

Intraocular Lens Based Vision Simulator Using Computational Holographic Displays

by

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Intraocular Lens Based Vision Simulator Using Computational Holographic Displays

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Dedicated to my family.

ABSTRACT

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Cataract is defined as the cloudiness of the crystalline lens structure due to aging, trauma, etc. It is one of the most common preventable eye diseases in the world. The only treatment is surgery. During the surgery, the cataractous lens is removed and an artificial intraocular lens is implanted into the eye. There is an ongoing research effort to improve intraocular lens optical characteristics and depth-of-field to make patients totally spectacle-free after cataract surgery. However, modeling the vision through a cataractous lens and demonstrating various intraocular lens options before surgery are both significant challenges.

In the first part of this thesis research, we developed an eye model to assess the performance of different intraocular lens options. Furthermore, we used the results of the eye model in a custom holographic vision simulator developed in our group to predict the postoperative vision of patients before surgery. In the clinical assessments, we employed a computational holographic display as it is the only technology that can generate digitally programmable exit pupils and simultaneously display content at different depths. Our artificial eye model was used to evaluate the optical performance and the point spread functions of different intraocular lenses. Those PSFs were then used to compute phase holograms to display on our custom holographic vision simulator.

We characterized three different intraocular lenses using a standardized resolution target and point light source illumination using an light emitting diode. Monofocal, bifocal, and trifocal intraocular lenses exhibited a contrast decrease of 5%, 42%, and 45% and a resolution limit of 4.49, 4.00, and 4.00 lp/mm at 35cm (using 8% modulation transfer function criteria), respectively. Monofocal intraocular lenses had the best resolution and contrast. The multifocal lenses produced prominent photic phenomena such as halos and glare around bright light sources, reducing contrast and resolution. We compared our results with the literature and confirmed that the visual functions of intraocular lenses could be successfully modeled and simulated using a holographic display before surgery.

In the second part of this research, we used our eye model and developed a cataractous lens simulator for progressive levels of opacification, which is achieved by applying a reversible chemical procedure on the intraocular lens surfaces. After the lens is fully or partially immersed in acetone, subsequent testing of the lens in distilled water results in a progressive change in opacification level within minutes. We objectively measured the quality of vision by obtaining modulation transfer function curves, transmission, and spectroscopic measurements at different opacification levels. By simulating variable opacification across the IOLs, we tested how vision changes from less dense to more dense cataractous regions in a holographic display system with programmable small exit pupils. All results were consistent with the expected vision degradation caused by natural opacification.

ÖZETÇE

Holografik Ekran Kullanan İntraoküler Lens Temelli Görüntü Simülatörü

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Katarakt, yaşlanma, travma vb. nedenlerle kristalin lens yapısının bulanıklığının tanımlandığı bir durumdur. Dünyada en yaygın önlenabilir göz hastalıklarından biridir ve tek tedavisi cerrahi operasyondur. Cerrahi sırasında kataraktlı lens çıkarılır ve yapay bir göz içi lens göze yerleştirilir. Katarakt cerrahisi sonrası hastaları tamamen gözlüksüz yapabilmek için göz içi lenslerinin optik özelliklerini ve derinlik odaklarını geliştirmeye yönelik sürekli bir araştırma çabası vardır. Bununla birlikte, cerrahi öncesinde kataraktlı bir lensin görüşünü modellemek ve çeşitli göz içi lens seçeneklerini göstermek en önemli zorluklardır.

Bu tez çalışmasının ilk bölümünde, farklı göz içi lensi seçeneklerinin performansını değerlendirmek için bir göz modeli geliştirdik. Ayrıca, göz modelinin sonuçlarını, cerrahi öncesi hastaların gözündeki sonucu tahmin etmek için grubumuzda geliştirilen özel bir holografik görüş simülatöründe kullandık. Klinik değerlendirmelerde, sayısal olarak programlanabilir çıkış gözbebekleri üretebilen ve aynı anda farklı derinliklerde içerik gösterebilen tek teknoloji olan bir hesaplamalı holografik ekranı kullandık. Yapay göz modelimiz, farklı göz içi lenslerinin optik performansını ve nokta yayılım fonksiyonlarını değerlendirmek için kullanıldı. Bu nokta yayılım fonksiyonları daha sonra özel holografik görüş simülatörümüzde göstermek üzere faz hologramlarını hesaplamak için kullanıldı.

Standartlaştırılmış bir çözünürlük hedefi ve LED kullanarak nokta ışık kaynağı aydınlatması kullanarak üç farklı göz içi lensi karakterize ettik. Monofokal, bifokal ve trifokal göz içi lensler sırasıyla %5, %42 ve %45 kontrast azalması ve 8% modülasyon transfer fonksiyonu kriterlerine göre 35 cm'de sırasıyla 4,49, 4,00 ve 4,00 lp/mm çözünürlük sınırı gösterdi. Monofokal göz içi lensler en iyi çözünürlüğe ve kontrasta sahipti. Çok odaklı lensler, parlak ışık kaynaklarının etrafında hale ve parlamalar gibi belirgin fotik fenomenler üretti ve bu da kontrastı ve çözünürlüğü azalttı. Sonuçlarımızı literatürle karşılaştırdık ve göz içi lenslerin görüş fonksiyonlarının cerrahi öncesi bir holografik ekran kullanılarak başarılı bir şekilde model ve simüle edilebileceğini doğruladık.

Araştırmanın ikinci bölümünde, göz modelimizi kullanarak ve içinde ilerleyici opasifikasyon seviyeleri olan bir kataraktlı lens simülatörü geliştirdik, bu opasifikasyon, göz içi lens yüzeylerine geri dönüştürülebilir bir kimyasal işlem uygulayarak elde edildi. Lens tamamen veya kısmen asetonda daldırıldıktan sonra, lensin distile su içinde test edilmesi sırasında opasifikasyon seviyesinde dakikalar içinde ilerleyici bir değişiklik meydana gelir. Modülasyon transfer fonksiyonu eğrilerini, iletimi ve farklı opasifikasyon seviyelerinde spektroskopik ölçümleri alarak görüş kalitesini nesnel olarak ölçtük. IOL'ler üzerinde değişken opasifikasyon simüle ederek, programlanabilir küçük çıkış gözbebekleri olan bir holografik ekran sisteminde yoğunluğu azdan yoğunla doğru nasıl değiştiğini test ettik. Tüm sonuçlar, doğal opasifikasyonun neden olduğu beklenen görüş bozulması ile uyumluydu.

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ABBREVIATIONS

IOL	Intraocular Lens
SLM	Spatial Light Modulator
CPD	Cycles Per Degree
PSF	Point Spread Function
D	Dioptries
SD	Standard Deviation
IQR	Interquartile Range
CGH	Computer Generated Holography
PCO	Post Capsular Opacification
MTF	Modulation Transfer Function
CSF	Contrast Sensitivity Function
HNED	Holographic Near Eye Display
LASER	Light Amplification by Stimulated Emission of Radiation
LED	Light Emitting Diode
ISO	International Organization for Standardization

Chapter 1:

INTRODUCTION

At least 2.2 billion people globally suffer from disability related to vision, with 65.2 million suffering from moderate to severe distant vision impairment or blindness because of cataracts [World Report on Vision, 2021]. These statistics suggest that cataract is one of the leading causes of vision impairment. Cataract surgery consists of a single procedure under local anaesthetic that may be done as a day case. The surgery involves removing the opaque lens and implanting an IOL [Ramke et al., 2017]. It is highly cost-effective [Baltussen et al., 2004]. It considerably improves quality of life since early surgery can prevent visual damage from increasing or, if done later, can restore eyesight [Finger et al., 2017].

There are several options for the types of intraocular lenses to be implemented. A monofocal IOL with a single, fixed focal length has been a common choice for this implementation. Another option is multifocal IOLs, which are engineered to provide good unaided distant and close vision. They realize multiple foci with diffractive optics, zones of varying refractive power, or their combinations [Calladine et al., 2006]. If a multifocal IOL has two foci, it is called bifocal; if it has three, it is called trifocal.

Although they were engineered to have multiple foci for spectacle-free vision and have considerably good optical performance, in 7 trials, multifocal IOLs were related to undesirable visual phenomena in 97 of 334 instances (29%), compared to just 26 of 328 cases (8%) with monofocal IOLs [de Silva et al., 2016]. The positive projection of non-useful patterns distinguishes these visual phenomena, also called pseudophakic photic phenomena, on the retina. These photic phenomena are artifacts that appear as glares or halos under lighting conditions [Buckhurst et al., 2017]. These phenomena are linked to patient discontent [Woodward et al., 2009; Javitt et al., 2000] and are common causes of multifocal IOL explantation [Kamiya et al., 2014]. As a result, the connection of multifocal IOLs with photic phenomena remains one of the most common reasons for postoperative complaints [de Silva et al., 2016; Khandelwal et al., 2019].

Photic phenomena of multifocal IOLs, such as halos and glare, differ among patients [de Silva et al., 2016], with some experiencing pseudophakic photic phenomena [Tchah et al., 2017] months after surgery. These photic phenomena can be reported

subjectively using questionnaires or assessed scientifically using wavefront aberrometry analyses to distinguish the experiences between patients. The literature also mentions that contrast sensitivity may be a more objective/helpful method for assessing visual function [Henein et al., 2022]. Some researchers evaluated photic phenomena measurements in the laboratory and devised a quantitative metric to differentiate them objectively [Carson et al., 2018]. Others also assessed factors associated with glares and halos in patients [Espaillat et al., 2021].

A system that mimics the human's optical, physical, and sometimes physiological aspects is called an optomechanical eye model or an artificial eye model. Although fully mimicking the eye is still an ongoing research topic, there have been many attempts to build a physical eye model [Amorim et al., 2022]. These models are an option to simulate and evaluate photic phenomena objectively. The developed models were initially simplified based on the famous Gullstrand reduced eye model [Gullstrand et al., 1909]. However, more accurate and detailed models were developed with technological advancements. Eye models are commonly used as performance criterion for glasses, contact lenses, and intraocular lenses, as well as to test and fine-tune optical measurement systems, innovate surgical treatments [Amorim et al., 2022].

Throughout the years, there have been many artificial eye models developed by researchers to evaluate the optical properties of IOLs, including but not limited to aberrations, imaging quality, side effects, and contrast decrease [Amorim et al., 2022]. The optomechanical eye model used in this study was based on a small, packed device developed in a previous study [Gobbi et al., 2006]. The model was optimal for providing data on visual performance with IOLs to objectively compare IOL models and establish a direct association with the proportionate visual performance achieved.

Nevertheless, these eye models were typically used in in vitro trials and not directly in clinics to evaluate the subjective performance of patients, especially in cataracts. The main obstacle to using these models for cataract patients is the cloudiness of their crystalline lens, which only lets through the image partially, preventing the patients from seeing the whole image. So, showing the phenomena mentioned above to patients before surgery is still an essential challenge with state-of-the-art technology.

Holographic displays have the potential to provide unparalleled capabilities for direct view displays as well as virtual and augmented reality applications. Holography is the only technology that can create real-like virtual 3D images. This gives holography

the possibility of being used in ophthalmological studies. With CGH solutions, the diagnosis, treatment, and follow-up of several eye diseases can be made uniquely [Kavakli et al., 2021]. The use of CGH for cataract patients is still limited. So far, only Kavakli et al. have used augmented reality and CGH technology in their head-mounted display to screen cataract patients [Kavakli et al., 2021]. Another unique solution holographic displays provide for cataracts is their ability to transmit the desired image through a pinhole in a crystalline lens. This gives holography the power to fully show any image to patients with mild to severe cataracts if a non-cataractous region exists in their lens.

The study of visual outcomes in people with cataracts and post-capsular opacification is critical in ophthalmology. Cataracts, which are caused by the clouding of the eye's natural lens, are a prevalent cause of vision impairment and substantially influence the quality of life of those who suffer from them. As the world's population ages, the frequency of cataracts is predicted to grow, necessitating better knowledge of how cataract severity affects visual function. Furthermore, we may contribute to continuous advances in diagnostic tools, surgical procedures, and intraocular lens designs by better understanding the link between cataract severity and visual quality.

1.1 Contribution of The Thesis Research

In this thesis, we present two novelties. Firstly, we combine holographic display technology with our newly designed optomechanical eye model and propose a novel benchtop setup for cataract patients. This benchtop setup preoperatively estimates the photic phenomena and side effects patients might experience after IOL implantation, such as decreased contrast or resolution, through characterizations done with the optomechanical eye model. Secondly, we simulate vision under different conditions in progressively increasing cataract and post-capsular opacification patients by utilizing our optomechanical eye model and applying a reversible chemical procedure to intraocular lenses.

The following summarizes the key contributions of the study made in this thesis.

- *A novel method for simulating multifocal IOLs through holographic display.*
Despite their design to provide multiple foci for clear vision without glasses and their overall good optical performance, multifocal IOLs have been associated

with undesirable visual phenomena. There are several IOL simulators to show vision and these undesirable visual phenomena through IOLs to cataract patients prior to surgery. However, they do not work well for patients with moderate or dense cataractous lenses due to impaired vision and heavy scattering. Clinical evaluations are only available postoperatively, which is too late for patients and doctors to decide. As proposed in our group's previous study, we used an eye-box steering method with a pupil tracker to transmit the holographic images with the desired side effects to the patient's non-cataractous regions and use the holographic display to correct the refractive errors [Kavakli et al., 2021]. Unlike our group's previous study, which only focused on monofocal IOLs, we added the side effects of multifocal IOLs and photic phenomena that occurred due to their imperfections to our simulations. To our knowledge, we are the first to apply the characterizations made by an artificial eye model to holographic content and show the results to subjects regardless of the severity of their cataracts. Experiencing the vision of an IOL through a holographic simulator helps subjects understand the effects of their IOL choice and shows them what to expect after surgery. Our method will significantly contribute to ophthalmologic studies and is essential to increasing patient satisfaction after cataract surgery.

