Changes in temperature inside an optomechanical model of the human eye during emulated transscleral cyclophotocoagulation

SIMON REGAL,1 JOE TROUGHTON,1 ROGER DELATTRE,1 THIERRY DJENIZIAN,1,2 AND MARC RAMUZ1,*

1Mines Saint-Etienne, Center of Microelectronics in Provence, Department of Flexible Electronics, F-13541 Gardanne, France
2Al-Farabi Kazakh National University, Center of Physical-Chemical Methods of Research and Analysis, Almaty, Tole bi str., 96A, Kazakhstan
*ramuz@emse.fr

Abstract: Currently, many diseases of the eye are treated by laser surgery. An understanding of light propagation and the heating of eye tissue during laser exposure is crucial to improving the outcome of these procedures. Here, we present the development of physical and computational models of the human eye by combining optical light propagation and thermal characteristics. For the physical model, all parts of the eye, including cornea, lens, ciliary body, sclera, aqueous and vitreous humors, and iris, were fabricated using a 3D printed holder and modified polydimethylsiloxane. We also present a computational model based on finite element analysis that allows for a direct comparison between the simulation and experimental measurements. These models provide an opportunity to directly assess the rise in temperature in all eye tissues. The simulated and physical models showed good agreement for the transmission of light at varying incident angles. The heating of optical components was investigated in the retina and the ciliary body during simulated laser surgery. Temperature increases of 45.3°C and 30.6°C in the retina and ciliary bodies, respectively, were found in the physical model after 1 minute of exposure to 186 mW of 850 nm laser radiation. This compared to 29.8°C and 33.9°C increases seen under the same conditions in the simulation model with human eye parameters and 48.1°C and 28.7°C for physical model parameters. These results and these models are very promising for further investigation of the impact of laser surgery.

© 2020 Optical Society of America under the terms of the OSA Open Access Publishing Agreement

1. Introduction

Ophthalmic laser surgery uses optical and thermal processes to improve patient vision. The control of these processes is essential to ensure only the desired location is treated, without affecting surrounding tissue.

Significant work already exists in which analytical models of the eye are presented. Models of laser light propagation [1,2] and thermal models of the eye, based on finite element analysis, [3–11] have been demonstrated. In addition, there is a wide array of work presenting physical models of human eyes. These models generally include the cornea, lens, aqueous and vitreous humors, iris, and retina. Simple approaches have a fixed iris and lens [12–16], while more advanced models include a tuneable iris and/or lens [17–20]. These models tend to include a system to record the image formed on the retina for comparison to real eyes. Some of this work has been developed to reproduce the human eye as closely as possible from an optical point of view [16–19], while others models are developed to investigate the impact of the type and placement of the intraocular lens for cataract surgery [12,14,15]. Studies have also been carried out on real human and animal eyes to investigate the temperature rise during illumination. These have investigated either the complete eye [21] or only the fundus [22], and have aimed
to set limitations for the time and light power to be used in order to minimize damage to the surrounding tissue or pain to the patient [23–25].

In the published work, many limitations are found. For the physical models discussed, only the normal path of light to form an image on the retina is generally considered, meaning interactions with other parts of the eye, such as the sclera and ciliary bodies, are neglected. Whereas the simulation model, based on our previous work, only takes into account the optical part and neglects the thermal part. There is also a lack of study of different wavelengths of light, particularly those relevant to laser eye surgery. The study of real eyes, particularly using cadaver tissue, for modeling laser treatments, is significantly limited by the fragility of human tissue. Such material quickly decays, with the onset of changes in optical properties occurring almost immediately after death.

The work here presents separately developed computational and physical models of the human eye. The computational model is developed using COMSOL Multiphysics® software, based on previously published work [26], and the physical model comprises components formed of a combination of 3D printed polylactic acid (PLA), injection molded polydimethylsiloxane (PDMS), and soft lithography. The models presented here go beyond previous work as they include the parts of the eye – cornea, lens, aqueous and vitreous humors, sclera, iris, and ciliary body – considered from both an optical and thermal point of view. The explicit inclusion of these components ensures interaction of light with the sclera and ciliary body is considered.

To the best of the authors’ knowledge, no previous work has combined both optical and thermal physical models to provide a system for accurately modelling the effects of laser surgery.

