHg measurement took longer to reach a stable level compared with conditioning with the higher 25.9 or 51.7 mmHg load (which gave similar results). There was no significant effect of the length of the interval between conditioning and measurement (10, 30, or 60 min) on the output response.

C. Sensor Properties: Drift and Temperature

For the first 3 s, the sensor response was noisy (see Fig. 6). After this, some small amount of drift or creep occurred in the sensor output. The average time for all loads to remain within 10% of the 15–20 s average was 10.4 s.

The sensor's output increased only very slightly over the 17° temperature range. When comparing calibration curves at 23 °C and 36 °C, the error associated with calibrating at room

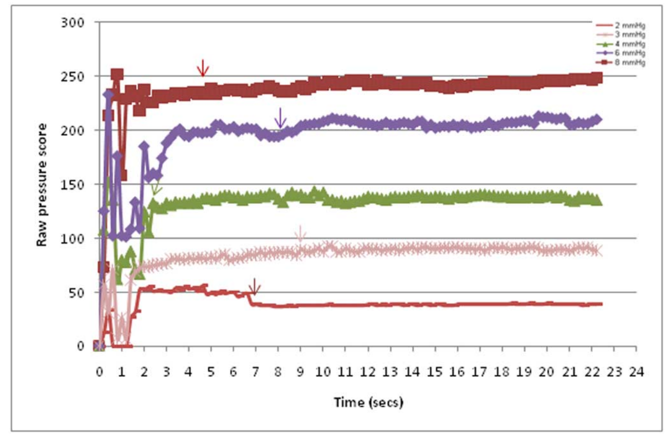

Fig. 6. Drift curves for loads of 2, 3, 4, 6, and 8 mmHg over approximately 22 s. Arrows indicate when the load remained within a 10% range around the 15–20 s average.

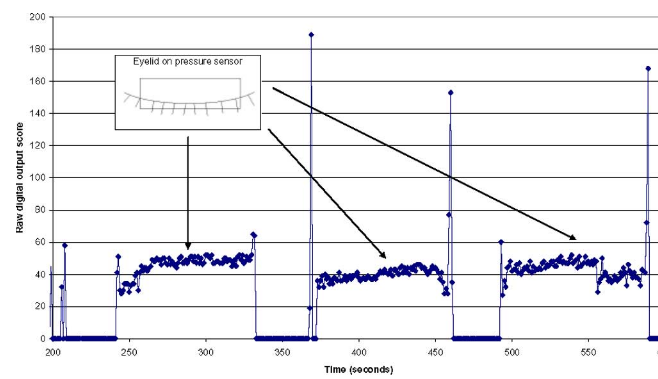

Fig. 7. Eyelid pressure measurement with the upper eyelid being placed on and off the sensor three times.

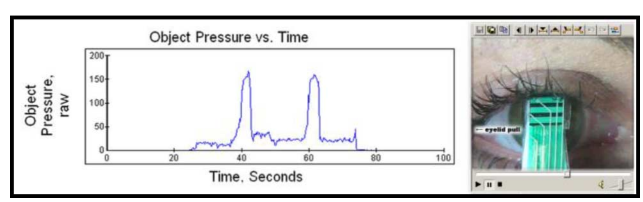

Fig. 8. Eyelids pulled twice to increase applied pressure.

temperature (23 °C) and measuring at ocular surface temperature (36 °C) was a slight overestimation of pressure (average 2.5%).

D. Examples of Upper Eyelid Pressure Measurements

Sample eyelid pressure measurements are shown in Fig. 7, where the upper eyelid is placed on the sensor and removed three times. This shows obvious and consistent response of the sensor each time the pressure is applied by the upper eyelid. A further example is the effect of tightening the upper lid, which is achieved by a technique similar to the lid-pull technique for removing a rigid contact lens. The sensor output shows that the pressure applied by the upper eyelid increases when the lid is pulled (see Fig. 8).

IV. Discussion and Conclusion

A novel system was developed to use Tekscan #4201 tactile pressure sensors for measuring upper eyelid pressure. This included designing a custom contact lens, trimming and resealing the pressure sensor, attaching it to the contact lens and support beam, and filing a flat peripheral area on the contact lens to which the sensor could be attached. Initially, the sensor–contact lens combination had the support beam perpendicular to the contact lens so that the sensor lay flat on the lens, but was bent to run along the support beam. However, pressure measurements were noisy, most likely due to the shearing effects between the back and front Mylar sheets of the sensor. When the support beam was attached at an angle, so that the sensor remained flat from the support beam onto the contact lens, the variability in measurements was significantly reduced.