- *A novel procedure to mimic opacification on human lenses as a reversible and controlled process.* Objectively observing the degradation of vision caused by the opacification of the human lens, such as cataracts or post-capsular opacification, is a crucial need for ophthalmology. There have been previous studies where cataracts were simulated physically or virtually, utilizing methods such as virtual reality and optical setups. However, there has never been a case where this opacification was simulated as progress in virtual simulations, and the only physical example was an irreversible process that would not be repeated on the same substance. Different from other cataract simulators, for the first time in literature, we came up with a procedure to simulate the progression of cataracts in a repeatable manner. These simulations assess visual outcomes in individuals with cataracts and post-capsular opacification, employing an artificial eye model and a chemical process on IOLs. The influence of varying degrees of cataracts

on visual quality was objectively evaluated by acquiring modulation transfer function curves, photodiode measurements, and spectroscopy results. By simulating opacification in a structured fashion, we examined the variations in results within the patient's pupil, transitioning from regions with lower cataract density to those with higher density, using our holographic display as a testing platform. An integral aspect of our study pertains to its potential clinical implications. With cataracts and PCO typically occurring slowly and patients often becoming aware of their condition only in severe cases, our approach holds promise as an educational tool within the realm of ophthalmology. This method could facilitate early diagnoses by demonstrating the progressive impact of opacification on vision, allowing for timely interventions that prevent a significant decline in patients' quality of life.

- This project's efforts have resulted in two extensive journal articles, which are under review. The upcoming publication of these articles promises to add useful insights to the relevant ophthalmology domains.

Chapter 2:

BACKGROUND**2.1 Cataracts, Surgery, and Intraocular Lens Implantation****2.1.1 Cataracts**

Cataract is characterized as a condition in which the lens becomes cloudy, resulting in visual impairment. Although cataracts can be throughout the lens, the three predominant types are cortical, nuclear, and subcapsular cataracts (primarily posterior). Additionally, two distinct types, namely anterior polar and equatorial cataracts, are observed in experimental research settings as shown in Figure 2.1.

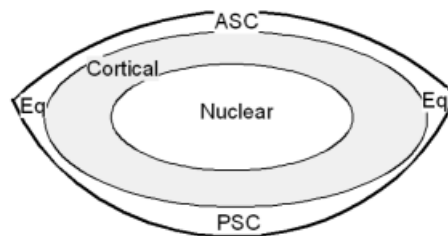


Figure 2.1 Potential cataract locations. ASC = anterior subcapsular cataract; Eq = equatorial cataract; PSC = posterior subcapsular cataract [Löfgren et al., 2017].

The precise underlying mechanism of cataract formation remains elusive. Numerous known factors have been identified, each contributing to a fraction of the overall incidence. However, it is widely acknowledged that advancing age plays a significant role in cataract development. Through the utilization of twin registries, it was learned that genetics plays a role in cataracts development [Hammond et al., 2001]. Furthermore, diabetes, trauma, smoking, exposure to ionizing radiation, and solar ultraviolet radiation have all exhibited significant associations with cataract development in crystalline lens [Seddon et al., 1995]. Regrettably, the pathophysiological understanding of most of these factors remains not enough.

2.1.2 Cataracts Diagnosis and Treatment

The diagnosis of cataract is typically established when patients report a decline in visual acuity or experience glare. The presence of opacities interferes with vision through light absorption and scattering.

Various techniques are available for the detection and quantification of cataracts. The slit-lamp microscope serves as the primary non-invasive tool for this purpose. In quantitative studies, Scheimpflug photography is preferable as it enables the examination of the entire lens with focus.

Throughout the past two millennia, surgical removal of the opaque lens tissue obstructing the visual pathway has been the sole method of treating cataract. Initially performed through a procedure known as couching, and presently carried out via phacoemulsification, which involves fragmenting the lens tissue using ultrasound and removing it from within the capsule.

The primary objective of cataract surgery is the implantation of an IOL into the human eye. This procedure is replacing the cloudy natural lens with an artificial one and typically a straightforward surgery lasting approximately 30 minutes. Cataract surgery is often performed as a procedure under local anesthesia, allowing patients to return home on the same day.

The surgical procedure concludes with the insertion of an IOL. In the 1940s, Dr. Harold Ridley of England invented the IOL, introducing plastic lenses to replace the bulky phakic glasses that patients had to rely on following cataract surgery [Ridley, 1952].

2.2 *Types of Intraocular Lenses*

IOLs play a pivotal role in the field of cataract surgery, serving as artificial replacements for the cloudy natural lenses that are removed during the procedure. An example of the procedure to implant IOL to eyes is shown in Figure 2.2. IOLs have significantly improved visual outcomes and quality of life for individuals with cataracts. With advancements in technology and design, various types of IOLs have emerged to address specific visual needs and preferences. This section will explore the different types of intraocular lenses, highlighting their characteristics and applications in the realm of ophthalmology.

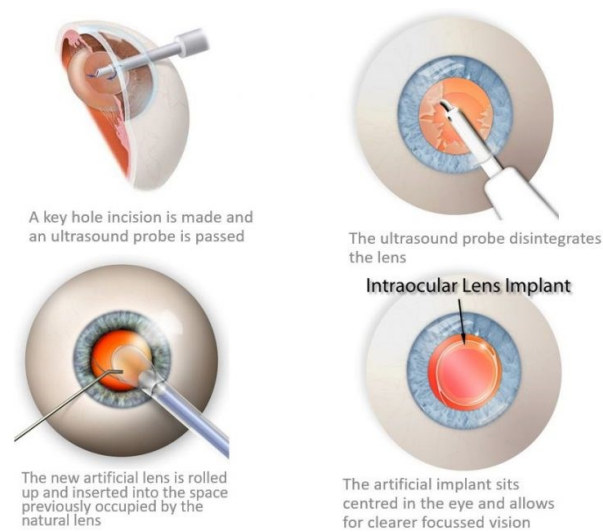


Figure 2.2 Implantation procedure of an IOL.

2.2.1 Monofocal Intraocular Lenses

Monofocal IOLs represent a wide category of lenses. These lenses are specifically engineered for a clear vision at a single focal distance, often optimized for distance vision. So, individuals electing to undergo monofocal IOL implantation will still need the use of glasses or contact lenses to keep their visual acuity during tasks demanding near and intermediate vision, such as reading or working on a computer. Nevertheless, monofocal IOLs offer nearly perfect visual clarity and heightened image quality within single focal length, particularly efficient for activities like driving. In view of their dependable performance and cost-effectiveness, monofocal IOLs present a viable option for patients who prioritize distance vision and have a willingness to supplement their visual requirements for near tasks with appropriate spectacles or contact lenses.

Monofocal IOLs have two main types, aspheric and spherical lenses. Spheric monofocal IOLs possess a uniform curvature across the entire optical surface, resulting in a consistent focal power throughout. While effective in addressing the patient's primary visual needs, spheric IOLs may cause certain aberrations, such as spherical aberration, which can lead to decreased visual quality, particularly in mesophotic conditions. On the other hand, aspheric monofocal IOLs consist of a modified curvature that more closely mimics the natural curvature of the cornea. This design aims to reduce spherical aberration and enhance contrast sensitivity, leading to better visual outcomes. As a result, aspheric monofocal IOLs have been suggested as a viable alternative to eliminate the drawbacks associated with spherical IOLs. However, it is important to

note that the selection between these two types ultimately depends on various patient-specific factors, such as preexisting ocular conditions and individual visual requirements.

2.2.2 Multifocal Intraocular Lenses

The optical features of multifocal IOLs allow light from an object to be refracted or diffracted for both distant and close vision, with some light dispersion happening. These IOLs are classified as refractive or diffractive whose light focusing characteristics are demonstrated in Figure 2.3. But most IOLs nowadays are the combination of these two types and that type will be our concern in this thesis study.

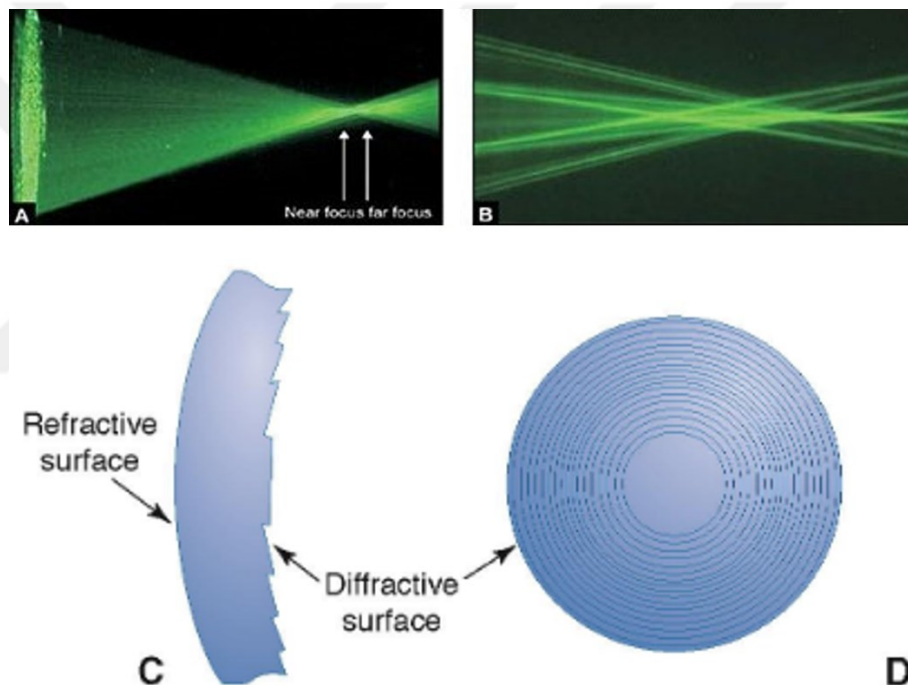


Figure 2.3 Demonstration of optical characteristics of: (A) diffractive multifocal IOLs, (B) refractive multifocal IOLs [Fine et al, 2008], (C) refractive surface on IOL, and (D) diffractive surface on IOL.

There are several ways to divide the light into focal points: For refractive surfaces, multiple anterior surfaces or a posterior spherical combined with multiple anterior surfaces can be used. If refraction and diffraction want to be combined, then, posterior diffractive surfaces can be used with anterior refractive surfaces. Finally, a biconvex lens on the anterior surface can be used for longitudinal aberrations [Alio et al., 2017].

Multifocal Intraocular Lens Types

Refractive models accomplish multifocality with annular zones of varying refractive powers, often providing good enough distant and intermediate vision. These models are sensitive to pupil size changes, need accurate centering, and can result in halos, glare, and decreased contrast sensitivity [Alio et al., 2017, Alio et al., 2014]. To enable functional vision at all distances, some refractive systems integrate a constant shift in curvature across zones [Rosen et al., 2016].

Diffraction models have concentric zones with diffractive microstructures that become closer as they go further from the center. Although these types typically provide good far and close vision, intermediate vision is not always perfect. They rely less on pupil movements and tolerate centering fluctuations more, but they frequently have a more influence on contrast sensitivity [Alio et al., 2017; Alio et al., 2014]. Despite a reduction in contrast sensitivity compared to monofocal IOLs, multifocal IOL patients often retain contrast within the normal range [Alio et al., 2017].

Multifocal IOL optics can be rotationally symmetric or asymmetric. Aspheric designs also eliminate spherical aberrations and improve contrast sensitivity [Alio et al., 2017; Maurino et al., 2015; Alio et al., 2014; Denoyer et al., 2007].

Outcomes

Patients' satisfaction with multifocal IOL implantation depends on the choice of IOL, which considers their problems, visual demands and comprehension of the various optical designs and visual performances of multifocal IOLs. Furthermore, the surgeon's knowledge of good surgical technique is critical [Alio et al., 2017; Alio et al., 2014].

A recent systematic review of studies with peer review found that overall patient satisfaction was quite high. Across all study groups, 0.91 or more of patients acquired spectacle independence of 0.8 or higher for distant vision, fully for intermediate vision, and 0.7 for close vision. The study also revealed that 100% of patients achieved binocular uncorrected visual acuity of around 0.30 logMAR for distant, 0.04 for intermediate, and 0.03 for close visual acuity [Alio et al., 2017; Rosen et al., 2016].

Trifocal IOLs are especially well-suited to those who require good intermediate vision. Today's technology has enabled the development of more advanced multifocal IOLs, such as the increased depth of focus, which results in an elongated focal point. These lenses use unique approaches to increase patient expectations across all distances by increasing the range of vision without dividing light beams. These sophisticated

lenses might also demonstrate a reduced occurrence of visual effects like photic phenomena. However, there needs to be published information about the results of these lenses, and ongoing research is evaluating their possible advantages and limits based on clinical performance [Rosen et al., 2016].

Complaints and Treatment

Because multifocal IOLs are more susceptible to residual refractive errors, residual ametropia is a typical source of patient unhappiness. Incorrect biometric analysis, inadequate IOL power selection, calculation method restrictions, or problems in IOL placement can all result in residual ametropia. Enhancements through myopic and hyperopic laser in situ keratomileusis or photorefractive keratectomy after cataract surgery have been established as effective, reliable, and secure management options. If excimer laser therapy is not accessible, IOL exchange or piggyback lens implantation are other options [Alio et al., 2017; Alio et al., 2014].

In patients with long-term multifocal IOLs, PCO is a common occurrence, frequently resulting in impaired vision and photopic phenomena [Alio et al., 2014; Rosen et al., 2016]. Individuals who received hydrogel IOLs, rounded-edged IOLs, IOLs positioned in the sulcus, and underwent a large capsulorhexis experienced greater occurrences of posterior capsule opacification (PCO) compared to those with different materials [Findl et al., 2010]. Capsulotomy with an Nd: YAG laser is a safe and effective treatment option for PCO [Alio et al., 2014; Rosen et al., 2016; Findl et al., 2010]. However, before having Nd: YAG laser therapy, it is critical to address or rule out other potential causes of patient discontent [Alio et al., 2014; Rosen et al., 2016].

Dry eye is a multifactorial condition that affects the tear film and the ocular surface, causing pain, impaired vision, and photopic phenomena. A dry eye can be caused or exacerbated by cataract surgery, which reduces corneal sensitivity via the incision. Dry eye treatment guidelines include eyelid cleanliness and the use of artificial tears.