2. Material and methods

Figure 1(a) presents an expanded view of the physical model, extracted from the CAD software, for the main structure. This model is composed of a main body, ancillary optical components, and a tuneable microfluidic iris. Figure 1(b) shows a cross-section of the assembled design in the CAD software and Fig. 1(c) shows a fully fabricated and assembled device.

Fig. 1. Composition of the physical model of the human eye. a) Exploded view with all the parts of the model listed, presented from the CAD models; b) Cross-sectional view of the assembled model from the CAD designs; c) Fully assembled model; d) Pictures of the molds and individual parts fabricated by PDMS injection molding; e) Fabrication process to create the iris microfluidic iris. (A) Spin coating of SU8-2075 photoresin on a wafer, (B) UV exposure of photoresist, (C) Development of photoresist with propylene glycol monomethyl ether acetate, (D) Spin coating of PDMS after silanization of the mold surface, (E) PDMS peel off, (F) Bounding of the microfluidics channel on flat PDMS by oxygen plasma activation.
2.1. Main structure

The main structure is shown in yellow part in Fig. 1(a). The concave depression in the center of the design simulates the retina at the back of the eye. The design was made using the CAD software Inventor (Autodesk, Inc, USA) and was printed using an Ultimaker 3 Extended (Ultimaker B.V., Netherlands). The design was printed using PLA, which is a common material used for 3D printing, offering good surface quality and reliability. For this model, the vitreous chamber was filled with a Dulbecco’s Phosphate Buffered Saline (BSS) (Sigma-Aldrich, USA). When this model is assembled, the shape of the retina, along with the sclera, form a cavity, seen in Fig. 1(b), mimicking the vitreous chamber found in an eye.

2.2. Optical components

The optical components include all parts that transmit light in the eye. Model components were made by injection molding of PDMS. PDMS was used as it has a refractive index of 1.4275 ± 0.0005 at 850 nm, and is almost transparent (>90%) from 400 nm to 1400 nm Fig. 3(a). In addition, the PDMS transmission spectrum can be customized by mixing with dedicated dyes, Silc Pig® Pigments (Smooth-On, Inc, USA), allowing accurate modelling of other components. Figure 1(d) shows the molds and the phantom models obtained for each optical component. Each part was designed based on typical measurements of human eyes [27], and the PDMS injection molding process produced reproductions of these parts.

2.3. Tuneable iris

Based on the work of Kimmle et al. [28], the tunable iris was fabricated by soft lithography as described elsewhere [29]. Modulation of the iris aperture size is achieved by the injection of an infrared light absorbing dye (NIR880D, QCR Solutions Corp, USA) into a spiral shaped microfluidic channel, controlled using an microfluidic pump (11 Pico Plus Elite, Harvard Apparatus, USA). As the dye is injected, the channel is filled, from the outside inwards, effectively closing the iris. When the dye is removed, the iris opens up again, giving an aperture diameter range of 0 to 9 mm. See Visualization 1 for a video of this action. The fabrication process of this device is presented in Fig. 1(e). The fluidic channel and the wall widths were 250 µm and 50 µm respectively while the channel depth was 100 µm.

2.4. Light propagation and thermal measurement procedures

All measurements of optical and thermal properties of the physical model were performed using custom-made test bench with a 3D printed model stand mounted on a High-Precision Rotation Mount (Thorlab, United States). An 850 nm LED (VSLY5850, Vishay Semiconductors,USA) was mounted in the stand and directed into the iris for optical measurements, while an 850 nm continuous wave infrared laser diode (SPL TR85, OSRAM Opto Semiconductors, Germany) was used for thermal measurement. Figure 2(a) presents a cross-sectional view of the test bench. The light source was located 1 mm in front of the cornea. The light was emitted without any further focusing or collimating elements. The laser diode is specified by the manufacturers to have a spectral width of ± 3 nm around the emission center, with a beam divergence (FWHM) of 6 x 19° ($\theta_{\perp} \times \theta_{\parallel}$). For optical measurements, a VEMD5060X01 photodiode (Vishay Semiconductors, USA) was mounted at the back of the retina, and for thermal measurements a thermal camera (Flir E6, Flir, USA) was mounted on the test bench instead Fig. 2(a & d). Optical measurements were performed for angles between 0° and 14° with a step size of 2° with the photodiode located on the retina. Figure 2(a) and 2(b) show the rotation of the physical model from 0° (Fig. 2(a)) to 20° (Fig. 2(b)). Light source is fix and the physical model rotates with the photodiode at the back of the retina being the center of rotation. Figure 2(c) present the 3D view of the complete test bench for a 20° angle of rotation. Before placing in the model, the photodiode was placed directly
in front of the LED to measure the light incident on the model for transmission calculations. Thermal measurements were performed on the retina (when the model was at $0^\circ$ rotation) and on the back of ciliary body (when the model is at $20^\circ$ rotation). Figure 2(d) shows the location of the camera in relation to the physical model. Two measurements were made using the thermal camera. The first before the illumination, to establish a reference temperature, and the second after 1 minute of illumination. Both images were recorded with the laser diode turned off. Additionally, the laser power was varied between 2.7 mW and 186 mW. Measurements were later compared with results of the simulation model to assess the model’s accuracy and limitations, discussed below. All thermal measurements were conducted with air in the vitreous chamber, as it was not possible to include the vitreous humor because the main structure was removed for ciliary body measurements.

