Abstract.

Bioinspired lenses that rely on changes of curvature to achieve focus are interesting candidates for miniaturised tuneable lenses as they require fewer mechanical moving parts compared to their conventional counter-parts. The lens described in this manuscript closely mimics the design and actuation principle of the vertebrate lens. It consists of a liquid lens encapsulated in a transparent polymer membrane. Application of a radial strain changes the curvature of the lens thereby changing its focal length. The unstrained lens has a focal length of 50 mm, which rises to a value of 100 mm at a maximum radial strain of 0.67 %. This range compares favourably to both biological lenses and other published examples of biomimetic lenses. Finally we point out a few routes to improve the quality of the lens and expand its focal length range.

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1. Introduction

Conventional tuneable lenses usually rely on the displacement of individual lenses to achieve focus. However the miniaturisation of such systems represents a major challenge. The difficulties reside in the fabrication and assembly of small part, and is most notably limited by the increased significance of surface friction at small length scales. Alternative optical systems, inspired by the mechanism of accommodation in the vertebrate eye, achieve focus by a change of lens shape. These systems contain fewer moving parts, and so, are better suited to miniaturisation and as such a wide
range of bioinspired lenses have already been developed. Compared to other bioinspired lenses, the lens presented in this manuscript, made using a simple process consisting of the encapsulation of a liquid in a flexible polymer membrane, and actuated using a radial stretching mechanism, mimics closely the design and actuation principle of the vertebrate lens. These features make this new lens a good candidate for a variety of applications. The actuation of the lens only requires in-plane movements making it an interesting candidate where the depth of the device is critical (such as mobile phones for example). Moreover its simple fabrication relies on the encapsulation of a high refractive index liquid in a non-miscible thermo-curable polymer. Since this bioinspired lens offers the possibility to encapsulate aqueous solutions, it paves the way for the development of truly accommodative artificial intra-ocular lenses. Today, more than half the population over the age of 65 has cataracts, a condition that causes a cloudiness of the crystalline lens of the eye. Current treatment replaces patients’ crystalline lenses with fixed-focal-length lenses called intra-ocular lenses. However most intra-ocular lenses, that are also used for patients with presbyopia, are not capable of accommodation. Using a flexible lens that closely mimic the vertebrate lens could restore full vision to people affected by these conditions.

In section 2 we describe the vertebrate lens and we give an overview of other bioinspired lenses. The design and fabrication of the lens are presented in section 3. Section 4 details the results and discussions.

2. Vertebrate lens and bioinspired equivalent

2.1. Vertebrate lens

The natural vertebrate lens is a transparent, biconvex structure in the eye that serves to refract light to be focused on the retina. It comprises the lens capsule, a smooth, transparent membrane that encloses a liquid, which contains water-soluble proteins. Accommodation is the process by which the lens changes its optical power so as to maintain a focused image of an object on the retina as the distance from the object varies. This is achieved through altering the lens's shape and thus changing its surface curvature and refractive power [1]. The Helmholtz mechanism is the most widely accepted theory for lens accommodation [2]. For distant vision (Figure 1a), light rays entering the eye are nearly parallel so little diffraction is needed to bring an image into focus. According to Helmholtz's theory, the ciliary muscles surrounding the lens relax, causing the suspensory ligaments holding the lens to become taught as they resist the force exerted by the internal pressure of the liquid inside the eye. The lens stretches and becomes thinner and flatter, thus reducing its power. For close vision, light rays entering the eye are divergent and so greater refraction is required to bring an image into focus. The ciliary muscles contract, this resists the internal pressure on the lens and allows the suspensory ligaments to slacken. The lens springs back into a thicker and more rounded form, thus increasing its power. Figure 1b shows actuation principle
of the lens described in this manuscript and Figure 1c shows an image of the lens.

2.2. Bioinspired lenses

Much recent work on optical devices has focused on the development of optofluidic lenses: those using liquids as their optical medium. This is evidenced by the large number of recent publications on the subject. Although most applications of this technology are the subject of ongoing research, cases of successful commercialisation are beginning to emerge [3]. Optofluidic lenses may be broadly categorised as those tuned by altering the shape of the lens and those tuned by altering the properties of the liquid. Three of the most promising classes of optofluidic lens are described below.