Insufficient pupil size can negatively impact visual clarity after the insertion of multifocal IOLs, as the size of the pupil determines which regions of the multifocal IOL come into play. Cyclopentolate drops, or a 360° argon iridoplasty, can be used to treat patients with tiny postoperative pupils with impaired near vision.

Visual function can be compromised by IOL decentration, which is influenced by factors such as the extent of displacement, the design of the IOL, and the size of the pupil. Argon laser iridoplasty is frequently used to avoid IOL explantation. However, if

explanation is required, it is best to do so within the first six months of surgery to avoid scarring tissue and problems. A capsular tension ring during explantation may also make the surgery easier [Alio et al., 2017; Rosen et al., 2016; Fernandez-Buenaga et al., 2012].

2.3 *Intraocular Lens Evaluation*

Throughout the years, people have come up with many different evaluations to understand the optical performance of intraocular lenses. These evaluations may be categorized into two main branches: subjective and objective.

Subjective evaluations generally consist of questionnaires filled out by the patients after the cataract surgery to assess their satisfaction and quality with their new vision. Various questionnaires were created for this purpose; some of the most popular ones are the VF-14 questionnaire, the Cataract TyPE Specification questionnaire, and the Cataract Symptom Score.

The objective evaluations can be further categorized into clinical and optical bench tests. In the following sections, we will discuss the details and different types of these tests.

2.3.1 *Through Clinical Tests*

Defocus Curves

The defocus curve test is a common practice among ophthalmologists to assess patients' vision through an IOL after cataract surgery [Sheppard et al., 2013]. It measures visual acuity in different optical defocus ranges by utilizing a series of lenses with positive and negative power. This way, distances from near to far are simulated, and in every step, the patient's visual acuity is measured.

The procedure starts by finding the best-corrected distance visual acuity of the patient. This visual acuity is then graphed on the Y-axis at zero, which symbolizes infinity. The best-corrected distance visual acuity is then adjusted using minus lenses. Viewing through a negative lens produces divergent light beams with the same refractive effect as moving the eye chart closer to the viewer. A -1D lens indicates a 1m viewing distance in this manner. Similarly, a -2.50D lens indicates 40cm or around 16 inches.

The distance between the viewer and the eye chart can be measured in centimeters, inches, or diopters. The defocus curve for close objects is created by plotting the acuity while seeing through a sequence of increasingly negative lenses. An example defocus curve chart is given in Figure 2.4.

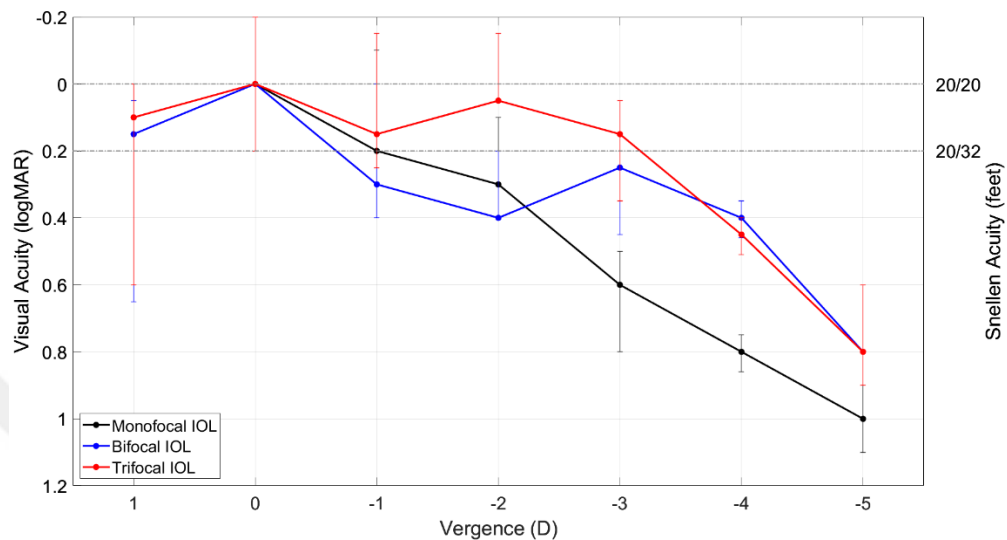


Figure 2.4 Example of defocus curve results for 3 different IOLs.

Although there isn't a standardized method for evaluating intraocular lenses with the defocus curve, it is suggested that randomization of the reading chart is necessary for the most accurate results. Previous studies show that there is a significant difference in results when the reading chart is randomized or not [Ram et al., 2007].

Contrast Sensitivity

The contrast sensitivity test assesses the patient's ability to discriminate between an item and its background. This aspect of visual function is essential in settings with restricted lighting, such as low-light surroundings. For example, enough contrast sensitivity is required to provide clear vision in cases of fog or excessive glare.

In an academic setting, contrast sensitivity differs from visual acuity in that it assesses the patient's clarity of vision at a distance. The patient may have excellent visual acuity but poor contrast sensitivity, or vice versa. Because both phrases refer to the ability to see details, contrast sensitivity and visual acuity may need clarification. Nonetheless, it is critical to distinguish between them. Contrast sensitivity includes evaluating two factors, size, and contrast, instead of visual acuity, which examines size.

Contrast sensitivity may be tested and measured in an academic setting using two basic methods: the Pelli-Robson Contrast Sensitivity Chart and the Sinewave Grating.

To identify contrast sensitivity difficulties, the Pelli-Robson Contrast Sensitivity Chart is often used. It is made up of horizontal lines of uniformly sized capital characters. This vision test is often performed with the patient wearing corrective eyeglasses or contact lenses, if necessary. The contrast of each letter decreases as the patient progresses from left to right along the chart. Beginning at the top of the chart, the individual reads each row until the letters are no longer visible against the white backdrop. Furthermore, the Mars Letter contrast sensitivity test uses a chart like the Pelli-Robson test, although with a smaller chart size and closer viewing distance. The Pelli-Robson chart is given in Figure 2.5, for the reference.



Figure 2.5 Pelli-Robson chart used for contrast sensitivity test.

The use of sinewave grating targets is a more advanced method of testing contrast sensitivity (see Figure 2.6). These gratings appear as fuzzy lines with alternating bright and dark hues of grey and varying line thickness. During the exam, the patient is shown various gratings on a computer screen or a wall chart. This method provides information on the patient's visual ability in low, medium, and high contrast circumstances. The doctor can determine the patient's CSF by putting the data on a graph. Notably, CSF is more informative than visual acuity in determining a person's spatial vision ability.

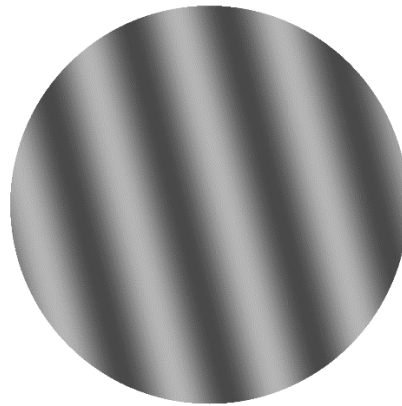


Figure 2.6 A sinewave grating example for determining visual contrast sensitivity as a function of spatial frequency.

Contrast sensitivity test is inferior in popularity compared to defocus curve test. Most hospitals omit this test after the surgery and solely focus on visual acuity of the patient. Regardless, contrast sensitivity changes due to intraocular lens implantation are still a crucial factor to assess the lens quality and it should be made a common practice to evaluate it after cataract surgery.

Photic Phenomena

Objectively characterizing the severity of photic phenomena (halo, glare, and starburst) issued by IOLs is still challenging in clinical assessments. There are no standard charts or metrics to quantify these phenomena. In clinics, the severity of photic phenomena is generally assessed by questionnaires given to patients after surgery, asking them to rate the occurrence from mild to severe.

Previous studies also tried to objectively characterize photic phenomena by utilizing a vision simulator [Ukai et al., 2021]. This vision simulator lets the user change the parameters of the simulated photic phenomenon, such as the size and intensity of the halo, glare, and starburst. By iteratively asking the patient how close the simulated photic phenomenon looks to the ones they experience, the study tried to parametrize and understand the severity. Although this was a more objective assessment than just conducting a questionnaire, the accuracy and precision of the results still heavily depended on the patient and how well they could evaluate their vision.

2.3.2 Through Optical Bench Tests

Optical benchtop tests are crucial for assessing the performance of an IOL objectively before surgery. A component mimicking the eye should be used for the most realistic results, which is generally called an artificial eye model or optomechanical eye model. There are two commonly accepted eye models on which newer and more complex eye models are based: the Navarro eye model [Navarro et al., 1999] and the Gullstrand reduced eye model [Gullstrand et al., 1909]. Although these models work with very good performance, they are not the standard for IOL assessments.

Standard Eye Models

The standard for assessing IOLs [ISO, 1999] specifies a model eye with an artificial cornea that is nearly aberration-free. This is accomplished using commercially available achromatic doublets [Norrby et al., 2007]. Any aberrations in such a model eye are caused by the IOL being measured. In the literature, several eye models were designed considering these standards. In general, these eye models usually have common types of parts and are suitable for benchtop or handheld usage: a lens in place of cornea, an aperture working as the iris, a second lens for the crystalline lens, a camera corresponding to retina. Depending on the model eye's usage, these parts can be dynamic or static, changing the distances between each other.

These components are assembled inside a chamber which is filled with aqueous or vitreous fluid, simulating the medium inside the eye. Although aqueous and vitreous fluids do not correspond to the same fluid possibly distinguishable, most eye models consider them the same for simplicity and are modeled as chambers filled with a single type of fluid.

Cornea

The cornea is the main optical power of the human eye. A healthy cornea shows a refractive index of 1.376 at 0.589 μm [Navarro et al., 1999]. It has two separate curvature radii, anterior and posterior, where the former is around 7.80 mm, and the latter is around 6.50 mm. Its thickness ranges from 0.5 to 0.6 mm, and its asphericity conic constant is between -0.15 and -0.30 on both surfaces [Dubbelman et al., 2006, Xiong et al., 2017].

Generally, a lens made of polymethylmethacrylate, or fused silica mimics the cornea. This kind of lens usually has a refractive index between 1.4 and 1.5, which is

enough for approximation but slightly overestimates the actual value of the cornea. Another alternative to polymethylmethacrylate lenses is glass corneas, which increase the eye's power. However, previous studies showed that those kinds of corneas still could not outperform the former lens [Norrby et al., 2007].

Although most researchers aim to mimic the anatomical function of the cornea, its shape depends on the application area of the model eye. The most popular cornea-shape option is a curved lens based on the Gullstrand model [Gullstrand et al., 1909]. The mimicking lens's curvature radius is generally close to 7.7 mm–7.8 mm and 6.4 mm–7.5 mm for the anterior and posterior surfaces, respectively. The typical thickness of the lens is 0.5 mm for a genuine eye approximation.

The eye models based on ISO 11979-2 recommendations for IOL evaluation generally aim to have minimal or a specific amount of spherical aberration in the cornea [ISO 11979-2]. They utilize an aspheric achromat doublet lens as a cornea for this purpose but sometimes slightly modify the ISO cornea model to approximate the human eye's asphericity.

Although many other modifications can be made to the lens mimicking the cornea, the discussed parts are the main specifications.

Iris and Pupil

Many optical properties, such as the diffraction of the eye or optical aberrations and so on are directly affected by the size of the pupil [Kawamorita et al., 2005]. To not neglect the effects of pupil size on the function of the eye, a circular aperture acting as an iris with a pupil is included in the eye models. The actual human iris is not smooth, but most eye models overlook this fact. A contrary example is usage of a curved ring made with a rough texture or tapered edges which mimics the iris-lens interface, reducing unwanted diffraction [Ohnuma et al., 1999, Sheehy et al., 2002].

Most of the applications require varying size of the pupil to simulate different illumination conditions. For this reason, most model eyes have either adjustable or replaceable pupils with a diameter between 2.0 mm to 8.00 mm. The typical increment of pupil size in the eye models is 1 mm or 0.5 mm.

The iris itself has a thickness of around 0.35 to 0.45 mm and this is another important factor for properly mimicking the eye. This requires some solutions for how the iris is attached to the model eye since such thin irises can cause difficulty in handling [Li et al., 2021]. One of the solutions is directly milling the iris to device

enclosure [Shen et al., 2009]. Another solution is using the holder of crystalline lens and mounting iris and lens on the same holder with replaceable properties [Sheehy et al., 2002].

As mentioned above, some eye models use replaceable or adjustable which are either manual or automatic [Schneider et al., 1997]. An example of the automatic case was done previously where a feedback mechanism fed by the light sensor of the eye model controlled the pupil size [Santiago-Alvarado et al., 2015]. Another example featured a phase plate with a light limiting section around it, allowing the combined device to regulate aberration while simultaneously serving as a pupil [Yamaguchi et al., 2006].

The entrance pupil which is also used commonly should not be mistaken for the pupil discussed in this section. The main purpose of entrance pupil is to limit the light entering the eye and it is usually placed in front of the eye which is different from the pupil.

Crystalline Lens

The crystalline lens has great importance in eye models since its fluctuations are the main cause of defocus and spherical aberrations of the human eye [Fernández et al., 2007]. The discussion considering crystalline lens and mimicking its continuous power change can be discussed thoroughly by itself. However, our main purpose in this thesis is to evaluate IOLs and we will only discuss the parts concerning IOLs.

In most of the eye model cases, rather than an explicit crystalline lens, and IOL is used regardless of the application including IOL evaluation or not. In the case of IOL evaluation, the IOL is used to simulate an aphakic eye implanted with an IOL. In the other case, IOL is just used for simulating the crystalline lens [Xie et al., 2014]. There are also models where a crystalline lens and an IOL used together [Arianpour et al., 2013].

The crystalline lens is attached inside the eye model through a holder as discussed in the previous section. In some cases, this holder is detachable from the eye model and in other cases it is directly fixed. In general, the haptics of the IOL are utilized to suspend it with a platform like holder but some special cases need the IOL without the haptics [Arianpour et al., 2013]. An important example concerning our thesis is where two membranes are used to sandwich an IOL between them to evaluate posterior and

anterior capsulotomy [Trinavarat et al., 2001]. However, this model was not used for assessing visual functions of IOLs but rather assessing other functionalities.

