INTRODUCTION

Glaucoma is the leading cause of irreversible blindness, affecting more than 60 million people worldwide. Glaucoma is a group of optic neuropathy diseases in which the optic nerve is damaged due to an unwanted force applied from a build-up of pressure within the eye. In a healthy eye, there is a continuous production of aqueous humour, a clear fluid which helps maintain the eye shape for proper refraction and provides nutrients to the area. The aqueous humour is produced by the ciliary bodies which then flows through the eye then eventually is drained out through the trabecular meshwork and exists out of the eye through the Channel of Schlemm into the episcleral blood vessels (Johnson & Kamm, 1983). In cases of glaucoma, this outflow pathway becomes increasingly blocked overtime causing a backup of fluid within the eye leading to an increase in pressure. Overtime this increase pressure causes the thinning of the neuro-retinal rim and the deterioration of retinal ganglion cells. A normal range of intraocular pressure (IOP) is between 10 and 22 mmHg with the probability of diagnosis of glaucoma reaching nearly 100% when approaching 35 mmHg; anything less than 10 mmHg is an issue of the eye being under deflated by an underproduction of aqueous humour (Davanger, Ringvold, & Blika, 1991; Thomas, Vajaranant, & Aref, 2015).

Monitoring the IOP is the primary form to track the progression of glaucoma, and Goldmann applanation tonometry (GAT) is the current gold standard for monitoring the patient's IOP levels (Arend, Hirneiss, & Kernt, 2014). This in office procedure is completed with a device that measures the force required to flatten a specified area of the cornea. The GAT requires local anaesthesia and a trained professional to perform the test. As a result, patients with glaucoma generally have their IOP measured in a clinical setting every few months and only within typical office hours. However, IOP levels have been found to vary as much as 10 mmHg or more in a single day due to multiple factors such as activity levels, nutrients, blood pressure and body position (Arora et al., 2014; Kim & Caprioli, 2018). In recent years, there have been more advancements in continuous IOP monitoring systems to get a better understanding of the progression of glaucoma.

A recent implantable device was designed to monitor the real change in pressure using an intraocular lens. The patient would...
undergo a surgical procedure to implant the device into the eye either as a stand-alone procedure or during a cataract procedure. Embedded into the intraocular device is a channel with one end open to the aqueous environment of the eye and the other a closed end reservoir filled with gas. As the patient’s IOP increases, the aqueous humour fluid within the eye would flow into the embedded channel within the intraocular device until equilibrium is reached. There are ruler marks on the device along the channel in order to correlate the location of the fluid–air interface directly back to the IOP. Multiple images can be taken throughout the day, using a cellphone camera with a microscope attachment, to gather more information on the progression of the disease (Araci, Su, Quake, & Mandel, 2014). A different device was created which utilized a parylene-based sensor coated with titanium and gold which could read the IOP based on the impedance-phase shift measured. The sensing portion of the device was to lay on the surface of the eyeball, underneath the conjunctiva layer, with an implantable tube punctured into the posterior chamber of the eye. As the IOP fluctuates, the applied pressure flows up the implantable tube and deforms the parylene-based sensor and changes the monitored impedance (Chen, Rodger, Saati, Humayun, & Tai, 2008). A micro-implantable device that is surgically placed in the anterior portion of the eye utilizes an optical cavity that when excited by broadband near-infrared light the sensor would reflect a pressure-dependent resonance signature that could be related back to the IOP. As pressure increases, there is a blue-shifted spectrum, and when pressure decreases, there is a red-shifted spectrum correspondence. Trials were completed on rabbits, and positive results were obtained for monitoring the IOP fluctuations over a prolonged period of time. However, at the current stage reading the IOP sensor requires a technically skilled professional with a large apparatus that cannot be done at home (Lee et al., 2017). These devices give more direct IOP readings however do require surgical intervention which adds risk to the patient.