![Fig. 2.](image)

**Fig. 2.** Drawings of the test bench used for optical and thermal measurements, illustrating the rotation of the model used to vary the angle of the incident light. a) and b) Top cross-section view of the structure used for optical measurements at 0 degrees (a) and 20 degrees (b) The red arrow models the light beam. c) 3D representation of the system at 20 degrees rotation. d) Photograph of the setup used to perform thermal measurements. The back part of the model was removed during thermal measurements on the ciliary body.

### 2.5. Simulation

A computational model, developed using Comsol Multiphysics® finite element analysis software, was used as a comparison to experimental data. This model follows and improves upon previous work [26]. The optical and thermal parameters used are presented in Table 1. The thermal behavior of these components has been added to the model following data published elsewhere [11]. The simulation is based on the bio-heat transfer equations set out by Pennes, for the calculation of the heat transfer in biological tissues [30].
Table 1. Optical and thermal parameters used in the computational model, values taken from previously published works

| Eye component     | Refractive index | Absorption coefficient (mm⁻¹) | Scattering coefficient (mm⁻¹) | Thermal conductivity (W m⁻¹ K⁻¹) | Specific heat capacity (J kg⁻¹ K⁻¹) | Density (kg m⁻³) | Reference  |
|-------------------|------------------|-------------------------------|------------------------------|----------------------------------|-------------------------------------|------------------|------------|
| Cornea            | 1.38             | 0.014                         | 0                            | 0.58                             | 4178                                | 1050             | [26], [11] |
| Aqueous humor     | 1.341            | 0                             | 0                            | 0.578                            | 3997                                | 1050             | [26], [11] |
| Lens              | 1.458            | 0.003                         | 0                            | 0.4                              | 3000                                | 1000             | [26], [11] |
| Vitreous humor    | 1.341            | 0                             | 0                            | 0.594                            | 3997                                | 1000             | [26], [11] |
| Iris              | 1.387            | 2.234                         | 1.438                        | 1.68                             | 3650                                | 1100             | [26], [11] |
| Sclera            | 1.364            | 0.284                         | 2.558                        | 0.58                             | 4178                                | 1000             | [26], [11] |
| Retina            | 2.62             | 1.771                         | 0.565                        | 0.15                             | 3680                                | 1000             | [26], [11] |
| PDMS              | 1.4275           | Tuneable                      | Tuneable                     | 0.15                             | 1460                                | 970              | [31]       |
| PLA               | 0.11             | 1590                          | 1252                         |                                   |                                     |                  | [32]       |

In addition to these, boundary conditions are set for modelling the interaction between the model system and the external environment. These are:

- The sclera is heated by blood at the back of the eye. The heat transfer coefficient is 65 W m⁻² K⁻¹ [11] and the blood temperature is 37°C.
- In the cornea, there are convection and radiation losses. The heat transfer coefficient is 14 W m⁻² K⁻¹ [11] and the outside temperature is 25°C, with a corneal emissivity of 0.975 [11].
- Tear evaporation is set at 40 W m⁻² [11].

In order to make direct comparison between the physical and computational models two additional elements were included the simulation: an LED and a photodetector, matching those used in the physical model.

3. Results and discussion

The dimensions of each component of the physical model were taken from literature [27,33,34] and are summarized in the Table 2. The refractive index values of each component were taken from previous work [26]. These published works contain data from human and porcine subjects. The refractive index of the ciliary body is approximated by the choroids due to a lack of data in the literature. The equations used to calculate the refractive powers can be found elsewhere [35].