In most clinical cases, the main concern regarding IOLs is their alignment in the eye. To assess the effect of alignment in the visual performance of the IOL, the holder is designed to possess the ability to rotate and translate in multiple angles and directions, allowing for the adjustment of the lens position relative to the optical axis [Dubbelman et al., 2001, Barry et al., 2001, Barbero et al., 2003]. While the decentration is generally simulated between 0 and 2 mm with 0.1 mm increments, tilt is simulated between 0 and 5 degrees with 0.01 degrees increments. Such IOL position adjustments are done manually, and, in some cases, they are done automatically utilizing control mechanisms with position sensors [Gatinel et al., 2013].

Retina

One should not underestimate the power of imitating retina and its viewing properties accurately in an eye model. To imitate the retinal surface, many layouts have been employed. There are four types of retinas: sensor retinas, diffusely reflecting retinas, transparent retinas, and active retinas. In this discussion we will keep our focus on sensor retinas since it was the main method chosen in our artificial eye model. A sensor, such as a charge-coupled device or a complementary metal-oxide semiconductor, is always a part of retina imitation.

There are advantages and disadvantages when choosing the sensor type between charge-coupled device and complementary metal-oxide semiconductor. CCDs offer better image characteristics and lower noise than CMOSs, but they need more power and have bigger pixel sizes [Förster et al., 2015]. Larger pixel sizes are more sensitive to light, allowing for smaller aperture diameters, which allows for less aberration. They may, however, be less suited for modeling photoreceptor size. There are other sensor choices used by previous studies like photodiodes, radiometers etc. [Ohnuma et al., 2011; Regal et al., 2020]. Each sensor type serves a different application.

Since most sensors are not suitable to be used in aqueous or vitreous fluids, a glass lamina is placed on the retina plane for protecting them from the medium inside the eye model. This leaves the sensors outside of the eye model. The focal plane usually occurs at the anterior surface of the glass lamina used as the retina plane. Previous studies claim that the deformations of the optical system can be avoided better if the image is focused slightly behind the lamina [Barcik et al., 2008]. The image is transferred to the

sensing surface by relay optics which captures the focused image, the magnification ratio depends on the specifications of the sensor which is mentioned below. To overcome an important limitation discussed in the following paragraphs, the lamina surface doesn't have to be perfectly planar but has some displacement such as tilt or decentering [Amorim et al., 2022].

The pixel size of the sensors used also holds an important place. Its preferential size is generally between 2.0 – 3.0 μm in foveal region. If the pixel size and the resolution related to it is not enough, optical magnification before the image reaches the sensor is the solution.

An avoidable limitation in using sensors as retina lies upon their geometry. The planar surfaces of the sensors do not let the eye model accurately simulate the curved image formed by the retina. A planar surface cannot naturally correct the effect of field curvature since it can't allow simultaneous focus of on-axis and off-axis beams [Amorim et al., 2022]. If only the area around the fovea is concerned like in our simulator case, then the planar sensory surface is acceptable since our concern is only of the central vision [Dimmick et al., 1941]. However, if the eye model application considers the area beyond macula lutea, the curvature of the retina should not be overlooked. Unfortunately, fabrication of a curved sensor with dimensions as small as the retina is extremely difficult to achieve [Guenter et al., 2017].

One solution to overcome the mismatch between the curved retina and planar sensor is to use an extra lens, compensating the field curvature and as a result focusing the image correctly on a planar surface. But this solution requires additional optical components inside the eye model which is not desirable since with every additional optical component, new aberrations are introduced to the model, diverging from the actual human eye aberrations [Amorim et al., 2022]. As an alternative approach, posterior chamber depth can be adjusted between on-axis and off-axis measurements to compensate for the perpendicular distance [Coughlan et al., 2017]. However, this method is not suitable for fully image formation purposes and is only limited to point-like measurements.

2.4 *Intraocular Lens and Cataract Simulators*

There are some IOL simulators to show vision through IOLs to cataract patients prior to surgery. However, they do not work well for patients with moderate or dense cataractous lenses due to impaired vision and heavy scattering. One such case is an IOL simulator that uses a biconcave lens back-to-back with an IOL placed in a thin wet chamber. The compact lens assembly has a limited field-of-view but allows patients to see the world through a multifocal intraocular lens [Na et al., 2022]. Similarly, the Rassow telescope follows the same architecture as the intraocular lens simulator but utilizes a convex lens [Wahl et al., 2019]. VirtIOL (10Lens S.L.U., Barcelona, Spain) is another device for simulating vision through intraocular lenses [Studený et al., 2018]. It allows the subject to obtain the defocus curves of patients as if they were wearing the IOL of their choice. SimVis (2EYEVISION, Madrid, Spain), which prescribes multifocal corrections by utilizing the temporal multiplexing of the lens power [Akondi et al., 2018]. VAO (VOPTICA, Madrid, Spain) is another device with integrated adaptive optics for simulating multifocal IOLs with various measurements and functionalities such as ocular aberrations, subjective refraction, and visual tests [Hervella et al., 2019].

2.5 *Digital Light Control*

This section introduces the main method used in our holographic display based intraocular lens simulator. In this thesis, we utilized computer generated hologram calculation for achieving digital off-axis holography to send the characterized contents created to non-cataractous regions on the patient's pupil.

2.5.1 *Computer Generated Holography*

Choice of the Method

Holographic displays offer unique capabilities for direct-view displays and virtual and augmented reality applications. Among all technologies, holography stands out as the sole technology capable of generating programmable exit pupil patterns [Chang et al., 2019]. In a prior study, our group leveraged this characteristic to conduct preoperative vision screening combined with cataract screening and verification for cataract patients. This screening involved the implementation of a distinct approach

known as the eye-box steering method, previously developed by our group [Kavakli et al., 2021].

In this thesis, the aim was to expand upon the previous experiments conducted by our group so for the simulator, CGH was chosen as the main method.

Theory

Computer-generated holograms, synthetic holograms, and computer holograms are various names for a type of hologram that is formed as graphical output from a digital computer. The computer may calculate the amplitude transmittance of the hologram and see the output on a CRT or plot it on paper by supplying a mathematical representation of a wavefront or an object through an array of points. CGH, like traditional holograms, can be classified as image holograms, Fourier transform holograms, or Fresnel holograms based on the connection between the object and the complex wavefront captured in the hologram [Lee et al., 1978].

In this thesis work, the CGH calculations were done based on the Fourier transform holograms displayed in near eye. Therefore, in the following subsections, holographic near eye displays and CGH computation process specifically for Fourier transform holograms will be discussed.

Generic Holographic Near Eye Display

The holographic display-based simulator discussed in this thesis can be classified as a HNED. In general, a HNED has 4 main parts: a light source, a phase modulator, which is an SLM in our case, an eye box plane where the final hologram occurs and optical elements in between these. HNED architecture with main components is given in Figure 2.7.

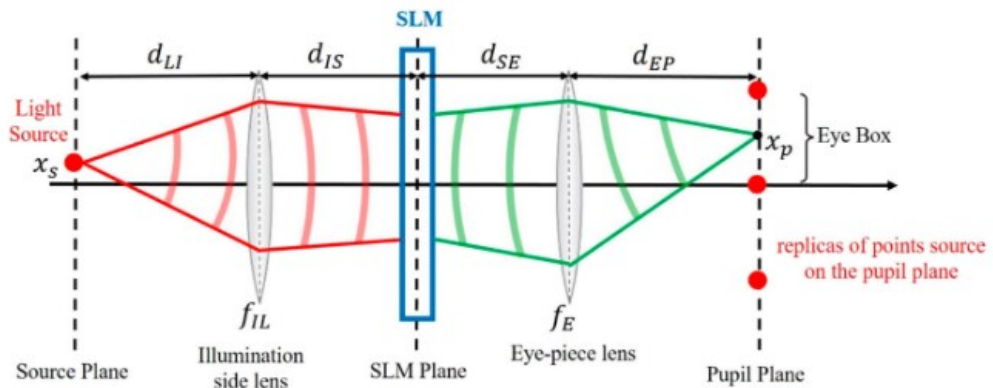


Figure 2.7 Generic HNED Schematic [Kazempourradi et al., 2019]

Taking the schematic as a reference, the path of the light through an HNED is as follows: Firstly, the point light source emits the chosen light, which is generally coherent (laser) or partially coherent (LED). The emitted light propagates to the illumination lens with a focal length of f_{IL} . Passing through the illumination lens, light becomes collimated and propagates until it reaches the SLM. SLM is a thin multiplicative optical element that modulates the light according to the given phase pattern. After modulation, the light propagates, passing through the eyepiece lens and reaching the pupil plane. On the pupil plane, the modulated light forms an eye box.

This summarizes the process from emission to eye box formation. In the following subsection, we will discuss the details of how the phase pattern used on SLM is calculated through a CGH algorithm.

CGH Computation Process

The CGH computation process to achieve phase patterns for Fourier transform holograms consists of several steps. A block diagram of the CGH algorithm used for our holographic display simulator is given in Figure 2.8. This algorithm was developed previously by our group to achieve optimized speckle-free image formation.

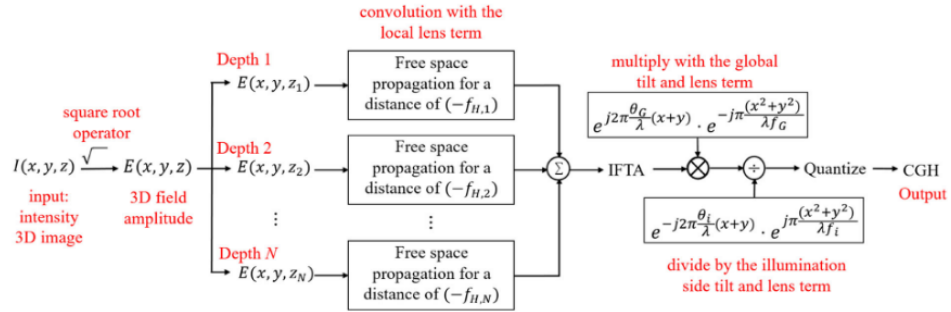


Figure 2.8 CGH Algorithm block diagram [Kazempourradi et al., 2019]

In this computation process, scalar wave optics theory and paraxial approximation was utilized. Firstly, the algorithm takes an intensity 3D image consisting of a 2D image and its corresponding depth map as input. The viewpoint of this image is the center of the eye box (x_p). The square root operator is applied to input image to achieve field amplitude and then this field amplitude is sliced into N planes with different distances (z_i) according to the depth map information. Each plane is then convolved with a lens term given below:

$$f_{H,i} = \frac{(f_E(d_{EP} + d_{SE}) - d_{EP}d_{SE})^2}{z_i f_E^2} \quad (2.1)$$

Where f_E is the eye-piece lens focal length, d_{EP} and d_{SE} are the distances between the eye-piece lens and pupil plane, SLM plane, and eye-piece lens, respectively. z_i is the distance between each plane defined by the depth map. This convolution operation computes an intermediate field for each slice using fast Fourier transformation (FFT)-based free-space propagation. A full-complex hologram is achieved with the superimposition of intermediate fields and encoded into a phase only CGH. To preserve the quality, rather than just discarding the magnitude, an iterative Fourier transform algorithm is used to compute the phase only CGHs. Eye box location and the light source are controlled by superimposition of lens and global tilt terms on the CGH. The terms are: Global lens term of focal length,

$$f_G = \frac{d_{SE}(f_E - d_{EP}) + d_{EP}f_E}{f_E - d_{EP}} \quad (2.2)$$

Global tilt angle (in radian),

$$\theta_G = \frac{x_p f_E}{f_E d_{SE} + f_E d_{EP} - d_{SE} d_{EP}} \quad (2.3)$$

Where x_p is the eye box location. Focal length of illumination lens,

$$f_i = \frac{d_{LI}(f_{IL} - d_{IS}) + d_{IS}f_{IL}}{f_{IL} - d_{IS}} \quad (2.4)$$

Where f_{IL} is focal length of the illumination lens. d_{LI} and d_{IS} are distances between source plane and illumination lens; illumination lens and SLM plane, respectively. And finally, illumination tilt angle,

$$\theta_i = \frac{x_s f_{IL}}{f_{IL} d_{LI} + f_{IL} d_{IS} - d_{LI} d_{IS}} \quad (2.5)$$

We can make physical interpretations for these terms: The incoming illumination wave is efficiently transformed into a collimated plane wave, which is normally incident using the illumination lens and tilt terms. The global lens and tilt terms image this normally incident plane wave at the center of the eye box. To image the point light source onto a plane at z_i , the lens term f_{Hi} is added to the CGH. This forms a virtual object point at the desired place, concluding the CGH computation process.

2.5.2 Liquid Crystal Phase Modulators

Modulator Choices

The liquid crystal phase-only modulator is a popular type of modulator to display phase-only CGH in optical architectures. An electric signal controls the orientation of the birefringent liquid crystals within each pixel, resulting in a change in the phase delay imposed on the incoming light beam by each pixel. These modulators generally have response times ranging from 20 to 200 Hz. They are, however, plagued by some difficulties, such as space-bandwidth-product requirements, and nonlinear modulation characteristics that demand calibration. Following subsections go into further detail on these topics. Nonetheless, SLMs were chosen as the phase modulator of the holographic display simulator due to having significant benefits such as excellent resolution, exact bit-depth modulation, and good light efficiency.

Space-Bandwidth Product

One of the biggest limitations in holographic displays utilizing a SLM is the space-bandwidth product requirements. Some holographic display techniques employ a table-top television-like arrangement to create ghost-like 3D objects floating in mid-air, allowing several users to move and spin around them [Sasaki et al., 2014]. Unfortunately, large area SLMs with micron-level pixel pitches are required for such systems, resulting in significant Space-Bandwidth-Product requirements that surpass the capabilities of present SLM technology [Yaras et al., 2010]. As a result, demonstrations are now confined to very tiny items and tight viewing zones, with little room for major short-term improvements.

Other Non-idealities

Due to production restrictions, SLM panels often have minor flaws in flatness [Xun et al., 2004]. This problem may be solved in the case of phase SLMs by monitoring the wavefront distortion and applying an opposing phase shift to the displayed patterns. Alternatively, more thorough correction approaches [Cizmar et al., 2010] can address additional faults in the optical system at the same time, such as non-flat lighting beams or lens aberrations.