Non-invasive IOP monitoring systems have been explored by utilizing soft contact lenses with embedded strain gauges allowing for the changes in the corneal curvature to be measured and related directly back to the IOP (Leonardi, Leuenberger, Bertrand, Bertsch, & Renaud, 2004). In some instances, a rigid contact lens was modified to include a circular patch of piezoresistive bilayer film base membrane, able to detect deformations in the eye caused by an internal pressure change. This device was able to continually monitor the patient’s IOP down to a fluctuation of 1 mmHg; to obtain, this accuracy each lens was required to be calibrated and tailored to each patient with dependencies on the corneal radius, corneal thickness and hydration level of the eye (Sanchez et al., 2011). Sensimed Triggerfish is an approved continuous monitoring contact lens able to detect changes within the corneal deformation and utilize this deformation as an indicator to the progression of the IOP. The device has a 7-μm-thick circular sensing resistive gauges embedded within soft silicone contact lens ranging from just under 600 to 250 μm thick from centre to peripheral edge. The disposable device monitors the fluctuations between the embedded sensors with arbitrary units which are sent via an antenna embedded into the contact lens to an external antenna placed around the outside of the eye which is then stored in a portable recorder the patient must carry (Mansouri & Shaarawy, 2011). This device is estimated to cost between $526 and 682 in single-use equipment with an additional cost of $7,310 for a reusable data recorder, recorder cable and software, to record and track the data (Dunbar, Shen, & Aref, 2017). The non-invasive IOP monitoring systems indirectly determine the IOP however are easier and safer to use for patients, as no surgical intervention is required.

This paper presents a new inexpensive, non-invasive IOP monitoring system that utilizes microfluidics within a soft contact lens to indirectly monitor the fluctuations throughout the day. The contact lens will be able to deform uniformly with the changing curvature of the eye as the IOP fluctuates. An indicator fluid within the microfluidic channel can shift in position with respect to the deforming volume as a result of a change in curvature of the eye. Marks embedded within the contact lens will allow the user to track the change in location of the fluid and take readings multiple times a day (Campigotto, Leahy, Zhao, Campbell, & Lai, 2019; Lai, Xie, & Campbell, 2017). A main advantage of utilizing microfluidic channels embedded within a contact lens is the inexpensive manufacturing cost.

2 | DESIGN

The proposed IOP sensor is in the form of a soft contact lens which can be worn by patients as a regular contact lens throughout the day and disposed of after use. There is a microfluidic channel embedded within the contact lens, partially filled with incompressible fluid that gives a visual representation of the IOP, based on the position of the fluid. The sensor is created using Sylgard 184 Polydimethylsiloxane (PDMS) with an embedded 100 μm by 100 μm microchannel which form a double spiral design. The overall thickness of the contact lens device is 250 μm, and the lens has an overall diameter of 14.2 mm. The microfluidic channel was placed around the outer edge of the contact lens with a clearance of 4 mm away from the centre of the lens, as to not obstruct the visual field of the user; this allows for the average maximum pupil diameter of 8 mm, in low light (Spector, 1990). A compromise in the channel size was made between increasing the fluid movement with a smaller channel diameter and making the channel large enough to manufacture and decrease the error in tracking fluid location during postprocessing. To maximize the initial area of the microchannel, a spiral design was created with a tapered inlet and outlet for ease of manufacturing. The inlet and outlet are left open as it was determined that the fluid is able to move much further when utilizing that pressure differences that occur at those entrances. The spiral lens shape increases the initial volume of the channel while limiting areas with high energy concentrations which will potentially affect the fluid flow.

Polydimethylsiloxane (PDMS) was chosen as the contact lens material as it is a known biocompatible material already used in contact lenses and does not cause irritation or infection to the human eye. PDMS has the advantage of being highly oxygen and water permeable, allowing for improved patient comfort, optically transparent,
and is easy to manipulate during fabrication to obtain ideal material properties (Guidi, Hughes, Whinton, Brook, & Sheardown, 2014; Lamberti, Marasso, & Cocuzza, 2014). However, PDMS is hydrophobic by nature, and in order to have proper interaction with the tear film within the eye, the lens must be hydrophilic (Nichols & Sinnott, 2006; Trantidou, Elani, Parsons, & Ces, 2017). Therefore, a temporary solution of surface treating the PDMS with a Corona Plasma Surface Treater was used to change the surface to become more hydrophilic (Eddings, Johnson, & Gale, 2008). The incompressible fluid used during initial experimentation was avocado oil as it is dark enough in colour to be visibly seen in the microchannel and has a good interaction with the PDMS as it does not diffuse out of the PDMS, further reasoning for the use of avocado oil for initial testing is explained below. However, further research must still be conducted to determine the finalized indicator fluid before the device can be used in the human eye.