The basis of hydraulically tuned lens is a circular, liquid filled chamber, covered by a flexible membrane. As pressure is applied to the liquid, the membrane deforms, altering its radius of curvature and consequently the lens’s focal length. Much research into lenses of this type has been published: Ahn and Kim [4] presented a lens, tuned by the deformation of a 40µm thick glass diaphragm, by pressurised oil from an external micropump. The high stiffness of the lens material limits the achievable surface
Tuneable bioinspired lens

curvature, leading to a relatively long focal length of 300-600 mm. An improved lens was reported by Chronis et al. [5], where the use of a flexible silicone rubber membrane permitted greater surface curvature giving a focal length of 1-6 mm. An even shorter focal length of 0.6-3 mm was achieved by Jeong et al. [6], whose biconvex lens used two liquid filled chambers to vary surface curvatures independently. The above designs necessitate an external pressure source and liquid reservoir as well as interconnecting tubing. More compact lenses are achievable if the liquid is contained within the lens chamber and pressurised indirectly. Two such designs are proposed by Ren et al. [7, 8], in which pressure is applied via a servo driven squeezing string and a mechanical lever. The SOLID technology [9] that enables the encapsulation of a liquid under a Parylene film, a transparent polymer typically deposited under vacuum, has also been used to make tuneable lenses [10]. However the rigidity of Parylene does not allow for large variation of optical power without permanent deformation of the encapsulating membrane. The solution proposed in this manuscript, using an elastomer as the encapsulating material addresses this issue.

The contact angle between a solid and a conductive liquid may be changed by an applied potential difference across the interface [11]. By this process the shape of a drop of liquid may be altered, this is the principle governing the tuning mechanism of electrowetting lenses. One possible lens configuration involves the encapsulation of two non-miscible liquids of different refractive index in a closed cell. One liquid is conductive (usually an aqueous solution) whilst the other is non-conductive (usually oil based). The liquid-liquid interface adopts a spherical profile whose radius of curvature, and hence focal length, may be varied by altering the relative wetting angle of the conductive liquid on the cell walls via an applied voltage [12]. The first proof of concept of such a lens was reported by Gorman et al. [13], using a droplet of hexadecanethiol deposited on a gold electrode and immersed in electrolyte. An improved design, proposed by Berge and Peseux [14], centred and constrained the liquid droplet on the optical axis. Increasingly sophisticated designs have been presented by Kuiper and Hendriks [15], and Krogmann et al [16].

An optofluidic lens may be tuned by varying the optical properties of its liquid rather than by altering its geometry. Refractive index may be changed by subjecting the liquid to an external electrical field, acoustic field, temperature field, or mechanical strain. The main drawback of these methods is that the achievable change of refractive index is relatively small, limiting the degree of tuneability [17]. Microfluidic techniques, such as mixing and pumping, may be used to change the lens liquid or its composition. Although large variations in refractive index are possible, the response time is usually slow (on the order of seconds) [18]. Most electro-active lenses use liquid crystals, whose refractive index can be modified by the application of an external electric field, to provide tunability. The change in refractive index is almost instantaneous, giving fast response times [19].
3. Fabrication, actuation and characterisation of the lens

3.1. Fabrication of the lens

The basic principle behind the design of the lens is to encapsulate a shaped drop of a high refractive index liquid within a flexible and optically clear, solid polymer matrix. The refractive index of the liquid is different to that of the polymer such that refraction occurs at the solid-liquid interface. The actuation principle relies on radial stretching of the lens that changes the shape and hence the surface curvature of the drop. This, in turn, affects the degree of refraction which occurs at the surface interface, hence changing the lens's focal length.

Poly(dimethylsiloxane) (PDMS) (Sylgard 184, Dow Corning), a silicone based elastomer, is the material of choice for the lens body. This material is attractive because of its flexibility, optical clarity and ease of use. To manufacture a PDMS component, liquid elastomer is combined with cross-linker in a 10:1 ratio by weight and mixed thoroughly. The liquid is held under a vacuum to de-gas and remove air bubbles introduced during mixing. The liquid PDMS is poured into a mould and thermally cured at 65 °C for three hours. There is no shrinkage associated with the liquid-solid phase transition of PDMS so that the component will retain its shape. Provided the mould is smooth, no release agent is required to remove the solid component. New PDMS layers adhere well to cured surfaces leaving a clear interface; this enables the lens to be built up in multiple stages. Liquid Poly(ethylene glycol) (PEG400) (Sigma Aldrich), is used to form the liquid droplet inside the lens. PEG400 has a higher density than
Tuneable bioinspired lens