When showing simple phase gratings on phase SLMs, another typical problem is the appearance of many diffraction orders. The first set of diffraction orders is caused by the device's pixelated nature, in which the pixels form a regular grating, causing light

to diffract at regular-spaced angles according to the grating period (i.e., pixel pitch). Even when the SLM is switched off, the same phenomenon occurs. The second difficulty emerges when displaying a phase pattern because not all incident light is modulated, and some incoming light is reflected off the SLM panel. This component is commonly known as the zero-order component which could be overcome by spatial filtering in the optical setup.

2.5.3 Programmable Exit Pupil Patterns

Our group utilized a method called eye-box steering in previous studies [Kavaklı et al., 2021]. This method was an essential part of the thesis since the main targets of our research were cataract patients. By utilizing this method, we overcome the visual quality decrease due to cataract of patients' eye and show them the contents we created in the desired quality.

By computationally incorporating their wave shaping features in the wave calculation, CGH algorithms enable us to generate the influence of numerous optical components on the display system. As a result, CGH is the only technique that gives total computational control over the size, placement, and shape of the exit pupil (eye-box).

The previously discussed reconstruction procedure enables us to place the generated hologram anywhere between 0th and 1st diffraction orders. The following grating formula can be utilized to solve where our hologram is placed for each color.

$$d_{x,y} = \frac{m_{x,y}\lambda f}{a} \quad (2.6)$$

In the current optical configuration of the simulator, $d_{x,y}$ corresponds to $5.2mm$ in both axes, giving the distance between 0th and 1st order. f is the focal length of the eyepiece lens, which is $50mm$, and a is the pixel pitch of our modulator which is $4.5\mu m$. $m_{x,y}$ is the chosen orders between 0 and 1 to place the hologram.

A linear grating phase component along the x and y-axes, featuring periods that are proportional to m_x and m_y , was introduced to direct the focal point of the eye within the square region located between the 0th and 1st diffraction orders in the first quadrant. For the horizontal and vertical axes and each hue, a new value of m had to be computed. The optical configuration, exit pupil position corresponding to $m_x = m_y = 0.5$ for the blue wavelength ($473nm$) and the area was partitioned into 5-by-5 subregions can be seen

in Figure 2.9, published previously by our group. This results in 25 distinct exit pupils, each carrying the same generated content, overcoming the cataracts of the patient.

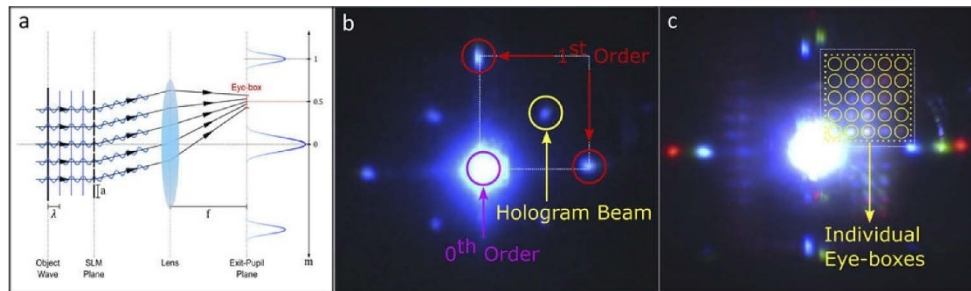


Figure 2.9 a) Diagram depicting the exit pupil position between diffraction orders. b) Photographic capture of a single exit pupil at 473 nm. c) Real-time capturing of all exit pupils at all wavelengths [Kavakli et al., 2021].

Utilizing the discussed method, in clinical assessments, any hologram can be steered on the cataract patient's pupil to achieve best results. This is an enormous advantage of holography method compared to other types of IOL simulators. Combined with densitometer data simultaneously captured in the process of assessment, one can be sure about their steering results.

Chapter 3:

**ARTIFICIAL EYE MODEL AND INTRAOCULAR LENS
CHARACTERIZATION****3.1 Introduction**

Cataract surgery involves removing the opaque lens and implanting an IOL [Ramke et al., 2017]. There are various types of IOLs. Monofocal IOL with fixed focal length has been a common choice. However, there are also multifocal IOLs engineered to provide good unassisted vision for both distant and near objects. These multifocal IOLs can have two foci (bifocal), three foci (trifocal) or extended depth-of-field (EDOF). Despite their design to provide multiple foci for clear vision without glasses and their overall good optical performance, multifocal IOLs have been associated with undesirable visual phenomena in 97 of 334 instances (29%) in 7 clinical trials. This contrasts with monofocal IOLs, which had undesirable visual phenomena in only 26 of 328 cases (8%) [de Silva et al., 2016]. The incidence of postoperative complaints related to photic phenomena and side effects continues to be one of the main concerns associated with multifocal IOLs [de Silva et al., 2016; Khandelwal et al., 2019].

Various studies have attempted to evaluate the effects of IOL on vision after surgery. [Blehm et al., 2021; Mencucci et al., 2018] However, most of these studies have relied on subjective feedback from patients, making it difficult to draw objective conclusions. In contrast, some studies have utilized optical bench setups to evaluate the performance of IOLs in vitro [Amorim et al., 2022]. While these evaluations provide more objective data, they are not easily accessible to cataract patients or ophthalmologists who work with them.

With today's technology, cataracts can be treated with a simple surgery; however, some severe complications persist afterwards, such as PCO or decentration. PCO is one of the most common complications requiring treatment post-operatively in up to 50% of patients [Liu et al., 1996]. Treatment of PCO can be done with YAG laser capsulotomy.

Although cataracts and PCO are easy to treat and there have been standardized methods to assess the symptoms caused by them, their progression is slow, making

them hard to keep track of. People with such complications may not notice them until cataracts or PCOs become severe, significantly decreasing their quality of life.

There have been previous studies that aimed to simulate how cataracts affect vision. Augmented and virtual reality were the common choices in these simulations since these technologies were often used in ophthalmologic studies [Aydingogan et al., 2021]. CatARact uses augmented reality technology to show individuals wearing a stereoscopic head-worn display of cataract symptoms [Krosl et al., 2020]. A study uses the Unreal Engine 3 game engine to develop a simulation of visual impairments, including cataracts [Lewis et al., 2011]. Another study using an Oculus Rift head-mounted display simulated visual disabilities. In contrast, navigating users in a virtual 3D city model [Vayrynen et al., 2016] Finally, a previous study utilized mixed reality to simulate low vision in real time [Acevedo et al., 2022]. These studies are precious simulations utilizing virtual reality. However, their shortcoming is that they do not address the progress of cataracts but instead simulate the nearly final stage of cataracts and their effect on vision. Unlike the discussed simulators, another study designed an optical device using protein denaturation in egg albumen [Mandracchia et al., 2016]. This study could successfully mimic the progression of cataracts and their effect on vision, but this progress was irreversible. Once the progress was complete, the device had to be set up from scratch for another cataract mimic.

In this chapter of the thesis, we will discuss the characterizations done on IOLs, how these characterizations were carried out and the analysis results. On next chapter, these characterizations were combined with holographic display technology to create a novel method to predict the effects of IOLs on patient's vision prior to cataract surgery. We also tried to address the shortcomings of cataract simulators by physically mimicking cataracts and their progression on an IOL. We conducted simulations to assess visual outcomes in individuals with cataracts, employing an artificial eye model and a chemical process on intraocular lenses. By repeating the same method on one side of the IOL, it is also possible to assess the vision with a progressive PCO. The influence of varying degrees of cataracts on visual quality was objectively evaluated by acquiring modulation transfer function curves and spectroscopy results. Furthermore, by simulating opacification in a structured fashion, we examined the variations in results within the patient's pupil, transitioning from regions with lower cataract density to those with higher density, using our holographic display as a testing platform.

3.2 Methods

3.2.1 Artificial Eye Model

To objectively test the effects of intraocular lens implantation on human vision and photic phenomena, we designed and built a 3D printed artificial eye model, which is adapted from a previously developed eye model [Gobbi et al., 2016]. As illustrated in Figure 3.1, the model consists of a PMMA scleral lens with +4D optical power that acts as an artificial cornea, an intraocular lens immersed in liquid, and a glass lamella that serves as the retina. Furthermore, since the pupil size strongly affects the vision function of the eye, we mimicked the iris with an adjustable aperture for different conditions. We designed and optimized the system in Zemax. The resulting distances are 7 mm between the cornea and the IOL, and 19.7 mm between the IOL and the retina with a total optical track length of 20.7 mm. The model was used to characterize the optical performance and photic phenomena associated with different IOLs [Akyazi et al., 2023].

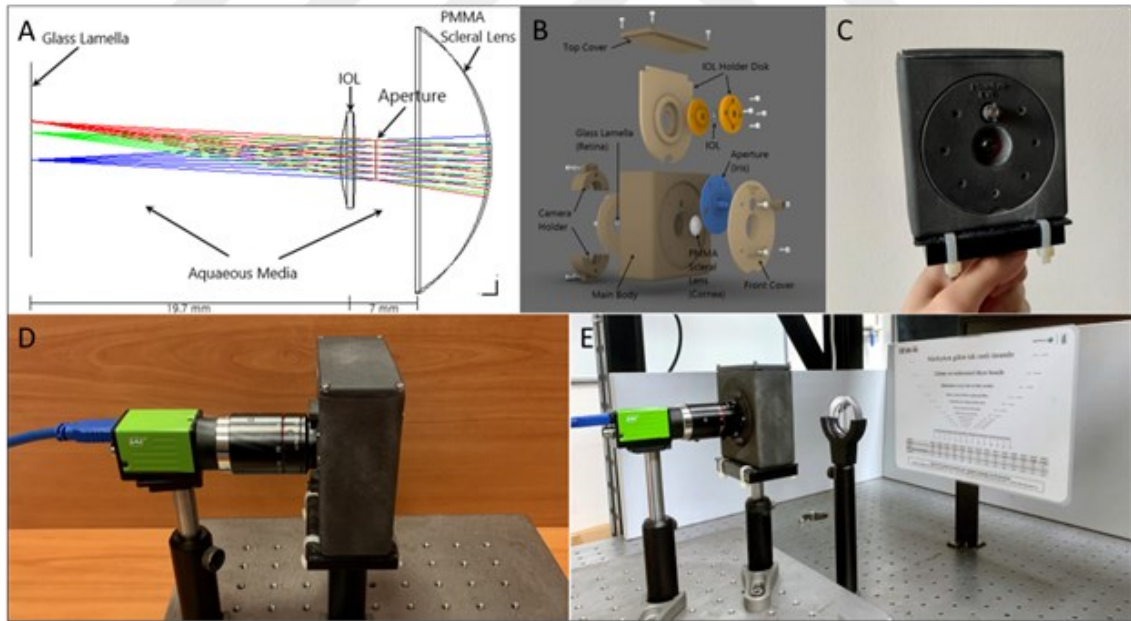


Figure 3.1: (A) Zemax model of artificial eye consists of a PMMA scleral lens, aperture, IOL, and glass lamella. (B) 3D schematic of the eye model. (C) 3D printed and liquid filled eye model. (D) imaging system including camera and microscope objective (E) Full optical setup used for IOL characterization.

3.2.2 *Artificial Cataractous Lens*

We developed a procedure utilizing the reaction between distilled water and acetone to create a reversible and progressive opacification on the IOL surface. As discussed before, this was done with the aim of simulating PCO and cataract on human eye and how it affects vision. The procedure consists of the following steps:

1. The IOL is cleaned with isopropanol to get rid of any previous opacification or dirt on it.
2. The cleaned IOL is dipped into distilled water and kept in water for about 15 seconds.
3. The wet IOL is directly dipped into pure grade (99.5%) acetone and again kept for about 15 seconds.
4. Then, the IOL is put aside for it to dry for 30 seconds.
5. Finally, once the processed IOL is put into a wet cell, it gradually becomes opaque over a period of 300 seconds as it stays in the water.

The advantage of this method over other methods is that this is a reversible process, so the opacification can be cleaned anytime to repeat and observe the effects on vision. Although this process wears out the IOL as it is repeated so the repetition limit is up to 20 times before the IOL starts to break apart.

Another advantage of this method is that by partially applying the acetone to the opaque surface of the IOL, a cataract-like pattern can be created to simulate an even more realistic cataract case. Depending on how long the acetone is applied on the surface, while some areas are having dense cataract, other areas can be having mild cataract or no cataract at all.

The opacification progression through time is shown in Figure 3.2. In intervals of around 40 seconds, the IOL was taken out of the wet cell and captured on a meter. One can observe that as the time passes, the IOL becomes visually opaque, barely transmitting light and the number behind it gets less visible, eventually not visible at all. The limit of 300 seconds was chosen because after 300 seconds, the change in degradation of the vision was barely noticeable with the human eye.

The vision through the IOL with various opacification levels was captured by utilizing the artificial eye model discussed previously. To objectively quantify the degradation of vision, several assessments were done. By capturing and analyzing a

slanted edge, MTF charts for different opacification levels were found. Using a spectroscope, the transmission throughout a spectrum between 200 nm to 1000 nm was measured. With a photodiode, the transmission as well as scattering for a light source of 530 nm wavelength was measured. Finally, by analyzing the color histogram of captures through the opacification progress, the shift in colors was found.