The concept of this device relies on the premise that according to an in vivo study conducted, there is a correlation of the IOP and the corneal curvature in humans of approximately 3 µm change per 1 mmHg for a radius of curvature of 7.8 mm (Douthwaite & Lam, 1997; Hjortdal & Jensen, 1995). Another study completed found that the sclera curvature deforms further with a correlation of approximately 100 µm change in curvature for a 1 mmHg increase (Pierscionek, Asejczyk-Widlicka, & Scha, 2007). The size of the contact lens device used in this paper is just larger than an average human cornea allowing the device to take advantage of the scleral-corneal junction that has been determined to have the largest amount of deformation while still providing patients with a small enough lens that can fit comfortably in the eye (Hjortdal & Jensen, 1995). As the curvature of the eye changes, the contact lens will mimic the deformation and either increase or decrease the volume of the embedded microchannel, forcing the fluid to displace. The movement of the fluid is then directly correlated back to the patient’s IOP (Lai et al., 2017). Makers in the contact lens allow for calibration when multiple images of the lens are taken (Figure 1).

Compared to other devices currently designed for IOP measuring, this microfluidic contact lens is inexpensive to manufacture and thinner in overall dimension, while still able to provide multiple IOP readings throughout the day. Thicker contact lenses have been shown to decrease the oxygen provided to the eye and when worn for extended periods of time or overnight the amount of oxygen going to the eye can cause changes in the corneal thickness (Braun & Anderson Penno, 2002). Removing any mechanical components allows for the contact lens to be manufactured much thinner improving the comfort and decreasing the overall cost. As well, tracking data and recording the points will be completed using a cellphone application that takes an image of the contact lens and determines

**FIGURE 1**  a, The progression of glaucoma from a healthy eye on the far left showing an undisturbed flow of aqueous humour through the eye in blue. As the outflow of aqueous humour becomes blocked, there is an increase in pressure which is directly correlated with a linear increase in the curvature of the eye. As this pressure build-up continues, a harmful force is permanently damages the optic nerve. b, An illustration of the microfluidic spiral lens described throughout this paper. As the curvature of the cornea changes with respect to the change in the IOP, the indicator fluid moves along the channel. The embedded channel within the lens will increase in volume with the increase in the corneal curvature, creating a suction effect that draws the fluid inward. The reserves affect can be shown with a decrease in the corneal curvature. Reference markers placed along the microchannel help in after processing to track the position of the indicator fluid and directly relate that displacement back to the fluctuation in the IOP.
the distance the fluid travels relative to the reference markers. Therefore, no additional equipment is required for the user to obtain at home readings of their IOP.

In future application of the device, the user will be fitted with a microfluidic contact lens and has initial readings of their IOP taken with a GAT. This will be used as the reference point, and from there, the fluctuations can be tracked based on the movement off the indicator fluid within the lens. The more frequently images are taken, the more data can be gathered to determine the fluctuations of the patients IOP. This can provide physicians with more information on how to optimize the patient’s treatment based on the progression of the disease.

## 3 | SIMULATION

Simulations were performed to determine the effect on the deformation of the microfluidic channel within the contact lens when internal pressure is applied to the eye. A finite element analysis (FEA) was performed using ANSYS and simulated as a frictional contact system between the contact lens and a human eye with a static friction coefficient of 0.42, based on balafilcon A PureVision contact lens, which was used as reference for the design of the microfluidic contact lens (Hill, Roba, Duncan, Tosatti, & Zurcher, 2012). The eye was simplified to be the cornea and half the sclera with no internal structures; this allowed for a symmetry constraint to be placed on the edge of the sclera, and a centre point in the cornea was fixed to only allow vertical displacement. The cornea and the sclera were assumed to have the same material properties with a linear elastic behaviour and a young’s modulus of 0.3 MPa and Poisson’s ratio of 0.49. The PDMS was assumed to have a young’s modulus of 1.32 MPa and a poison’s ratio of 0.49; although PDMS is a hyperplastic material and can undergo large deformations, for simplifications sake the simulation assumed a linear elastic material as the strain the system undergoes is minimal (Houaria & Chellali, 2014; Johnston, McCluskey, Tan, & Tracey, 2014; Nunes, 2011; Schneider, Draheim, Kamberger, & Wallrabe, 2009; Shahzad, Kamran, & Siddiqui and M. Farhan, 2015). The Young’s Modulus used for the PDMS was based on previously conducted research on the affect of PDMS curing temperature and curing ration on the material properties. The Young’s Modulus of PDMS will increase at higher temperatures used during the curing process and at lower curing ratios (Johnston et al., 2014; Wang, Volinsky, & Gallant, 2014). Figure 2 shows the results of the deformation within the system when a 10 mmHg pressure is applied along the inside face of the of the sclera and cornea.