There were a number of advantages of the custom-built hydrostatic calibration apparatus compared to commercially available systems. The sensor could be calibrated when attached to the contact lens, despite its thickness and shape. Using a plastic membrane at the end of the water column to contact and conform to the sensor surface closely resembles the contact applied by an eyelid. Also, there were no lower pressure limits imposed by the water column calibration apparatus, and this was important since we anticipated eyelid pressure to be relatively low. Calibration should be completed prior to every use of the sensors with a linear fit for pressure data between 1 and 10 mmHg.

The Tekscan I-scan manufacturer recommends that sensors that are new or have not been used for a length of time should be exercised by loading them three to five times. For best results, it is advised that the load be 20% greater than the maximum load to be applied in testing and should involve materials of compliance similar to the application. The benefits of conditioning are reduced drift and hysteresis, and increased reliability. The importance of conditioning was demonstrated with loads applied before and after prestressing the sensor (see Fig. 5). From investigation of the conditioning load and break time between calibration and measurement, it was concluded that the model #4201 sensors should optimally be conditioned with four loads of 25.9 mmHg for 1 min (with 30-s intervals between loads), less than 60 min prior to use.

Over the relatively short time required to measure eyelid pressure ($<$1 min), the influence of drift is insignificant provided that the time after loading is matched for calibration and measurement data (for example, between 10 and 30 s) and the first 10 s (when the output is unstable) are disregarded.

From the experiments concerning the influence of temperature, it was concluded that temperature does not have to be taken into account in the calibration and measurement of eyelid pressure. Only small errors were recorded when a measurement was taken at ocular surface temperature but calibrated with data recorded at room temperature. It is also questionable whether the sensor would heat up to ocular surface temperature while on the eye as the piezoresistive conductive ink inside the sensor is covered with Mylar plastic sheets. Mylar (polyester) is known for its excellent temperature resistance, with a coefficient of thermal conduction of 0.0001. So, it should act as an insulator for the pressure-sensitive ink. Unlike the temperature experiments where the whole sensor was placed in the incubator for a number of hours, for eyelid pressure measurements, the sensor–contact lens combination is only in contact with the eye and eyelid for a few minutes. Heat from the eye would also be absorbed by the Perspex contact lens, further limiting the influence of temperature variations on the sensor. Therefore, the effect of temperature on the Tekscan sensors reported in this study is most likely to be an overestimation of its influence.

Some electrical interference with the sensor output was observed from a fluorescent light, from the video camera liquid crystal display (LCD) panel, and when using metal clips to secure the sensor to the apparatus. The result was erratic readings from pressure cells without the application of pressure. To reduce this interference, the fluorescent ring light was replaced with a desk lamp, which could be positioned further away from the sensor and in proximity to the video cameras, was also avoided. Also, the sensor–contact lens combination was attached to the measurement apparatus with plastic clamps.

Several techniques have been developed to calibrate and measure upper eyelid pressure using piezoresistive sensors. Previous studies on measuring eyelid pressure were disadvantageous due to the complexity of the instrumentation and the techniques available. Using new piezoresistive sensors means that the total thickness of the device inserted between the cornea and eyelid is much smaller, being less than 0.7 mm (approximately 0.5 mm for the contact lens and 0.17 mm for the sensor). Evidence that the technique is able to measure upper eyelid pressure has been demonstrated in this study, though measurements have been reported in raw score values and not calibrated pressure units. Using the calibration equation for a pressure cell assumes that the entire cell is loaded by the eyelid margin (i.e., over a width of more than 1.14 mm). Current evidence suggests that the area of primary contact between the upper eyelid and eye surface is likely to be less than 1 mm [3], [12]. Once the contact area between the cornea and upper eyelid has been confirmed, eyelid pressure measurements can be scaled based upon this contact area.

Understanding the pressure exerted by the eyelids on the surface of the eye has a number of potential clinical applications. Trials using this piezoresistive sensor–contact lens system have demonstrated that the technique is able to measure the static eyelid pressure of the upper eyelid. Modifications of the system will be required to study lower eyelid pressure, closed eyelid pressure, and the pressure applied during the dynamic eyelid movements of blinking. The methods described in this paper provide the basis for new techniques for acquiring accurate information about eyelid pressure.

References

[1] M. G. Doane, "Interaction of eyelids and tears in corneal wetting and the dynamics of the normal human eyeblink," *Amer. J. Ophthalmol.*, vol. 89, no. 4, pp. 507–516, 1980.

[2] G. Hung, F. Hsu, and L. Stark, "Dynamics of the human eyeblink," *Amer. J. Optom. Physiol. Opt.*, vol. 54, no. 10, pp. 678–690, 1977.