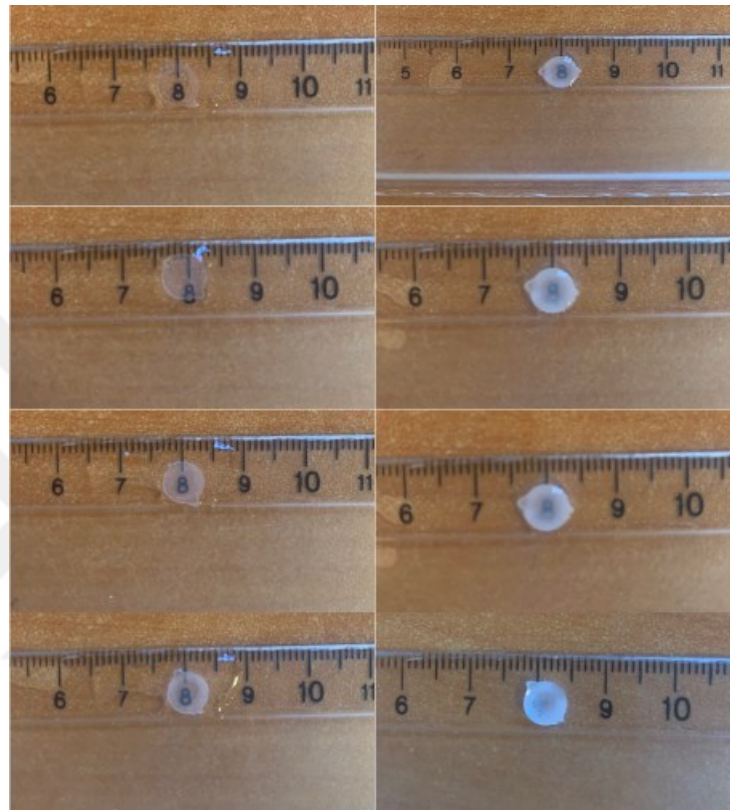


Figure 3.2 Opacification progress on IOL from 0 seconds to 300 seconds. The progress was captured every ~40 seconds.

3.2.3 Optical Characterization through IOLs

Our objective in using the eye model was to understand the impact of IOLs on the retinal image. We used the setup shown in Figure 3.1 with a varifocal lens (focal length ranging from 12 mm to infinity, manufactured by Kowa, Düsseldorf, Germany) to focus the image produced by the artificial eye model onto the scientific camera (3-sensor R-G-B prism area scan camera manufactured by JAI A/S, Valby, Denmark). We tuned the settings, such as gain and exposure, of the camera as close to natural conditions as possible and captured various scenes and charts for the following characterizations.

In this setup, we tested and characterized three IOLs: (i) Tecnis monofocal IOL, (ii) Tecnis multifocal IOL (with +3.25D add) as bifocal IOL, and (iii) Alcon Panoptix

IOL (with +2.17D and +3.25D add) as trifocal IOL. We first tested the setup by capturing real world scenes to see if we could use the artificial eye model to observe the outside world through IOLs as we had expected. This was followed by capturing various standard images such as Snellen charts, contrast sensitivity charts, and halometer charts to characterize the visual functions of the IOLs. The captured images were analyzed and compared to quantify the visual acuity, contrast sensitivity, and halo severity for each IOL.

Defocus and Resolution Limits

One of the most significant outcomes of multifocal IOLs is complete independence from eyeglasses. However, multifocality often results in a decrease in visual quality. Keeping this in mind, we first decided to evaluate the resolution limits of IOLs at various distances, and we fixed a distance where we achieved the best results using a USAF 1951 chart.

To evaluate the resolution limits of the IOLs, we performed two parts of the experiment. In the first part, we placed spherical trial lenses with powers ranging from +1.0 to -5.0 D with 1.0 D increments in front of the artificial eye model with a scientific camera setup. We captured the results of each IOL for each increment, with a reading chart placed at the correct distance away from the setup. We also repeated the same experiment using a perfect spherical lens instead of an IOL, for comparison. Finally, we calculated the PSF function for each IOL, corresponding to each defocus.

In the second part, we used the same scientific camera setup. However, this time we varied the focal length of the machine vision lens between the scientific camera and the artificial eye model to find the best corrected focus for each IOL. We used a USAF 1951 glass slide resolution target (Edmund Optics, Barrington, NJ) 35 cm away from the setup and captured the results for each IOL. We evaluated the captures for the least achievable contrast limit to resolve two separate lines by human eyes, which we determined to be an 8% contrast ratio limit. The best-achieved resolutions for each IOL were 4.49, 4.00, and 4.00 lp/mm for monofocal, bifocal, and trifocal IOLs, respectively. To better understand the clinical results, we used the viewing distance to convert the linear resolution of the picture to angular resolution, measured using the logMAR test. The conversion results were 0.02, 0.1, and 0.1 logMAR for monofocal, bifocal, and trifocal IOLs, respectively.

Contrast Reduction

The multifocality of IOLs is often associated with reduced image contrast. Although researchers have evaluated contrast reduction in the past, [Ravalico et al., 1993; Mesci et al., 2010; Carreno et al., 2020] it has not been a key feature in objective evaluations on benchtop setups.

To evaluate contrast reduction, we used the captured results of the USAF 1951 glass resolution target. To eliminate the variations, we calculated the contrast ratio in each resolution for separate orientations in 18 intersections and then averaged the results. We performed these calculations starting from Group 0, Element 1 (1.00 lp/mm) to Group 2, Element 3 (5.04 lp/mm) and obtained a contrast ratio for each element of the chart. In these calculations, we used the Michelson contrast:

$$C = \frac{L_{max} - L_{min}}{L_{max} + L_{min}} \quad (3.1)$$

where L_{max} is the maximum value on the white side of the intersection and L_{min} is the minimum value on the black side of the intersection. Since L_{min} cannot be lower than 0, the calculated contrast value could be between 0.0 and 1.0.

Considering external factors and the potential decrease in contrast due to the camera quality, we made the decision to eliminate any factors that could impact the decrease in contrast. For this purpose, without changing any settings of the camera, we captured the same chart without the artificial eye model. We conducted the same evaluations on this capture to obtain contrast ratios and contrast reduction due to setup impurities. We then rounded the best contrast achieved to 1.0 contrast ratio and rounded every other result accordingly. With this, we eliminated the effects on contrast decrease other than the IOL itself.

One of the most common contrast tests conducted in the clinics is Sinewave Grating Test [Ginsburg et al., 2003] where clinicians determine the minimum contrast that patients can distinguish between a grating and a uniform field with a fixed resolution. This resolution is expressed in spatial frequency with units of cpd. To make sense of our contrast decrease test, we converted the resolution to spatial frequency using the viewing distance.

Photic Phenomena

Multifocality can lead to unwanted photic phenomena in the form of halos and glare, which can affect vision. To evaluate the occurrence and characteristics of these phenomena, we used our artificial eye model combined with the scientific camera to capture and characterize them.

We placed the setup on an optical workbench and positioned an LED light source at a sufficient distance. Upon examining the halos produced in the simulator, we noticed that the effect of IOLs varies depending on the color of the light. To account for this we used red, green, and blue lights separately. Without adjusting the camera setting, we then captured the resulting phenomena for each IOL. To ensure objectivity, we avoided capturing saturated results by employing ND filters in the setup.

To ensure accurate comparison, we verified that magnification of the captures was consistent. The setup had a field of view of 25 degrees, which we used in our calculations. We converted the pixel sizes to their corresponding physical sizes and determined the size of the photic phenomena in each capture. We then calculated the radial average and plotted the intensities at their true pixel sizes. To fully characterize the photic phenomena, we applied a function fit to express the effects of IOLs on a point light source as a 2D point spread function for each color, and convolved the selected target scenes used to simulate what patients may see, as shown in Figure 3.3.

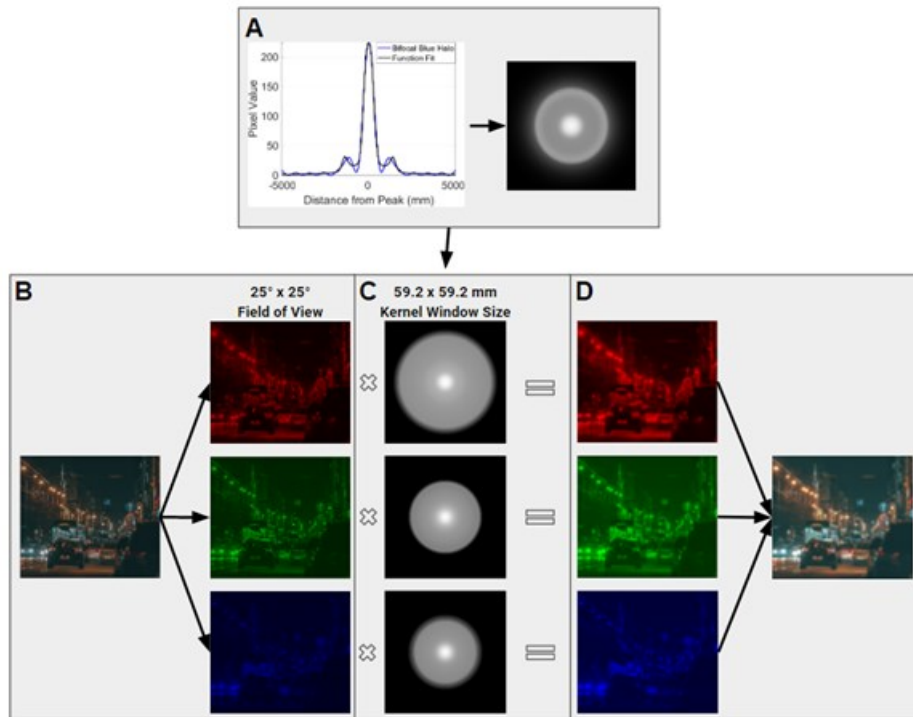


Figure 3.3: Experimental PSF of IOL and simulated image calculation algorithm using the PSF. (A) A function fit and radial averaging applied to experimental PSF cross-sections (e.g., Fig. 8) then the cross-section is converted into a radially symmetric 2D simulated PSF of the corresponding halo pattern; (B) The target scene separated into 3 color channels (red, green, blue); (C) Simulated PSF of the IOL for each color; (D) Results of each channel were combined to achieve a full color scene.

3.3 Results

3.3.1 Captures through Artificial Eye Model and Cataractous Lens

The USAF charts were captured respectively for monofocal, bifocal and trifocal IOLs (Figure 3.4). The contrast difference can be observed between the captured images. An example of cross section for contrast is shown as line AA', and its resulting graphs are presented in parts D, E, F. These USAF chart captures were then used to analyze for contrast decrease and resolution limit in the subsequent parts of the study.

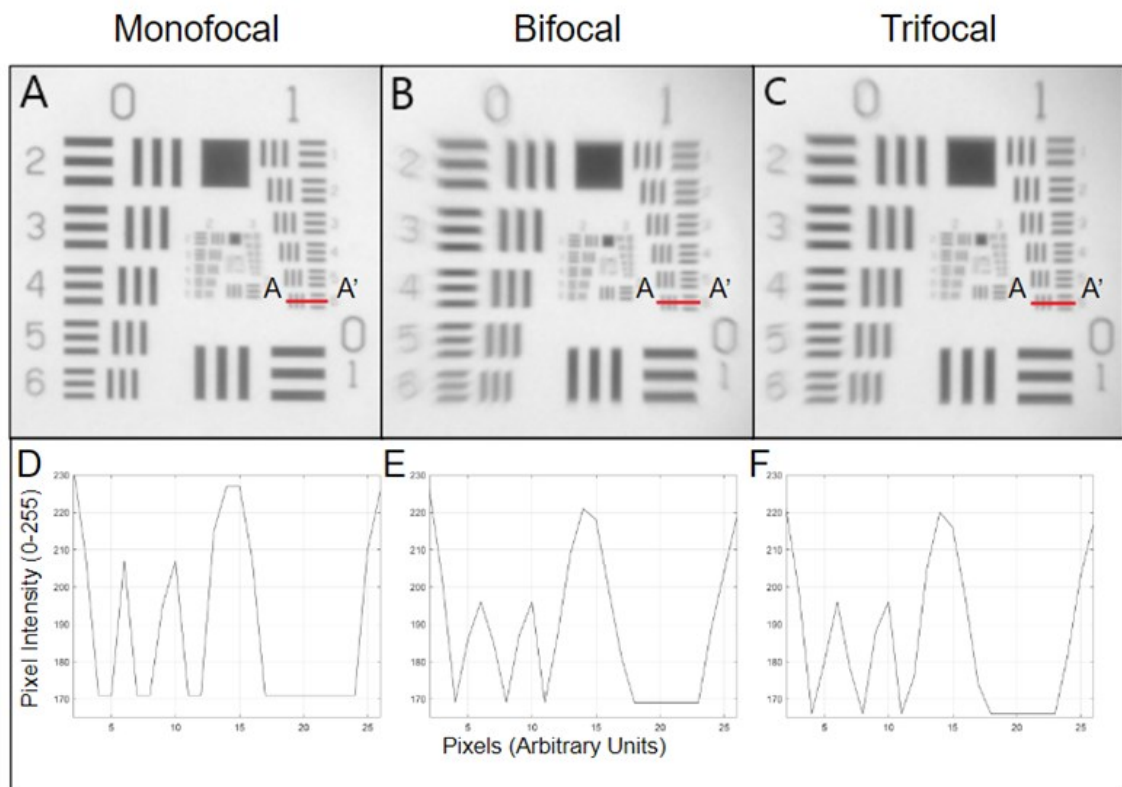


Figure 3.4: Top row: USAF Resolution Charts imaged with the artificial eye model setup at a focus distance of 35cm. Bottom row: Intensity cross-sections (AA'). Monofocal IOL demonstrated better resolution than the bifocal IOL and trifocal IOL.

Figure 3.5 shows the captures of point light sources of different colors (red, green, and blue) with the artificial eye model with monofocal, bifocal, and trifocal IOLs. The field of view of the setup was 25 degrees and the point light source was positioned 50 cm away from the setup. Prominent halos and glares were observed in the captures obtained with bifocal and trifocal IOLs. Notably, the size and magnitude of the halos varied depending on the color of the light sources. The halos produced by red point light sources were larger in size but lower in magnitude than those in green and blue point light sources.