The eye was modelled based on the dimensions of an average human eye, with the cornea having an assumed thickness of 550 µm and an anterior and posterior curvature of radius of 10.55 and 9.4 mm, respectively (Bekerman, Gottlieb, & Vaiman, 2014; Olsen, Aaberg, Geroski, & Edelhauser, 1998). While the sclera had an assumed thickness of 530 µm and overall diameter of 24.2 mm (Bekerman et al., 2014; Olsen et al., 1998), the simulation was completed with an applied pressure to the inside of the eye of 15, 20, 25, 30 and 35 mmHg. The CAD and STL files of the deformed models were obtained and analysed to determine the deformation of the embedded microchannel. A proportional, linear response was observed; as the internal applied pressure increased, the area of the microchannel increased with a rate of 0.15 mm²/mmHg, as shown in Figure 3. There was no significant observed thickness change in the contact lens between the various applied pressures; therefore, only the deformation of area within the microchannel was considered important.

As the pressure increases, so does the area of the microchannel in the contact lens, indicating that the structure is being stretched, with a total change in the area of the microchannel by approximately 28.6% between the lowest and highest applied pressure. To further illustrate the affect of the increasing total lens deformation on the embedded microchannel, the deformed models from ANSYS were exported and examined in MeshLab. The image of the left, in red, is the deformed contact lens after 15 mmHg of internal pressure was applied, and the image on the middle, in blue, is when 35 mmHg of internal pressure was applied (Figure 4). When overlaying the two deformed images, by setting specified markers for proper alignment, it can be visually noticed that the size of the channel increases in

![Figure 2](image-url)  
**Figure 2** Illustration of the ANSYS simulation performed with an internal applied pressure on the eye of 10 mmHg

![Figure 3](image-url)  
**Figure 3** The area of the embedded microchannel as a result of varying applied pressures to the internal portion of the cornea. As the applied pressure increases, the area deformation of the embedded microchannel of the contact lens is increased. A linearly proportional slope of 0.15 mm²/mmHg was determined
area, with the most amount of area changes at the inner tip of the spiral, highlighted with a red circle (Figure 4).

The simulation demonstrated that as the IOP increases the deformation of the contact lens increases which leads to an increase in area within the microchannel. Such an increase in microchannel area will make the incompressible indicator fluid to occupy shorter length in the channel. Under the surface tension force, the fluid is expected to flow inward towards the centre of the channel, whereas as the IOP decreases and the microchannel size decreases which will make the fluid flow towards the outlets. As such, the directions and distance of the fluid will travel can be used to indicate IOP fluctuations.

4 | FABRICATION

The contact lens is manufactured using a casting process with the moulds (see Figure 5) manufactured with a high-speed micromilling machine (Microlution 363-S). The young’s modulus of most soft contact lenses is between 0.39 and 1.52 MPa; therefore, the PDMS curing ratio and curing temperature and time were chosen in order to obtain a contact lens within this range (Childs et al., 2016). A cast-mould-based process was chosen for its ease to incorporate with the PDMS and obtain unique embedded microchannels. This process is already used as a form to manufacture current on market contact lenses so would be easily incorporated with other contact lens materials currently used, if further modifications are of interest (Boneberger, Haase, Schäfer, Stefan, & Zang, 1999; Evans & Thakrar, 1992; Manfredini, 2000; Yamada, 2006). Finally, the microfluidic contact lens would be a 24-hr disposable lens, so there is high emphasis put on ensuring the repeatability of results from lens to lens. Using a bulk mould-based system would allow for high uniformity across all manufactured lenses, which has been demonstrated in a previous study conducted by the research team on non-invasive Intraocular pressure monitoring with a contact lens, describing the repeatability of a similar lens material with a different microfluidic channel design (Campigotto et al., 2019).