The refractive index for the cornea and the lens were taken as equal to that of PDMS while for aqueous and vitreous humor the BSS refractive index was used. Based on the dimensions presented in Table 2, refractive powers of 41.44D, 24.34D and 61.24D for the cornea, lens, and complete eye respectively were obtained. These values are of the order of magnitude of the values found in the human eye [36–40].

Figure 3(a) shows the transmission spectrums of pristine PDMS, PDMS with the addition of the dye, and porcine sclera obtained experimentally from our previous work [26]. It is clear that, while the pristine PDMS shows a very different transmission spectrum to the sclera over the range of interest, the addition of the dye brings the two spectra into good agreement, with the PDMS spectrum falling within the experimental error [the grey shaded region of Fig. 3(a)] of the porcine sclera. It should be noted that studies of the transmission spectrum of human sclera can be found elsewhere in literature [41]. The same general trend is seen in the transmission data, but
Table 2. Anatomic dimension used for the creation of the molds used to reproduce each optical component in PDMS. The values are taken from the literature [27,33,34].

| Sample         | Thickness (mm) | Radius (mm)       | Reference |
|----------------|----------------|-------------------|-----------|
| Cornea         | 0.5            | Anterior radius = 7.75 | [27]     |
|                |                | Posterior radius = 6.5 |          |
| Aqueous humor  | 3.3            |                   | [27]     |
| Lens           | 3.5            | Anterior radius = 10.2 | [27]     |
|                |                | Posterior radius = 6 |          |
| Iris           | 1              |                   | [27]     |
| Vitreous humor | 15.5           |                   | [27]     |
| Ciliary body   | 1.21           |                   | [33,34]  |
| Sclera         | 0.6            | 12                | [27]     |
| Retina         |                | 11.4              | [27]     |

the values are higher. This is likely due to the sclera being thinner in humans than pigs and so allowing higher transmission [42]. While the model presented here was developed using the porcine data, adapting this to match the human data could be achieved by simply adjusting the mixing ratio of dye and PDMS. Figure 3(e) shows the difference between the experimental data and the value obtained for the phantom model components at 850 nm for various components of the model. The difference between experimental data and the corresponding phantom ranges between 2% and 7%. For nearly all components, the value of the phantom lies within the experimental error of the data. These results show the excellent agreement between the model presented here and published porcine data [26]. Additionally, data for the transmission spectra of human cornea and lens show a transmission greater than 90% over the range of interest [43], matching well the transmission spectrum of pristine PDMS presented in Fig. 3(b) and (c).

Figure 3(f) shows the relationship between pupil diameter and the amount of liquid injected. Using this mechanism, the transmission of the iris in the model can be determined as a function of the opening (pupil) diameter.

Figure 3(g) shows transmission spectrum of the fully open and fully closed iris. As expected, when the pupil was open (when no dye had been injected) transmission was over 90%, corresponding to the transmission of pristine PDMS. As dye was injected, and the pupil closed, the transmission reduced to a minimum of 17% between 700 nm and 1100 nm, when the pupil was fully closed, corresponding to the total area of the microfluidic channel walls which do not block the light [44]. The transmission can be further reduced by reducing the width of the fluidic channel walls.

Figure 4(a) shows the transmission of 850 nm light from an LED recorded on the fundus of the eye model by a photodiode, as a function of the current applied to the LED. The current variation causes a variation in light power, but no corresponding change in the transmission ratio over the range of 10 to 100 mA. This was consistent with the same data extracted from the simulation model, although there was an average difference of 6.6 ± 0.3% between the experimental and simulation data. This difference is likely due to non-ideal photodiode behavior in the physical model. In the literature, direct transmission values of 83.5% are reported [43]. The low transmission maximums (around 50% for the simulation and 45% for physical model) are due to the limited collection area of the photodiode, and the angular spread of the laser beam. While this means total transmission in the models is not recorded, any change in transmission is still well characterized.