<table>
<thead>
<tr>
<th></th>
<th>refractive index</th>
<th>Young’s modulus (kPa)</th>
<th>density (kg/m³)</th>
</tr>
</thead>
<tbody>
<tr>
<td>PEG 400</td>
<td>1.466</td>
<td>NA</td>
<td>1.128</td>
</tr>
<tr>
<td>PDMS</td>
<td>1.40</td>
<td>390-870</td>
<td>0.97</td>
</tr>
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Table 1. Materials properties, taken from the suppliers.

uncured PDMS and the liquids are non-miscible. As a consequence, a drop of PEG400 will sit stably under a layer of liquid PDMS and will assume a spherical shape so as to minimise its surface energy. PEG400 was chosen because, of the available liquids which were both non-miscible and denser than PDMS, it had the highest refractive index. Relevant properties of both materials are given in Table 1.

A plano-convex lens can be fabricated by depositing a droplet of PEG400 onto a flat membrane of cured PDMS. The drop is then encapsulated by a further layer of liquid PDMS and cured. Figures S1 and S2, in the supplemental material show a picture of the lens and the fabrication process respectively. The optical power of such lens is limited and therefore a second fabrication method was developed to fabricate a bi-convex lens. The fabrication steps detailed in Figure 2 are described briefly below. A mould (in our case, a steel ball bearing) is held above the base of a Petri dish via a magnetic stand. PDMS is poured into the Petri dish to just above the level of the bottom of the bearing. Once the PDMS is cured, the bearing is removed to leave a perfectly spherical indentation. This indentation is then filled with liquid and encapsulated using liquid PDMS, which is then cured. This method produces a drop with a biconvex shape, refraction occurs at both the front and back liquid-solid interfaces such that more powerful lenses are achievable. The curvature of the rear surface is determined by the choice of ball bearing as so is highly controlled and repeatable. The curvature of the front surface is determined by the volume of liquid deposited. Provided the radius of the drop is below the capillary length ($\kappa \sim 2$ mm for PEG), gravity is insignificant compared to surface forces such that the top surface of droplet will adopt a perfectly spherical shape. Another lens with cap shaped membranes was fabricated (see Figures S3, in the supplemental material) however due to the fabrication process chosen (see Figure S4, in the supplemental material), the membranes were so thin that the lens shape suffered from gravitational distortion when held vertically. Improvement of the process using appropriate moulds will enable the fabrication of a lens with cap shaped membranes truly mimicking the shape of the vertebrate lens.

3.2. Actuation of the lens

The optical power of the lens is defined by its focal length: the distance behind the lens at which initially parallel rays converge. This distance is determined by the amount of refraction that occurs at each material interface. The focus of our biomimetic lens is tuned by radial stretching of the polymer disk, which alters the shape, and hence
Tuneable bioinspired lens curvature, of the lens. This, in turn, affects the amount of refraction at each surface and the lens's overall focal length.

A rig was built to radially stretch the lens (see Figure 3b). The lens is held in place with equispaced, sprung steel clips. M2.5 bolts passed through clearance fit holes in an outer aluminium ring and screwed into threaded inserts affixed to the rear of the clips. Tightening the bolts pull the clips outwards, thereby stretching the lens. Care was taken to mount the lens centrally with no play in the bolts whilst ensuring that no pre-strain was applied to the lens. Measurement of the screw thread pitch related bolt rotation to radial displacement. The holes in the aluminium ring were configured to enable stretching to be applied in between one and four directions. For the majority of testing, 6 clips fitted around the lens circumference, corresponding to three axis stretching as shown in Figure 3b. The design of the stretching ring is unable to provide truly uniform radial strain, however since the the distance at which the displacement is imposed is large compared to the diameter of the liquid lens (drop) the stress profile in the material surrounding the drop should be uniform and thus the radial strain on the drop can be considered axisymmetric. Stretching of the lens is measured as a dimensionless value of strain, defined by the radial displacement of the mounting clips divided by the initial distance from the centre of the droplet to the mounting point. For the majority of lens testing, the bolts were tightened in 30° increments up to 180°. This corresponded to a maximum strain of 0.67 %. The rotation of the bolts was accurate to within ±2°, this corresponds to a maximum error of ± 0.0075 % in the value of applied radial strain for any one particular axis.