The mould was placed in a vacuum chamber for 40 min for degassing, with additional degassing required if necessary, until all the air bubbles had disappeared. The mould was placed into the oven at 50°C for 40 min and was cooled to room temperature, and then, the sensor body was released from the mould. A thin top layer was created by spin-coating PDMS on a flat surface and then curing the PDMS in the oven at 50°C for 30 min (Chen et al., 2008). The curing ratio, temperature and time of heating the mould are important parameters that were completed carefully, as these will change the material properties of the PDMS contact lens if inconsistent. The two pieces of PDMS were then bonded together to form the complete contact lens device (Eddings et al., 2008). Two PDMS-PDMS bonding methods were examined: using semi-cured PDMS bonded with a fully cured PDMS, versus both fully cured and bonded with the Corona Surface Treater. Using the semi-cured PDMS bonded to the fully cured PDMS provides better PDMS-PDMS adhesion, and the oil-to-PDMS interface has a high contact angle. The Corona Surface Treater increases the surface strength of PDMS by activating the layers of cross-linked PDMS. The silanol groups (OH) are exposed at the surface of PDMS and when adhered together form covalent siloxane bonds (Si-O-Si), also making the surface to become more wettable (Eddings et al., 2008). Further information on the experimentation is provided in Hydrophilic Testing in the following section.

4.1 | Indicator fluid

The indicator fluid used within the channel is a major factor in improving the performance of the contact lens sensor. For proof-of-concept testing, initially it was thought to utilize water, dyed with food colouring for easier visual tracking, as water is the main component of tear or aqueous humour. However, upon setting up the

![Figure 4](https://example.com/figure4.jpg) Using MeshLab the deformed models with a simulated internal pressure of 15 and 35 mmHg analysed in ANSYS was directly compared to each other. The far left displays the deformation at a simulated pressure of 15 mmHg, the middle displays the deformation at a simulated pressure of 35 mmHg, and the right displays the result when the two deformations are overplayed with the 15 mmHg shown in red and the 35 mmHg shown in blue. As it can be visually illustrated, the higher applied pressure to the eye, the larger the internal channel becomes. This is highlighted in red where the most amount of volume change occurs. The orange markers are the points set on each of the images to insure proper alignment.
experiment, there were already noticeable gaps in the water forming throughout the microchannel. As the topcoat of PDMS is only 50 µm thick, the water is able to diffuse out of the channel at such a quick rate that the results of experimentation would be skewed. PDMS membrane is less permeable to oil since oil molecules are generally larger than water molecules. It is expected that oil-based indicator fluid will be more stable. Considering that avocado oil has much lighter viscosity than most other oils, and it was readily available, could be tracked visually in the microchannel, and would interact safely with the PDMS and the porcine eyes, the indicator fluid was switched to avocado oil. Figure 6 shows the rate of diffusion of water and oil out of a PDMS microchannel with equivalently thick membranes and channel sizes. After 10 min, noticeable diffusion of water out of the microchannel was observed, whereas with the oil, no noticeable diffusion occurred after 60 min. As such, avocado oil is selected as the indication fluid for the initial prototype testing.

The experimentation conducted below in the Enucleated Porcine Experiment section took around one hour for each test; therefore, it can be noted that the diffusion of the oil did not affect the results.

Ideally, the indicator fluid used would have a low viscosity to flow easier through the channel, be visually opaque, biocompatible, and not be able to diffuse out of the PDMS. Further, examination into the indicator fluid will be completed to further optimize the sensor before used in practical application.

4.2 | Hydrophilic testing

A test was performed, using a syringe pump, where a small amount of coloured oil was pushed through a 5-mm, straight microchannel,
that either has or has not been plasma surface treated. The surface-
treated system has a lower contact angle between the surface and
the oil and had a flow rate of 0.011 mm$^3$/sec, while the non-surface-
treated surface has a higher contact angle with the oil and had a
flow rate of 0.009 mm$^3$/sec. Therefore, the plasma surface-treated
system was approximately 1 s faster, flowing through the apparatus,
than the non-plasma surface-treated system; lowering the contact
angle between the oil and the surface with plasma surface treatment
allows for a faster flow rate by approximately 0.0023 mm$^3$/sec.

5 | EXPERIMENTAL TESTING

5.1 | Enucleated Porcine Experimentation

Testing was conducted on fresh enucleated porcine eyes obtained
from a local butcher. To ensure that the enucleated eyes performed
accurately to a living specimen, the experiments were conducted within
24–36 hr. While the specimen were not being used, they were kept in a
cup of water and placed in the fridge approximately 5°C, to prevent the
eye from drying out and to slow the rate of deterioration (Girard, Suh,
Hart, Burgoyne, & Downs, 2007; Nibourg & Koopmans, 2014).