[3] D. R. Korb, J. V. Greiner, J. P. Herman, E. Hebert, V. M. Finnemore, J. M. Exford, T. Glonek, and M. C. Olson, "Lid-wiper epitheliopathy and dry-eye symptoms in contact lens wearers," *CLAO J.*, vol. 28, no. 4, pp. 211–216, 2002.

[4] C. B. Cosar, C. J. Rapuano, E. J. Cohen, and P. R. Laibson, "Chalazion as a cause of decreased vision after LASIK," *Cornea*, vol. 20, no. 8, pp. 890–892, 2001.
[5] M. Nisted and H. W. Hofstetter, "Effect of chalazion on astigmatism," *Amer. J. Optom. Physiol. Opt.*, vol. 51, no. 8, pp. 579–582, 1974.
[6] D. A. Plager and S. K. Snyder, "Resolution of astigmatism after surgical resection of capillary hemangiomas in infants," *Ophthalmology*, vol. 104, no. 7, pp. 1102–1106, 1997.
[7] R. M. Robb, "Refractive errors associated with hemangiomas of the eyelids and orbit in infancy," *Amer. J. Ophthalmol.*, vol. 83, no. 1, pp. 52–58, 1977.
[8] T. Buehren, M. J. Collins, and L. Carney, "Corneal aberrations and reading," *Optom. Vis. Sci.*, vol. 80, no. 2, pp. 159–166, 2003.
[9] T. Buehren, M. J. Collins, D. R. Iskander, B. Davis, and B. Lingelbach, "The stability of corneal topography in the post-blink interval," *Cornea*, vol. 20, no. 8, pp. 826–833, 2001.
[10] M. J. Collins, T. Buehren, A. Bece, S. C. Voetz, "Corneal optics after reading, microscopy and computer work," *Acta Ophthalmol. Scand.*, vol. 84, no. 2, pp. 216–224, 2006.
[11] M. J. Collins, T. Buehren, T. Trevor, M. Statham, J. Hansen, and D. A. Cavanagh, "Factors influencing lid pressure on the cornea," *Eye Contact Lens*, vol. 32, no. 4, pp. 168–173, 2006.
[12] A. J. Shaw, M. J. Collins, B. A. Davis, and L. G. Carney, "Eyelid pressure: Inferences from corneal topographic changes," *Cornea*, vol. 28, pp. 181–188, 2009.
[13] A. J. Shaw, M. J. Collins, B. A. Davis, and L. G. Carney, "Corneal refractive changes due to short-term eyelid pressure in downward gaze," *J. Cataract Refract. Surg.*, vol. 34, no. 9, pp. 1546–1553, 2008.
[14] D. Miller, "Pressure of the lid on the eye," *Arch. Ophthalmol.*, vol. 78, no. 3, pp. 328–330, 1967.
[15] D. Lydon and A. Tait, "Lid pressure: Its measurement and probable effects on the shape and form of the cornea–rigid contact lens system," *J. Br. Contact Lens Assoc.*, vol. 11, no. 1, pp. 11–22, 1988.
[16] H. Shikura, T. Yamaguchi, and S. Nakajima, "A new system for measuring the pressure between the eyelids and the cornea," *Invest. Ophthalmol. Vis. Sci.*, vol. 34, p. 1250, 1993.
[17] J. Otto, T. Brown, and J. Callaghan, "Static and dynamic response of a multiplexed-array piezoresistive contact sensor," *Exp. Mech.*, vol. 39, no. 4, pp. 317–323, 1999.
[18] D. R. Wilson, M. V. Apreleva, M. J. Eichler, and F. R. Harrold, "Accuracy and repeatability of a pressure measurement system in the patellofemoral joint," *J. Biomech.*, vol. 36, no. 12, pp. 1909–1915, 2003.
[19] M. Ferguson-Pell, S. Hagisawa, and D. Bain, "Evaluation of a sensor for low interface pressure applications," *Med. Eng. Phys.*, vol. 22, no. 9, pp. 657–663, 2000.
[20] A. A. Polliack, R. C. Sieh, D. D. Craig, S. Landsberger, D. R. McNeil, and E. Ayyappa, "Scientific validation of two commercial pressure sensor systems for prosthetic socket fit," *Prosthet. Orthot. Int.*, vol. 24, no. 1, pp. 63–73, 2000.
[21] S. A. Read, M. J. Collins, L. G. Carney, and R. J. Franklin, "The topography of the central and peripheral cornea," *Invest. Ophthalmol. Vis. Sci.*, vol. 47, no. 4, pp. 1404–1415, 2006.
[22] F. Rüfer, A. Schröder, and C. Erb, "White-to-white corneal diameter: Normal values in healthy humans obtained with the Orbscan II topography system," *Cornea*, vol. 24, no. 3, pp. 259–261, 2005.
[23] M. G. Harris and C. S. Chu, "The effect of contact lens thickness and corneal toricity on flexure and residual astigmatism," *Amer. J. Optom. Arch. Amer. Acad. Optom.*, vol. 49, no. 4, pp. 304–307, 1972.
[24] G. Y. Leung, V. Peponis, E. D. Varnell, D. S. Lam, and H. E. Kaufman, "Preliminary in vitro evaluation of 2-octyl cyanoacrylate (Dermabond) to seal corneal incisions," *Cornea*, vol. 24, no. 8, pp. 998–999, 2005.
[25] S. A. Read, M. J. Collins, L. G. Carney, and D. R. Iskander, "The morphology of the palpebral fissure in different directions of vertical gaze," *Optom. Vis. Sci.*, vol. 83, no. 10, pp. 715–722, 2006.
[26] J. M. Brimacombe, C. Anglin, A. J. Hodgson, and D. R. Wilson, "Validation of calibration techniques for tekscan pressure sensors," presented at the ISB XXth Congr., ASB 29th Annu. Meeting, 2005, Cleveland, OH.
[27] A. L. Randolph, M. Nelson, S. Akkapeddi, A. Levin, and R. Alexandrescu, "Reliability of measurements of pressures applied on the foot during walking by a computerized insole sensor system," *Arch. Phys. Med. Rehabil.*, vol. 81, no. 5, pp. 573–578, 2000.
[28] P. B. Morgan, M. P. Soh, N. Efron, and A. B. Tullo, "Potential applications of ocular thermography," *Optom. Vis. Sci.*, vol. 70, no. 7, pp. 568–576, 1993.
[29] C. Purslow and J. Wolffsohn, "The relation between physical properties of the anterior eye and ocular surface temperature," *Optom. Vis. Sci.*, vol. 84, no. 3, pp. 197–201, 2007.
[30] C. Purslow, J. S. Wolffsohn, and J. Santodomingo-Rubido, "The effect of contact lens wear on dynamic ocular surface temperature," *Contact Lens Anterior Eye*, vol. 28, no. 1, pp. 29–36, 2005.
[31] B. A. Holden and D. F. Sweeney, "The oxygen tension and temperature of the superior palpebral conjunctiva," *Acta Ophthalmol.*, vol. 63, no. 1, pp. 100–103, 1985.