Figure 4(b) presents the light received at the fundus of the eye as a function of the incidence angle of the LED. Here, the same trend is seen for the simulated and experimentally observed
Fig. 3. Optical validation of physical model a) Transmission spectrum from 400 to 1400 nm for pristine PDMS (blue), PDMS with the Silc Pig® dye (red), and experimental data recorded for porcine sclera (black) [26]. The shaded grey region shows the experimental error for the porcine sclera. b), c) and d) show similar measurements for cornea, lens and ciliary body, respectively. e) Comparison of transmission at 850 nm between experimental porcine data [26] and phantom model components made of PDMS for the different parts of the eye; f) Variation of pupil size with the fluid volume injected in the microfluidic channel. g) Transmission of light, as a function of the wavelength, in the ‘open’ (empty) and ‘closed’ (filled) iris.
Fig. 4. Optical and thermal comparisons between simulation and physics models. a) Measurement of the transmission recorded at the retina for variation of light intensity; b) Measurement of the light transmission at the retina for varying incident angle; c) Steady state temperature distribution along the pupillary axis from the computational model; d) Comparison of experimental and simulation data for the change in temperature after one minute of light exposure at different LED powers recorded at the retina. Simulation data shown for models using the thermal parameters of either human eye tissue, or the materials used in the physical model; e) Temperature distribution inside the eye when light is incident at an angle of 20 degrees, extracted from the computational model. Light power set at 186 mW. Temperature elevation at the sclera and ciliary body is seen; f) Comparison of experimental and simulation data for the change in temperature after one minute of light exposure at different LED powers recorded at the ciliary body. g) Thermal image recorded with a Flir E6 thermal camera after 1 minute of illumination on the retina at 186 mW; h) Thermal image recording with a Flir E6 thermal camera after 1 minute of illumination on the ciliary body at 186 mW.
transmission, although a small offset remains. As the eye is rotated, there is a decrease in light transmission. This can be explained as the change in the incidence angle means that light no longer passes directly through the cornea, but also through the sclera and ciliary body. As these components have absorption coefficients much higher than those of the cornea and lens, more light is absorbed.

Figure 4(c) shows the simulation of the temperature distribution inside the human eye along the pupillary axis for the steady-state case in this model. An increase in temperature along the axis of symmetry of the eye is observed. Close to the cornea, the temperature is 34°C while at the retina it is 37°C. This is caused by heat losses to the local environment at the cornea, and heating from blood behind the retina [11]. This agrees with similar results reported in literature [4,7,8,11], supporting the accuracy of the simulated model in the unilluminated steady state.

Figure 4(d) presents the change in temperature, ΔT, measured on the retina experimentally and in simulation, as a function of the illumination power. A thermal image of the model, recorded with the camera after 1 minute of illumination of the retina at 186 mW, is presented in Fig. 4(h). This was calculated as the difference in temperature before and after one minute of LED illumination. The LED power was varied from 2.7 mW to 186 mW. This corresponds to 3.55 J mm⁻² which is comparable to the energy used during laser surgery, which range from 2.65 J mm⁻² to 4.64 J mm⁻² per burn. [45,46] It can be seen that the temperature change is directly proportional to the LED power. For this investigation, the computational model was first run with the thermal parameters summarized in Table 1, and then again using the thermal parameters of the materials used in the physical model. It can be seen that the simulated results show a markedly lower ΔT for the simulation using tissue parameters than the experimentally observed values. However, there is good agreement with the simulated ΔT obtained using the model material’s values. The difference between the experimental data and those obtained from the simulation with tissue parameters is likely due to the use of a retina phantom material that had a lower thermal coefficient than in a real eye. The slope of these data represents the rise in ΔT per milliwatt of power supplied to the LED, Δ²T. While there is a significant difference in Δ²T for the simulated (Δ²T = 0.160 °C/mW) and measured (Δ²T = 0.237 °C/mW) data both show similar linear behavior. In addition, the fact that the simulation closely matches the experimental data when the thermal coefficients of the physical model were used (Δ²T = 0.258 °C/mW) suggest that the simulation is suitable, but that a material with closer matched thermal coefficients should be sought for future work. There does also exist some published data considering thermal changes in the eye of rabbits [21] and pigs [22], as well as several simulation studies [5,9,11]. The results obtained for the model here are consistent with these previous works notwithstanding differences in experimental conditions (wavelength, beam diameter, etc.).

Figure 4(f) presents similar measurements performed on the back of the ciliary body. A thermal image of the model, recorded with the camera after 1 minute of illumination of the ciliary body at 186 mW, is presented in Fig. 4(g). Here it can be seen that the simulation and experimental results are very close for both the simulation using the tissue parameters and the one using model parameters. Linear regression fits here give Δ²T = 0.173 °C/mW, Δ²T = 0.182 °C/mW, and Δ²T = 0.154 °C/mW for the measured data, the tissue-based simulation, and the model-based simulations respectively. The close agreement between the models and the experimental data here suggest that the lack of BSS has little impact on the thermal properties of the model, and justifies its exclusion during these thermal measurements.