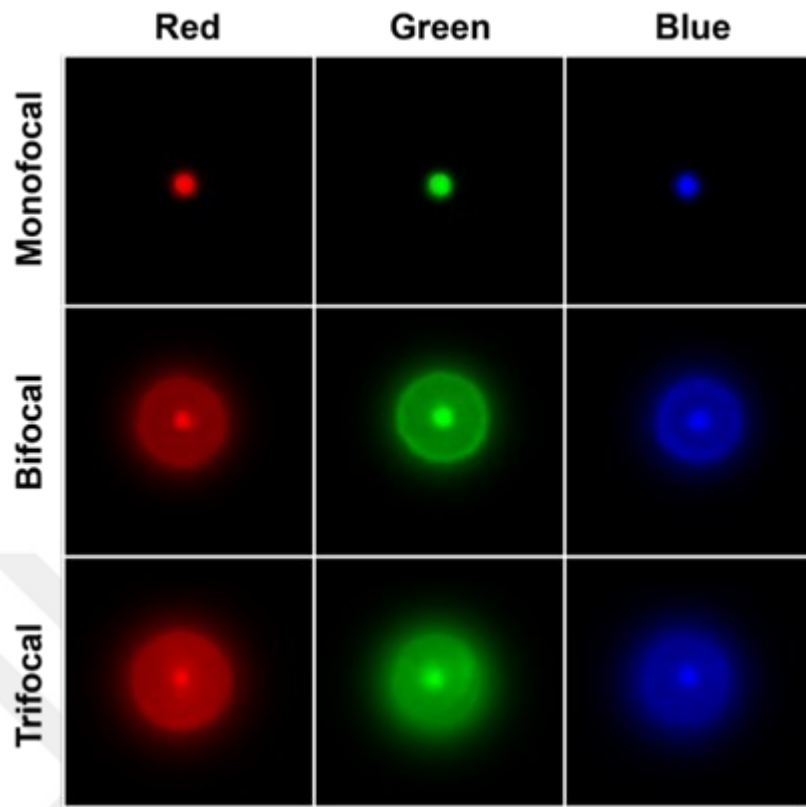


Figure 3.5: LED based point light sources (red, green, and blue) placed at 50cm are captured by the artificial eye imaging system. Significant halos and glares were observed in each color especially for bifocal and trifocal IOLs.

Several scenes were captured through the IOL with opaque surfaces. To simulate different scenarios, an office inside, a night-time courtyard of our university and a reading chart was captured with different opacification levels.

In Figure 3.6, we can see the captures of the office scene. One of the most important aspects of this capture is that we can clearly observe that the transmitted light decreases as the opacification level increases, turning the scene darker. Apart from the imperfections due to the original IOL, blur starts to occur and increase with the opacification. When we pay attention to the lights in the ceiling, even though they are saturated in the first capture with no opacification, as opacification increases, they become saturated and turn yellowish. These results are all known degradations to vision through cataracts.

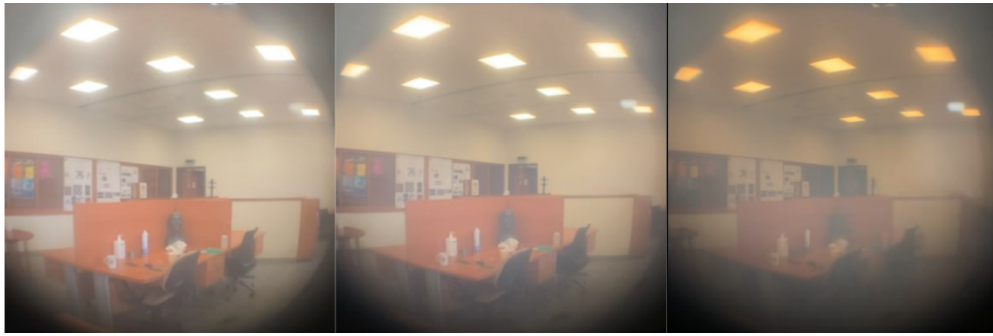


Figure 3.6 Captures of inside scene with lights through IOL with different opacification levels. (a) No opacification (0 seconds), (b) semi-opaque IOL (~120 seconds), (c) nearly fully opaque IOL (~240 seconds).

Figure 3.7 shows the night-time captures of a court in our university through different levels of opacification. Similar results can be observed with the captures of the office scene. The light sources are getting unsaturated, and the scene is becoming blurry and darker with a color shift towards red. Another prominent detail is the scattering of the light. When the edges of the capture observed, we can see that the light coming from the light scene scatters and overflows to sides, creating a glare like phenomena which points out that our opacification not only decreases the transmission of the light but also scatters it. This is another known feature of cataracts, so it further agrees with our simulations.

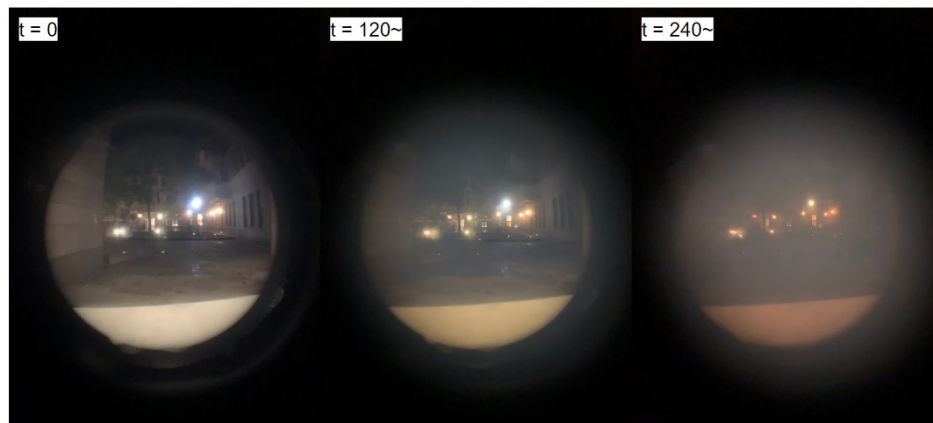


Figure 3.7 Captures of nighttime scene of outside through IOL with different opacification levels. (a) No opacification (0 seconds), (b) semi-opaque IOL (~120 seconds), (c) nearly fully opaque IOL (~240 seconds).

We also captured a sample reading chart from close distance (35 cm) to see how cataracts would affect the reading function of a patient. The write arrow in the first capture of Figure 3.8 indicates the 20/20 vision in close distance. We can see that while without any cataracts the IOL performance barely reaches there, as the opacification starts and increases, the captures taken with 40 seconds intervals show that the visual

acuity dramatically decreases. A more detailed analysis on the resolution of the process is done on the following parts of the thesis.

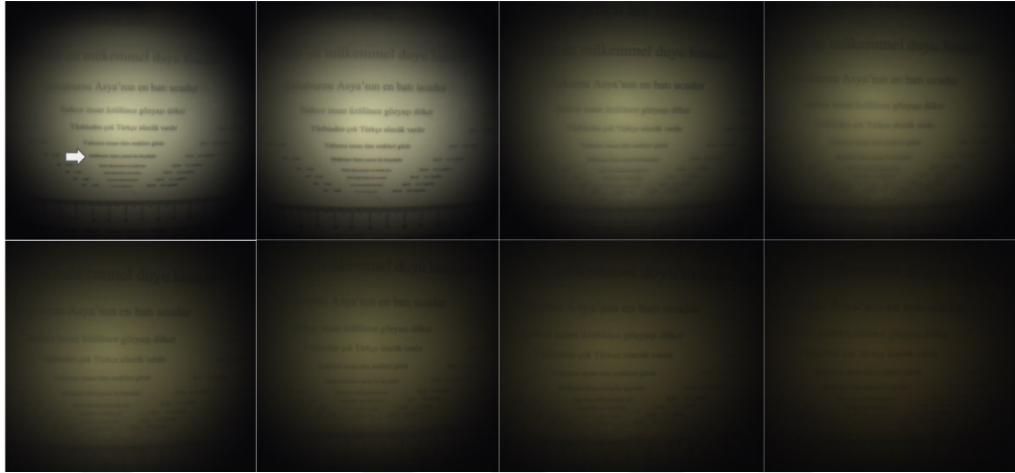


Figure 3.8 Captures of a close distance reading chart through IOL with different opacification levels through 0 seconds to 300 seconds with 40 seconds intervals.

3.3.2 Resolution, Contrast Decrease and Photic Phenomena Characteristics

To simulate the effects of IOLs on vision in each defocus, an artificial eye model, a Snellen reading chart, and a lens set were utilized. We fixed the distance of the Snellen chart to 2 meters and focused the setup according to the lens chosen as 0D, and the same chart was captured with increments of 1D in lens power until 5D defocus. The field of view of the setup was the same as the previous capture test. The PSF of each capture was calculated (Figure 3.9) to be used in simulations. We observed a continuous expansion in PSF for the monofocal IOL. The bifocal IOL PSF showed a slight glare at 0D defocus, which increased in magnitude as the defocus increased, leading to the appearance of a single halo in the PSF. -At 3D defocus, the light was focused resulting in a second focus in the defocus curve. In contrast, the trifocal IOL PSF had a slight glare at 0D defocus, but it showed the appearance of two halos in order as the defocus changed. At -2D and -3D, trifocal IOL PSF had a single and double halo, respectively, and a peak brightness in the center, showing that it has a focus in these defocus levels. The analysis of PSFs allowed us to identify the unique characteristics of each IOL in terms of resolution, contrast reduction, and photic phenomena.

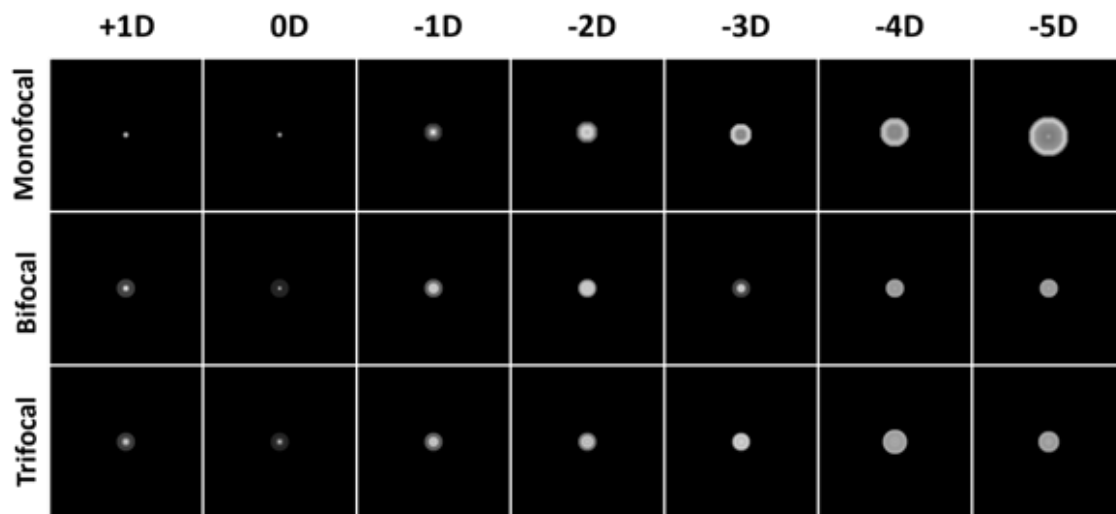


Figure 3.9: Defocus PSFs of each IOL with 1D increments from 1D to -5D. For each IOL, Snellen charts were captured using the artificial eye model. The resulting PSF functions point that monofocal IOL has a constantly decreasing focus starting from 0D towards -5D. . Bifocal IOL and trifocal IOL regain their focus on -3D and -2D respectively.

Apart from the defocus effects of IOLs on vision, we also aimed to characterize the contrast decrease. The contrast decrease in each IOL as a function of spatial frequency is calculated (see Figure 3.10). The monofocal IOL has its best contrast (95%) around 5 cpd spatial frequency, and the contrast decreases as the spatial frequency increases, reaching nearly 0, around 32 cpd of spatial frequency. The contrast pattern for bifocal and trifocal IOL was like that of the monofocal IOL but with lower starting contrasts at 5 cpd spatial frequency. Bifocal IOL starts with slightly better contrast (68%) than trifocal IOL (65%). As the spatial frequency increases both IOLs showed a decreasing trend and occasionally passed each other. The contrast lines for all IOLs crossed the 8% contrast limit chosen at 27, 24.7, and 24.5 cpd for monofocal, bifocal, and trifocal IOLs, respectively, which means that none of them reached 20/20 vision within this limit.

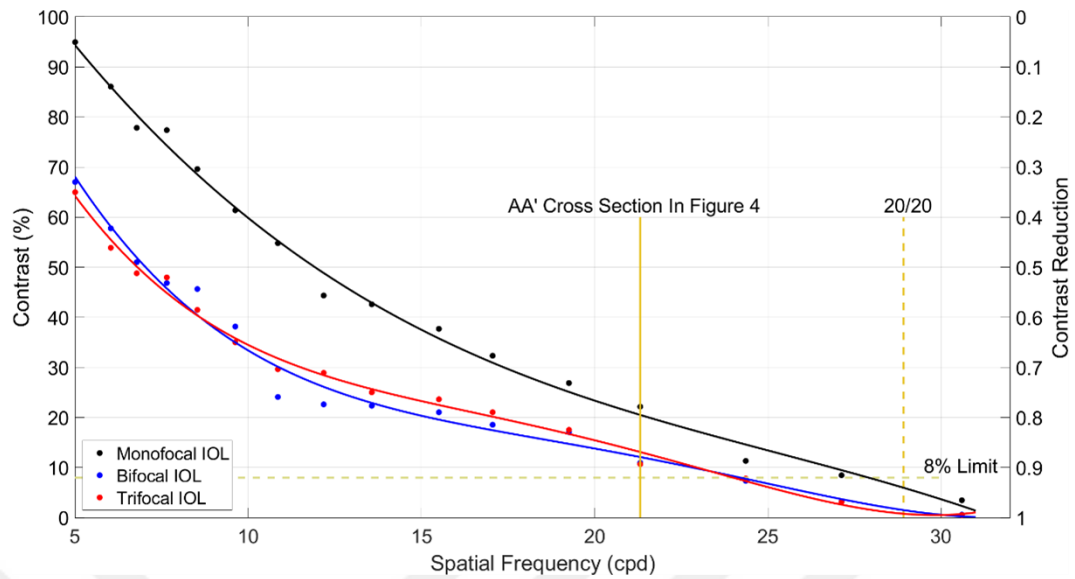


Figure 3.10: Contrast percentage curves of using the USAF charts in Fig.4 for different IOLs. Bifocal and trifocal IOLs show a similar curve and less contrast compared to monofocal IOL. Using an 8% contrast limit for human eyes, monofocal reaches 27.5 cpd while bifocal and trifocal IOLs reach 24.5 cpd in spatial frequency.