Alyra J. Shaw received the Bachelor of Applied Science degree (in optometry) from Queensland University of Technology (QUT), Brisbane, Qld., Australia, in 2001. She is currently working toward the Ph.D. degree at the School of Optometry, QUT.

Her current research interests include eyelid anatomy and pressure and the optical properties of the cornea.

Brett A. Davis received the Bachelor of Applied Science degree (in physics) from Queensland University of Technology (QUT), Brisbane, Qld., Australia, in 1990.

He is a Senior Research Assistant with the Contact Lens and Visual Optics Laboratory, Institute of Health and Biomedical Innovation, QUT. He has been engaged various research projects and interests include visual optics and optical design.

Michael J. Collins received the Dip.App.Sc. (in optometry), M.App.Sc., and Ph.D. degrees from Queensland University of Technology (QUT), Brisbane, Qld., Australia, in 1977, 1988, and 1996, respectively.

He is currently a Professor at the School of Optometry, QUT. He is involved in the research of the visual and optical characteristics of the cornea and contact lenses at the Contact Lens and Visual Optics Laboratory.

Prof. Collins is a member of the Optometrists Association of Australia, and a Fellow of the American Academy of Optometry and the Contact Lens Society of Australia.

Leo G. Carney graduated from the Department of Optometry, University of Melbourne, Melbourne, Vic., Australia, from where he received the M.Sc. and Ph.D. degrees in optometry, in 1970 and 1974, respectively, and the Dr. Sci. degree from Queensland University of Technology, Brisbane, Qld., Australia, in 2001.

He is currently a Professor Emeritus at the School of Optometry, Queensland University of Technology (QUT), Brisbane, Qld., Australia. He was the Head of the School of Optometry from 1992 until 2007. His current research interests include investigation of the physiological and optical evaluation of the anterior eye and the effects of contact lens wear on the structures of the eye. He has authored or coauthored about 200 papers dealing with aspects of anterior eye physiology, contact lenses, and visual optics.

Prof. Carney is a Councillor of the International Society for Contact Lens Research and a Fellow of the American Academy of Optometry and the Contact Lens Society of Australia.