Figure 4(e) shows a 2D view of the temperature obtained during one simulation. The temperature is maximal at the interface between the ciliary body and sclera. The absorption of light is mainly at the ciliary body due to its high absorption coefficient, and this absorption occurs at the interface between the ciliary body and the sclera. The simulation presented in Fig. 4(e) confirms this. For comparison with the experimental data, the temperature at the back
of the ciliary body, indicated by the arrow in the figure, was used so as to match the experimental configuration.

The physical model is of particular importance as it provides a shelf-stable alternative to real eyes for the investigation of laser treatments. One issue of note is the significant variation in optical properties of eyes between individuals, and even between each eye in a single individual [23–25]. Recently, research groups have proposed algorithms and devices to provide feedback-controlled treatment during laser surgery of the retina [23–25]. The development of these methods is based on experiments carried out on humans [23,25] and animals [24]. It would be possible to develop phantom models, of the different model components, with different absorption or scattering properties to capture this variability. In the same way, the input optical parameters of the simulation model could be adjusted easily. From this point of view, both models could be interesting for evaluating the different algorithms proposed, by varying the parameters of absorption, transmission, and scattering, of all parts of the eye.

As well as showing good reliability under the test conditions presented, these models are highly adaptable as they are based on optical parameters valid for light with wavelength from 400 to 1400 nm. For the computational model, a change in wavelength simply needs some small changes in optical parameters which can be found in literature [26]. Currently the physical model is valid only over the range of 600 to 1000 nm due the gap between the experimental and the model transmissions becoming greater than an acceptable tolerance outside this range. However, the physical model can be trivially adapted by adjusting the concentration of the light absorbing dye added to the relevant components during fabrication.

In this study, energies comparable to those used in current laser surgery were investigated in order to reproduce the thermal elevation seen. However, further testing should be performed on these models. For example, the size of the beam and the pulse frequency used in transscleral cyclophotocoagulation should be examined in order to obtain significant data on thermal variation. Furthermore, the spatial distribution and the size of the spot recorded on the focus area should be investigated further in future work in order to evaluate the unwanted damage in non-target tissues.

4. Conclusion

This work presents, for the first time, the development of both physical and computational models of the human eye capable of simultaneously modelling optical and thermal properties of the eye. Together it is hoped these models will prove a valuable resource in the design of future ocular laser procedures, and provide a route to modeling intra- and inter-individual variation using interchangeable components.

Acknowledgements

The authors would like to thank the Comsol Multiphysics service for its help and advice in developing the simulation model.

Disclosures

The authors declare that they have no conflict of interest.

References

1. S. Norrby, “The Dubbelman eye model analysed by ray tracing through aspheric surfaces,” Opt. Phys. Optics 25(2), 153–161 (2005).
2. J. M. P. Coelho, J. Freitas, and C. A. Williamson, “Optical eye simulator for laser dazzle events,” Appl. Opt. 55(9), 2240 (2016).
3. J. A. Scott, “A finite element model of heat transport in the human eye,” Phys. Med. Biol. 33(2), 227–242 (1988).
4. J. A. Scott, “The computation of temperature rises in the human eye induced by infrared radiation,” Phys. Med. Biol. 33(2), 243–257 (1988).
5. E. H. Amara, “Numerical investigations on thermal effects of laser-ocular media interaction,” *Int. J. Heat Mass Transfer* **38**(13), 2479–2488 (1995).
6. K. J. Chua, J. C. Ho, S. K. Chou, and M. R. Islam, “On the study of the temperature distribution within a human eye subjected to a laser source,” *Int. Commun. Heat Mass Transfer* **32**(8), 1057–1065 (2005).
7. E. Y. K. Ng and E. H. Ooi, “FEM simulation of the eye structure with bioheat analysis,” *Comput. Meth. Prog. Bio.* **82**(3), 268–276 (2006).
8. E.-Y. K. Ng, E.-H. Ooi, and U. Rajendra Archarya, “A comparative study between the two-dimensional and three-dimensional human eye models,” *Math. Comput. Model.* **48**(5-6), 712–720 (2008).