For each porcine eye, a single spiral contact lens was tested
twice. A Millar Mikro Pressure Catheter was threaded into the cen-
tre of the posterior chamber of the porcine eye to record the rea-
time IOP as a syringe pump controlled the inflow of water into
the posterior chamber of the eye to simulate the fluctuating IOP.
The spiral contact lens was centred in the middle of the cornea and
hydrated with drops of water throughout the experiment to prevent
the eye from drying out and keep proper lubrication between the
eye and the contact lens. The system was recorded using a USB cam-
era, with a resolution of 640 X 480 pixels, from above and from the
side to ensure the porcine eye was properly inflating. Figure 7 shows
a schematic of the experimental set-up used.
The pressure was increased from 10 to 34 mmHg and deflated manually with a syringe back to 10 mmHg. This process was conducted twice on each eye with a single spiral contact lens with a total of four eyes tested. Figure 8 illustrates the fluid movement at various known IOP captured from the top view of the USB.

It was observed that as the pressure increased, the fluid would move inward in the microchannel, which supports the conclusion, drew from FEA simulation that as the pressure increases, the channel size increases and the indicator fluid moves inward.

The enucleated porcine experimentation was completed at room temperature of 25°C. The temperature difference between the experimental set-up and the human body would be only approximately 10°C, so it is assumed to have no effect on the performance of the sensor. Once the PDMS is cured, the material properties are set and will not be changed. As for the indicator fluid inside the behaviour may be affected by the extreme temperature changes, such as below the freezing point or above the boil point, however, the human body regulates the temperature of the cornea to between 28 and 36°C, when the ambient temperature ranges between 0 and 45°C (Fabiani, Li Voti, Rusciano, Mutolo, & Pescosolido, 2016; Kessel, Johnson, Arvidsson, & Larsen, 2010). In real practice, it is assumed that these IOP readings will be completed by the user in the comfort of their home with a standard temperature range ranging from 22 to 28°C, meaning the corneal temperature should not vary too much.

6 | RESULTS AND DISCUSSION

The fluid movement in the spiral contact lens was plotted every 2 mmHg increase in IOP and again once the eye had been deflated back down to 10 mmHg. The experimentation was to evaluate the accuracy of the contact lens to track the increase in IOP, as well as the ability of the fluid to return back to its original position once the pressure was manually decreased back to 10 mmHg (Figure 9). The error was determined based on the standard deviation of the two trials performed for each eye.

The indicator fluid movement was found to have a positive, linear relationship with the increasing IOP with slopes of 44.9, 38.3, 40.7 and 39.3 µm/mmHg. The average indicator fluid movement per IOP was 40.8 µm/mmHg with a standard deviation of 29.4 between all the trials completed. The returnability of the fluid was 90%, 96% 92% and 83% for eye 1, 2, 3 and 4, respectively. This was determined by looking at the fluid position at the start of the testing when the pressure was 10 mmHg and at the end of the testing when the pressure was deflated manually back to 10 mmHg.

7 | CONCLUSION

The development of an inexpensive, continuous IOP monitoring system is greatly beneficial and necessary to the improvement of glaucoma patient’s lives around the world. The contact lens evaluated in this report, has promising results of capturing the mechanical strains undergone by the eye and outputs the intraocular pressure, accurately and in real time.

Overall, the contact lens-based IOP monitoring device appears able to provide robust measures of fluctuations in IOP. The indicator position is linearly responsive, highly sensitive and provides consistent measures across eyes and individual devices, to the point that the indicator fluid can displace noticeably by the naked eye. Based on testing of multiple porcine eyes with a single contact lens, the flexibility of the device is suggested to be able to overcome the variations in the corneal shape to allow consistent performance despite small anatomic differences among eyes. This implies that the device can be a universal fit for different human eyes with little effect on results. Future work will need to be completed on optimizing the indicator fluid, to a less viscous fluid that is still visually trackable with the naked eye and that is not able to diffuse out of the PDMS. Based on initial prototype testing, it can be determined that the sensor works with an oil, so the device will only be improved upon with a low viscous fluid used, as well as incorporating a cellphone application that can track the progression of the indicator fluid and instantly provide updates to both the patient and the physician. More information on the fluctuations of the IOP provides the physician with a better understanding of the progression of the disease, allows for higher accuracy of
catching the disease earlier on and allows the physician to providing more user unique treatments. The novel non-invasive contact lens-based device evaluated here can be manufactured at low cost and holds promise in the prevention of glaucoma-related vision loss and blindness.

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