3.3. Characterisation of the lens

The variations of the lens's focal length were measured using two methods. In the first, the position of the lens's point of best focus is determined directly by observation of the output of a CCD light sensor onto which an image is projected. In the second, the focal length is determined indirectly, by measurement of the lens's magnification of an image. The first method is considered a direct measurement method, whereas the second one is indirect. A ray-tracing program coded in Python [20] was used to correlate the focal length changes with the change in radii of curvature at the air-PDMS and PDMS/PEG400 interfaces.

To determine the point of best focus (direct measurement) the lens is placed between a focus target and a light sensor. The location of the target and lens are fixed whilst the sensor can be moved forwards and backwards relative to the the lens so as to be at the position of the point of best focus. The focus target is positioned at a fixed distance (1 - 1.5 m) in front of the lens. The light sensor and associated circuitry are taken from a commercial webcam (Logitech HD C310) with the housing and lens removed. The sensor is mounted behind the lens via a steel bracket whose position can be adjusted by a screw gauged micrometer. Telescoping brass tubing between the rear face-plate of the lens housing and the sensor housing (see Figure S5) prevents light leakage. A 3 mm
circular aperture, machined in the centre of the front face plater, is sufficiently large that the hole does not contribute to the focal length (pinhole camera effect) but is small enough to prevent excess light entering the system. Each element of the apparatus is mounted on an optical test rack, enabling their height and lateral position to be adjusted independently, to facilitate co-axial alignment. The light sensor output is monitored whilst the sensor position is adjusted via the screw gauge micrometer to the point of best focus. The gauge reading for this position is recorded; the micrometer gives readings to an accuracy of $\pm 0.01$ mm. Due to the subjective nature of the determination of the point of best focus, five different reading are taken, with the sensor being moved out of focus each time before being refocused. An example of images recorded is shown in Figure 4. This procedure is repeated after each incremental increase of lens radial strain. Once the travel limit of the micrometer is reached, the entire unit is moved back along the optical rack and the datum position adjusted accordingly. Additional sections are added to the telescoping tubing as necessary.

In a second method, the variation of the focal length is measured indirectly by examining the variation in magnification of an image of concentric circles. Apparatus is arranged, as shown in Figure 3c, such that light from the focus target forms an image behind the lens. This image is then photographed by a digital SLR camera. A photograph is taken after each incremental increase in strain. The diameter of each concentric feature is measured by comparison to a scale bar, photographed at the same distance at which the image is formed. The ratio of the image height $h_i$ to the object
Figure 4. Focal length measurement using the direct and indirect methods. top) screen shot of light sensor output for best focus position at different applied strains (direct method), bottom) graph showing the focal length as a function of the applied strain measured using the direct (plain line) and indirect (dashed line) methods.

height $h_o$ gives the lens magnification as given by: $M = h_i/h_o = f/(f - l_o)$ (see Figure 3a). Once the magnification is known, the focal length can be calculated according to:

$$f = \frac{Mh_o}{M - 1}$$

(1)

4. Results and discussion

4.1. Optical characterisation

Figure 4 shows the measured focal length of the lens as a function of strain using the two methods described above. Both curves follow the same trend. The values of the focal length measured using the indirect method present large variations for each set of measurement. The main source of error comes from the subjective nature of the measurement based on observation of the magnified images. The errors get amplified at high values of applied radial strain because the images become blurred, making the features harder to measure. This is due to spherical aberration as discussed later. Another source of error comes from the current actuation mechanism that is unable to
apply uniform radial strain. It can be observed in Figure 4 also that the lens recovers its initial state after being actuated.

The predicted focal length of the unstrained lens is $\sim 31$ mm (using the ray tracing code), which is lower compared to the values measured using either method. In addition, considering an incompressible liquid and hence a drop of constant volume, the maximum focal length variation should be less than 5% as calculated using the ray tracing code. Therefore another mechanism accounts for the large focal length range covered by the lens presented.

To accurately measure the variation in the shape of the drop during stretching, a series of side-on photographs of the lens (Figure 5) were taken at different values of applied radial strain. The software ImageJ was used to measure the radius of curvature of the drop by fitting circles of appropriate size to the lens profile. In the unstrained state, the radii are 5 mm (as defined by the radius of the ball bearing) and 2.77 mm (plain arc, Figure 5a). The radii of curvature increase to $\sim 5.1$ mm and 2.99 mm as defined by the plain arc in Figure 5b after applying a radial strain of 3%. However upon closer observation, it appears that the radius of curvature at the centre of the drop goes up to 4.32 mm (dashed arc, Figure 5b). This is due to a collapse of the PDMS membrane as explained below.