9. M. Cvetkovic, D. Poljak, and A. Peratta, “Thermal modelling of the human eye exposed to laser radiation,” in *2008 16th International Conference on Software, Telecommunications and Computer Networks* (IEEE, 2008), pp. 16–20.
10. R. H. Forushani, K. Hassani, and F. Izadi, “Steady State Heat Analysis of the Eye Using Finite Element Method,” *Bioméd. Res.* **37**(1), 347–350 (2012).
11. S. A. Mirnezami, M. R. Jafarabadi, and M. Abrishami, “Temperature distribution simulation of the human eye exposed to laser radiation,” *J. Lasers Med. Sci.* **4**(4), 175–181 (2013).
12. A. Barcik, J. Nowak, D. Siedlecki, M. Zajac, and J. Zarowny, “Physical model of human eye with implantable intraocular lenses,” in (2008), p. 71411A.
13. R. C. Bakaraju, K. Ehrmann, D. Falk, A. Ho, and E. Papas, “Physical human model eye and methods of its use to analyse optical performance of soft contact lenses,” *Opt. Express* **18**(16), 16868–16882 (2010).
14. A. Drauschke, E. Rank, L. Traxler, and M. Forjan, “Mechanical eye model for comparison of optical and physiological imaging properties,” in *Proceedings of 15th International Conference MECHATRONIKA* (2012), pp. 1–6.
15. A. Arianpour, E. J. Tremblay, I. Stamenov, J. E. Ford, D. J. Schanzlin, and Y. Lo, “An optomechanical model eye for ophthalmological refractive studies,” *J. Refract. Surg.* **29**(2), 126–132 (2013).
16. P. Xie, Z. Hu, X. Zhang, X. Li, Z. Gao, D. Yuan, and Q. Liu, “Application of 3-Dimensional Printing Technology to Construct an Eye Model for Fundus Viewing Study,” *PLoS One* **9**(11), e109373 (2014).
17. D. Liang, K. Xiang, J.-W. Du, J.-N. Yang, and X.-Y. Wang, “Biomimetic optical system using polymer lenses with tunable focus,” *Opt. Eng.* **53**(10), 105101 (2014).
18. A. Santiago-Alvarado, A. Cruz-Félix, A. Hernández Méndez, Y. Pérez-Maldonado, and C. Domínguez-Osante, “Design and characterization of a tunable opto-mechatronic system to mimic the focusing and the regulation of illumination in the formation of images made by the human eye,” in T. George, A. K. Dutta, and M. S. Islam, eds. (2015), p. 94671Y.
19. S. Petsch, S. Schuhladen, L. Dreesen, and H. Zappe, “The engineered eyeball, a tunable imaging system using soft-matter micro-optics,” *Light: Sci. Appl.* **5**(7), e16068 (2016).
20. L. Jung, J. Xiao, V. Malarchuk, C. Lu, M. Li, Z. Liu, J. Yoon, Y. Huang, and J. A. Rogers, “Dynamically tunable hemispherical electronic eye camera system with adjustable zoom capability,” *Proc. Natl. Acad. Sci. U. S. A.* **108**(5), 1788–1793 (2011).
21. H. Sailer, K. Shinoda, G. Błatsios, K. Kohler, L. Bondzio, E. Zrenner, and F. Gekeler, “Investigation of thermal effects of infrared lasers on the rabbit retina: a study in the course of development of an active subretinal prosthesis,” *Graefe’s Arch. Clin. Exp. Ophthalmol.* **245**(8), 1169–1178 (2007).
22. N. Heussner, M. Vagos, M. S. Spitzer, and W. Stork, “A prediction model for ocular damage – Experimental validation,” *J. Therm. Biol.* **52**, 38–44 (2015).
23. R. Brinkmann, “Real-time temperature determination during retinal photocoagulation on patients,” *J. Biomed. Opt.* **17**(6), 061219 (2012).
24. D. Lavinsky, C. Sramek, J. Wang, P. Huie, R. Dalal, Y. Mandel, and D. Palanker, “Subvisible Retinal Laser Therapy: Titration Algorithm and Tissue Response,” *Retina* **34**(1), 87–97 (2014).
25. E. Seifert, J. Tode, A. Pielen, D. Theisen-Kunde, A. Framme, J. Roider, Y. Miura, R. Birngruber, and R. Brinkmann, “Selective retina therapy: toward an optically controlled automatic dosing,” *J. Biomed. Opt.* **23**(11), 1–12 (2018).