The surface profile of the lens was measured at different values of applied radial strain using a Taylor Hobson Form Talysurf surface profiler fitted with a standard gauge arm and 0.5 mm ball stylus. Profiles were measured over a 10 mm distance, centered over a diameter of the internal liquid droplet. After each run the bolts were incrementally tightened by 30° to increase the applied radial strain. Measurements were taken of both the front and rear surface for seven different strain values. Examples of three of the profiles measured are presented in Figure 5c. There is a noticeable depression in the surface of the lens centered above the liquid droplet. The diameter of the depressed surface region is 7 mm whilst the diameter of the internal liquid droplet is 4.9 mm. The depth of the depression increases as the lens is stretched. This observation is explained by the incompressible nature of the liquid droplet whose volume is conserved during stretching of the lens. As the droplet is radially strained it becomes flatter and thinner, its front and back surfaces are pulled together and this pulls in the surrounding rubber, creating the depressed surface.

However, there is also an initial 0.007 mm central depression in the lens surface before any radial stretching is applied. The cause of this is thermal expansion of the liquid droplet during the fabrication of the lens. The PDMS rubber is cured at a temperature of approximately 65°C. The liquid droplet expands at this temperature and the PDMS assumes the shape of the heated droplet before curing. When the lens is removed from the oven the liquid droplet cools and shrinks, thereby pulling in the surrounding PDMS and creating the surface depression (see Figure 5c). The volumetric change of the liquid droplet associated with cooling from 65°C to room temperature is consistent with the volume of the initial surface depression.

As a conclusion, the variation of the lens's focal length is governed by the changing
refractive contributions of each of its surfaces. The biconvex liquid droplet converges light. Its curvature is high, but refraction is limited by the low refractive index of the polymer-liquid interface. During stretching, the change in the droplet's shape is not enough to significantly alter its surface curvature such that its contribution to the lens's refractive power is small. As the droplet is stretched, it becomes thinner, pulling in the surrounding polymer and creating depressed concave regions on the lens's outer surfaces. These partially diverge the light being focused by the droplet, thereby lengthening the lens's focal length. Although the curvature of these depressions is small, the refractive index of the air-polymer interface is large, such that their contribution is significant. During radial stretching, the depth and curvature of these depressions increase, their divergent contribution grows and the lens's focal length increases. These observations account for the larger focal length range measured as verified using the ray tracing code.

Images formed by the lens are spherically aberrated to a significant extent. The curvature, and hence, divergent contribution, of the outer polymer surfaces is not uniform along the radius of the lens and causes light to focus at different points behind
the lens. A best focus exists at the point at which the diameter of the envelope of rays of light is minimised. This is known as the circle of least confusion, which was chosen for the measurement presented in this manuscript. As the lens is strained, the difference in focal length between the lens centre and lens edge increases, such that the magnitude of the spherical aberration and the diameter of the circle of least confusion grow. This was observed in the degradation of the quality of the images recorded from the light sensor output; as the lens radial strain was increased the images became more blurred. The decrease of image quality made the subjective determination of a distinct point of best focus harder. This is reflected in the wider range of measured focal lengths. The development of a lens with an aspherical surface will reduce the spherical aberration. In our case, it could be implemented by using a different mould (compared to the spherical bearing used for the fabrication of the lens presented in the manuscript). In addition, the actuation mechanism that relies on manually adjusting screws can lead to a non-uniformity of the lens that can also account for the spherical aberration observed. Therefore an actuation mechanism capable of applying uniform radial strain will also improve the performance of the lens.

4.2. Surface roughness - AFM

The polymerisation of PDMS on PEG400 raises question about the roughness of the interface. AFM (atomic force microscopy) measurements, using an Agilent 5500 AFM (Pico plus) in tapping mode and ultra-sharp cantilevers (NSC36/no Al) from MicroMasch with a resonance frequency of 170 kHz, were carried out to evaluate the RMS (root mean square) roughness of the interface. Measurements on areas of $8 \times 8 \mu m$ revealed RMS roughness of the order of $2.3 \text{ nm} \pm 0.3 \text{ nm}$ (see Figure 6). This low roughness, ideal for optical application as it keeps light scattering to a minimum, is due to the fact that liquid PDMS and PEG400 are not miscible. Therefore, due to the effect of surface tension, they adopt a shape which minimises their surface area to volume ratio. As a result, the PDMS surface cast against the liquid drop is extremely smooth.