26. S. Regal, D. O’Connor, P. Brige, R. Delattre, T. Djenizian, and M. Ramuz, “Determination of Optical Parameters of the Porcine Eye and Development of a Simulated Model,” *J. Biophotonics* **12**(11), e201802766 (2019).
27. D. A. Atchison and G. Smith, *Optics of the Human Eye* (Butterworth-Heinemann, 2002).
28. C. Kimmel, U. Schmittat, C. Doering, and H. Fouckhardt, “Compact dynamic microfluidic iris for active optics,” *Microelectron. Eng.* **88**(8), 1772–1777 (2011).
29. M. Lake, M. Lake, C. Narciso, K. Cowdrick, T. Storey, S. Zhang, J. Zartman, and D. Hoelzl, “Microfluidic device design, fabrication, and testing protocols,” *Protocol Exchange* (2015).
30. H. H. Pennes, “Analysis of Tissue and Arterial Blood Temperatures in the Resting Human Forearm,” *J. Appl. Physiol.* **1**(2), 93–122 (1948).
31. “PDMS,” http://www.mit.edu/~6.777/medprops/pdms.htm.
32. S. Farah, D. G. Anderson, and R. Langer, “Physical and mechanical properties of PLA, and their functions in widespread applications — A comprehensive review,” *Adv. Drug Delivery Rev.* **107**, 367–392 (2016).
33. M. D. Bailey, L. T. Sinnott, and D. O. Mutti, “Ciliary Body Thickness and Refractive Error in Children,” *Invest. Ophthalmol. Visual Sci.* **49**(10), 4353–4360 (2008).
34. Z. Wang, C. Chung, J. Lin, J. Xu, and J. Huang, “Quantitative Measurements of the Ciliary Body in Eyes With Acute Primary-Angle Closure,” *Invest. Ophthalmol. Visual Sci.* **57**(7), 3299–3305 (2016).
35. M. Katz, *Introduction to Geometrical Optics* (World Scientific, 2002).
36. W. Lotmar, “Theoretical Eye Model with Aspherics,” J. Opt. Soc. Am. 61(11), 1522–1529 (1971).
37. R. Navarro, J. Santamaría, and J. Bescós, “Accommodation-dependent model of the human eye with aspherics,” J. Opt. Soc. Am. A 2(8), 1273–1280 (1985).
38. H.-L. Liou and N. A. Brennan, “Anatomically accurate, finite model eye for optical modeling,” J. Opt. Soc. Am. A 14(8), 1684–1695 (1997).
39. T. Olsen, “On the calculation of power from curvature of the cornea,” Br. J. Ophthalmol. 70(2), 152–154 (1986).
40. R. P. Hemenger, L. E. Garner, C. S. Ooi, and B. Optom, “Change with age of the refractive index gradient of the human ocular lens: Poster # 56 (OR-109),” Optom. Vis. Sci. 71(Supplement), 175 (1994).
41. A. N. Bashkatov, E. A. Genina, V. I. Kochubey, and V. V. Tuchin, “Optical properties of human sclera in spectral range 370–2500 nm,” Opt. Spectrosc. 109(2), 197–204 (2010).
42. I. Sanchez, R. Martin, F. Ussa, and I. Fernandez-Bueno, “The parameters of the porcine eyeball,” Graefe’s Arch. Clin. Exp. Ophthalmol. 249(4), 475–482 (2011).
43. E. A. Boettner and J. R. Wolter, “Transmission of the ocular media,” Invest. Ophthalmol. Visual Sci. 1(6), 776–783 (1962).
44. “NIR880D-880-nm-NIR-Dye-QCR Solutions Corp,” https://qcrsolutions.com/wp-content/uploads/2018/04/NIR880D-880-nm-NIR-Dye-Technical-Data-Sheet-QCR-Solutions-Corp.pdf.
45. M. C. D. Aquino, K. Barton, A. M. W. Tan, C. Sng, X. Li, S. C. Loon, and P. T. Chew, “Micropulse versus continuous wave transscleral diode cyclophotocoagulation in refractory glaucoma: a randomized exploratory study: Micropulse cyclophotocoagulation,” Clin. Experiment Ophthalmol. 43(1), 40–46 (2015).
46. S. A. Pastor, K. Singh, D. A. Lee, M. S. Juzycz, S. C. Lin, P. A. Netland, and N. T. A. Nguyen, “Cyclophotocoagulation: A report by the american academy of ophthalmology11Prepared by the Ophthalmic Technology Assessment Committee Glaucoma Panel and approved by the American Academy of Ophthalmology’s Board of Trustees August 1, 2001,” Ophthalmology 108(11), 2130–2138 (2001).