4.3. Comparison to vertebrate eye

Much like the lens described in this paper, the focusing of light by the vertebrate eye is accomplished in two stages. The outer cornea has a fixed focal length, contributing approximately 70% of the refractive power of the eye [1]. Because the cornea interfaces with air, the refractive index of the surface, and hence, its optical power, is large in comparison to the contribution of the inner crystalline lens which is suspended in water like vitreous humour. By a mechanism of shape change, the contribution of refractive power of the inner lens is altered such that focus is achieved. For an average adolescent human, the optical power of the relaxed eye is around 60 dioptres with typical amplitude of accommodation of 10 dioptres [21]. This corresponds to focal length of 16-20 mm. By comparison, for the lens described in this manuscript nearly 100% of the convergent
power is provided by the inner lens like droplet, whilst tunability is achieved by altering the divergence of the outer, cornea like, PDMS surface. The advantage of this approach is that because the variable contribution to the lens's optical power is provided by the high refractive index interface, a much wider range of focal lengths is achieved with a smaller geometric variation (a range of 50 mm was measured compared to a range of 4 mm for the human eye). In this respect, the lens is similar to that found in the eye of the Great Cormorant, which, unlike the human eye, is able to change the curvature of its cornea to provide the wide range of refractive power necessary for clear vision both above and below water [22]. The disadvantage of this approach is that the fixed contribution to the lens's optical power is provided by the low refractive index solid-liquid interface. A much lower maximum lens power, corresponding to a longer minimum focal length is achievable (a minimum focal length of 50 mm was measured compared to a minimum focal length of 16 mm for the human eye). This limits the viewing angle of the lens.

4.4. Comparison to other bioinspired lenses

The lens presented in this manuscript compares favourably to other bioinspired lenses. For example lenses under thin deforming membranes have achieved focal lengths of 300-600 mm [5] using a 40 μm thick glass membrane or focal lengths of 1-6 mm [5], 0.6-3 mm [6] and 4-10 mm [23] for example using silicone rubber membranes. A bioinspired lens actuated using electroactive elastomers has achieved focal length between 17 and 23 mm [24]. For comparison with other type of bioinspired lenses, the reader is referred to a review on micro-optofluidic lenses [18]. Considering the low strain applied (a strain
Tuneable bioinspired lens

of 0.67 % enables to change the focus from 50 mm to 100 mm) the lens presented in this manuscript presents remarkable performance compared to other bioinspired lenses.

5. Conclusions

The design of a novel biomimetic lens, inspired by the mechanism of accommodation of the vertebrate eye, has been presented. The lens comprises a shaped drop of a high refractive index liquid encapsulated within a flat disk of optically clear flexible polymer. The lens is simple to fabricate and the polymer-liquid interface is inherently smooth to the nanometer level. This ensures there is little scattering of light and high optical transmissions can be achieved. The present technique of tuning via radial stretching requires manual adjustment. An alternative means of varying the droplet’s shape would be by changing the volume of liquid contained, possibly via a microfluidic pump. This would not only allow miniaturisation and automation but would enable the formation of a convex outer surface profile, thus increasing both the lens’s maximum refractive power and focal range. Currently, the performance of the lens is limited by its degree of spherical aberration, caused by the non-uniform curvature of its outer surfaces. This has been shown to become more problematic as the lens is strained and its focal length increases. This may be partially resolved by the inclusion of aspheric surfaces which could be achieved by using appropriate moulds and by using an actuation mechanism applying uniform radial strain. It should be noted also that the use of an automatic actuation mechanism will enable the evaluation of the actuation and relaxation time. The focal length of the lens has been found to be 50 mm in its relaxed state, rising to 100 mm at a maximum applied radial strain of 0.67 %. This range compares favourably to both biological lenses and other published examples of biomimetic lenses. Combined with its simplicity, mechanical stability, and scope for miniaturisation, the proposed system presents clear advantages over focus mechanisms reliant on lens displacement. Once current limitations have been addressed, the lens will present an attractive and viable alternative to conventional optical systems.

6. Acknowledgements

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