As the last step of the characterization, point light source captures are plotted (Figure 3.11). Each plot is averaged radially to eliminate the artifacts and aberrations of the setup, assuming each light source is a perfect circle. The monofocal IOL plots exhibit visible center peaks as a normal distribution, with the size of the peaks gradually decreasing as the color shifts from red to blue. Bifocal IOL plots show prominent halo peaks and glare apart from the center peak, with the magnitude of the halo increasing as the color shifts from red to blue, and the size of the glare decreasing. Specifically, the red-light source has a wider halo with less magnitude, while the blue light source has a narrower halo with more magnitude, and the green light source lies in between them. In trifocal IOL plots, glare is more prominent, while halos have a smaller magnitude. Again, the red-light source shows a broader photic phenomenon and less halo magnitude than the other colors. PSFs were calculated according to these halo charts to be used in photic phenomena simulations.

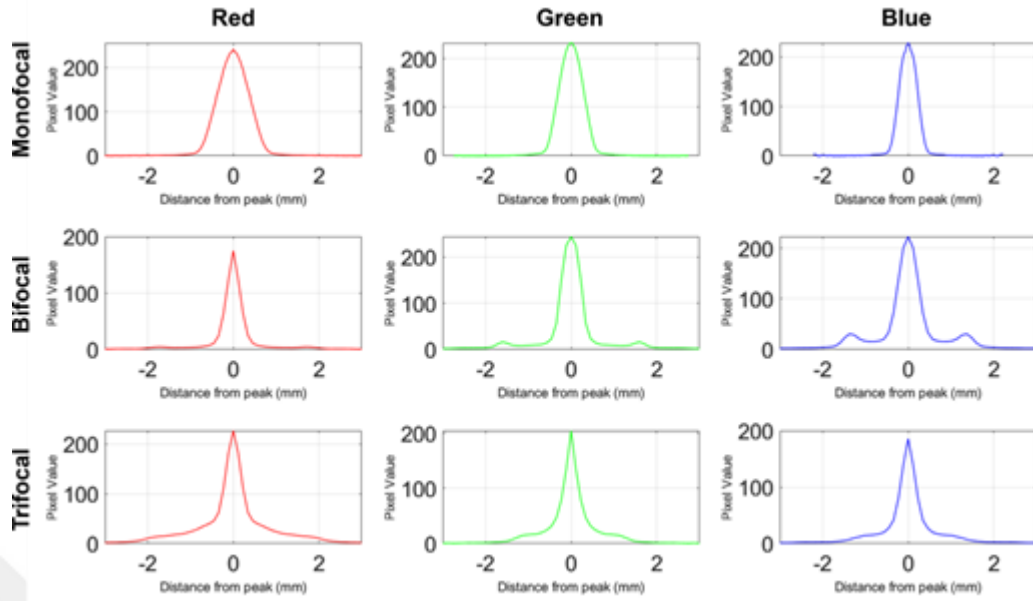


Figure 3.11: PSF cross-sections for red, green, and blue point light sources (shown in source plane coordinates at 50 cm). Monofocal IOL shows an expected peak at the center with no glare or halos. Bifocal IOL has prominent halos as small peaks around the main peak with little glare. Trifocal IOL tends to have more glare than a prominent halo. Red color in each case shows a wider peak compared to other colors (blue and green).

3.3.3 Vision Quality through Artificial Cataract

Several characterizations were done to objectively understand the opacification process and how it degraded the vision through IOL. In the first characterization, to understand how the resolution limit is affected by the opacification, a slanted edge was captured through the artificial eye model implanted with opaque cataract. The captures were taken 100 mm away from the eye model and the camera was 8.7 mm behind it. Knowing the pixel pitch of the camera, which is 0.00345 mm, we converted spatial frequency unit of cycles per pixel to cycles per degree to understand the graphs clearly. The conversion was done by the following equations. We first converted spatial frequency from cycles per pixel to cycles per millimeter with the following equation.

$$MTF\left(\frac{cyc}{mm}\right) = \frac{MTF\left(\frac{cyc}{pixel}\right)}{Pixel\ Pitch} \quad (3.2)$$

Then, for the other steps, we needed to calculate some optical parameters of the setup to use them in the conversion. We first calculated the magnification:

$$M = \frac{s_1}{s_2} \quad (3.2)$$

Where M is the magnification of the setup, s_1 is the distance between the slanted edge and the eye model, which is 100 mm, and s_2 is the distance between the eye model and the camera, which is 8.7 mm. Conducting this calculation, M was found to be 0.0087. Then, the focal length of the setup was calculated as follows:

$$FL(mm) = \frac{s_1}{1 + \frac{1}{|M|}} \quad (3.2)$$

Where $FL(mm)$ represents the focal length of the setup in units of millimeters which is 8 mm. After finding the necessary parameters of the setup, we finally converted the MTF unit to what we desire.

$$MTF (cpd) = \left(\frac{\pi}{180} \right) MTF \left(\frac{cyc}{mm} \right) FL(mm) \quad (3.2)$$

After being able to convert the spatial frequency, we plotted our finding on Figure 3.12. Thanks to the time dependence of the opacification, each capture represented different levels of opacification. At $t = 0$, there is no opacification on the IOL yet, and we can observe that its resolution reaches around 31 cpd, which is close to 20/20 visual acuity in a Snellen chart, when MTF criteria is chosen as 0.1. As the opacification increases, with 60 seconds intervals, we can see that the achievable resolution decreases to 26, 20, 14, 10 and finally 5 cpd respectively. Since 5 cpd is already a very low-quality resolution, the test wasn't proceeded after 300 seconds. If the trend of resolution decrease is evaluated, we can say that there is nearly a linear decrease as a function of time, after 180 seconds, the quality of vision becomes very low.

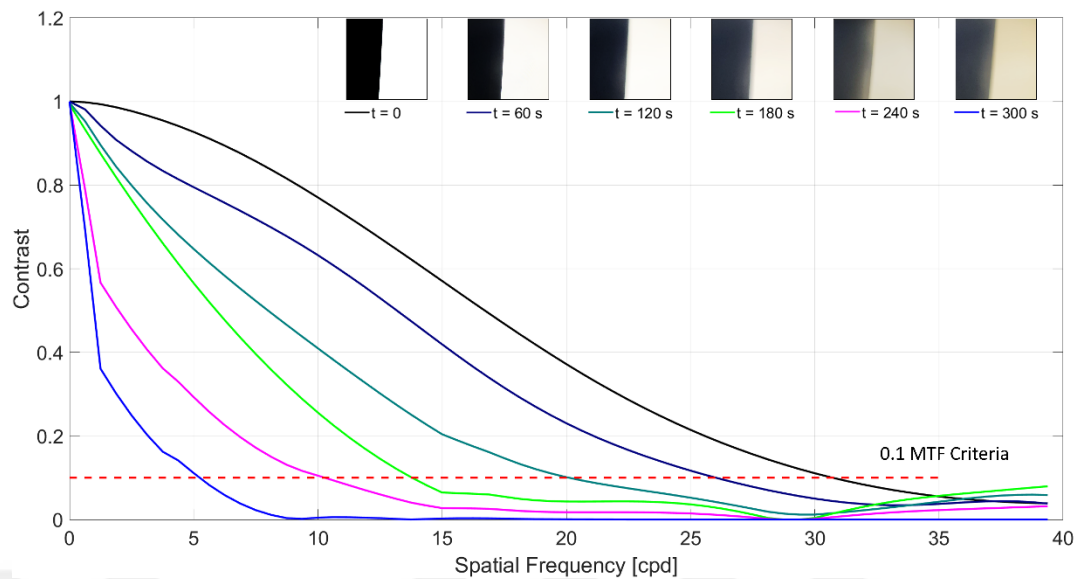


Figure 3.12: MTF functions of IOL with different artificial opacification levels.

One of the biggest known aspects of cataracts or PCO is the scattering surface. First verify and understand the scattering in our cataract mimicking, we utilized a photodiode and measured the transmission through the IOL with opacification process as a function of time. In the second step, we put an aperture in front of the photodiode and placed it such that it is at the focal plane of the IOL. This way, we eliminated the scattered light and measure how much of the transmitted light still stays on the optical path. For the light source, we used an LED light source with 530 nm peak wavelength and collimated it with an ideal lens to a beam of 2 mm diameter. The light source input power was 0.7 mW.

The transmitted light and the percentage of how much of it is scattered is plotted in Figure 3.13. As can be observed, the transmission starts around 0.6 mW and decreases to 0.2 mW at the end of 300 seconds. Whereas scattering starts around 10 percent and goes up to 50% with the passing time.

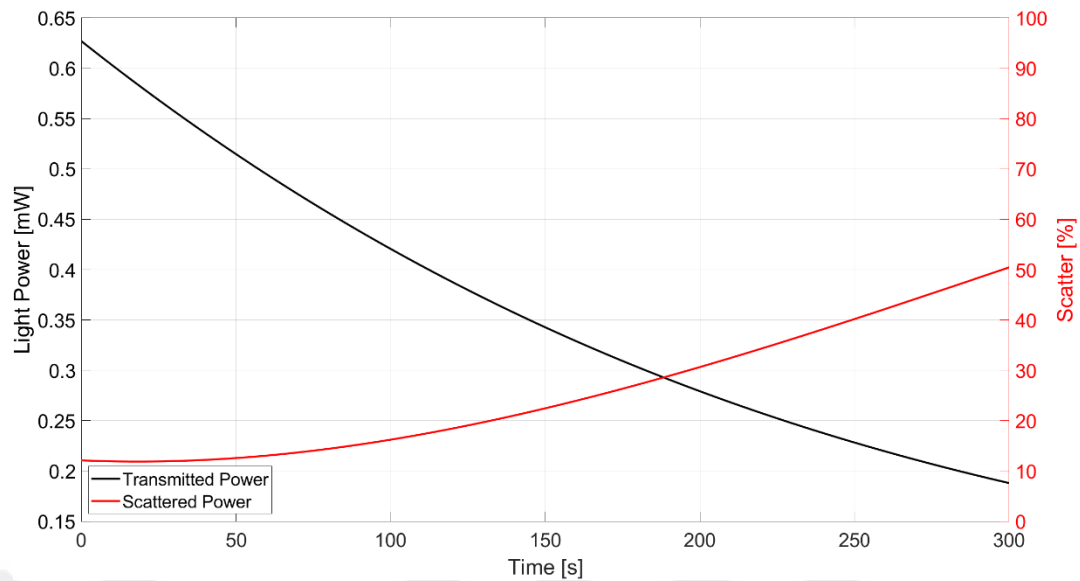


Figure 3.13: Transmission and scattering of artificial cataractous IOL with different opacification levels with respect to time.

Although the characterization done by the photodiode was very informing and gave us great insight into the behavior of opacification on IOL surface, it was limited to 530 nm wavelengths. In the next characterization, we utilized a spectroscopy and a light source having a spectrum from 200 nm to 1000 nm to see how opacification affects different wavelengths. Once again, with an interval of 60 seconds, spectroscopy measurements were on IOL with proceeding opacification and plotted on Figure 3.14. It was seen that, in all wavelengths, a similar degradation could be observed. The important aspect of these results was that some wavelength intensities decreased more compared to the others. For example, when the wavelengths around 700 nm are observed, we see that at start the intensity count is around 2800, and it decreases down to 1200. This gives a 57% decrease in intensity. When we observe the intensity count around 600 or 800 nm wavelengths, we see that the starting point is around 1500. Then as opacification increases, this value decreases to 500. This shows a decrease around 33%, telling us that the peak around 700 nm is exposed to the highest intensity loss and as we get away from that area, the percentage loss decreases.

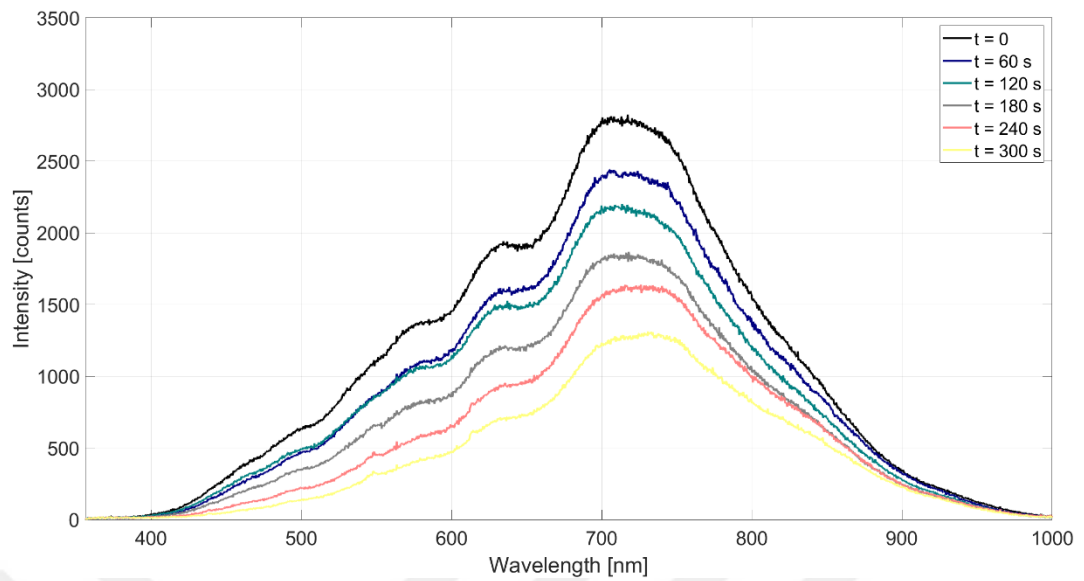


Figure 3.14: Spectroscopy measurement results of artificial cataractous IOL with different opacification levels.

The effect of opacification on different wavelengths was similar in the previous graph. We wanted to compare the behavior of wavelength shift during opacification. For this purpose, we only looked at the measurements at the start and the end. For an objective comparison, we normalized the intensity of the last measurement according to the first measurement and plotted on Figure 3.15. This showed us that as the opacification increases, there is a slight shift to higher wavelengths when we compare the peaks. No expansion around any wavelength related to the opacification was observed.

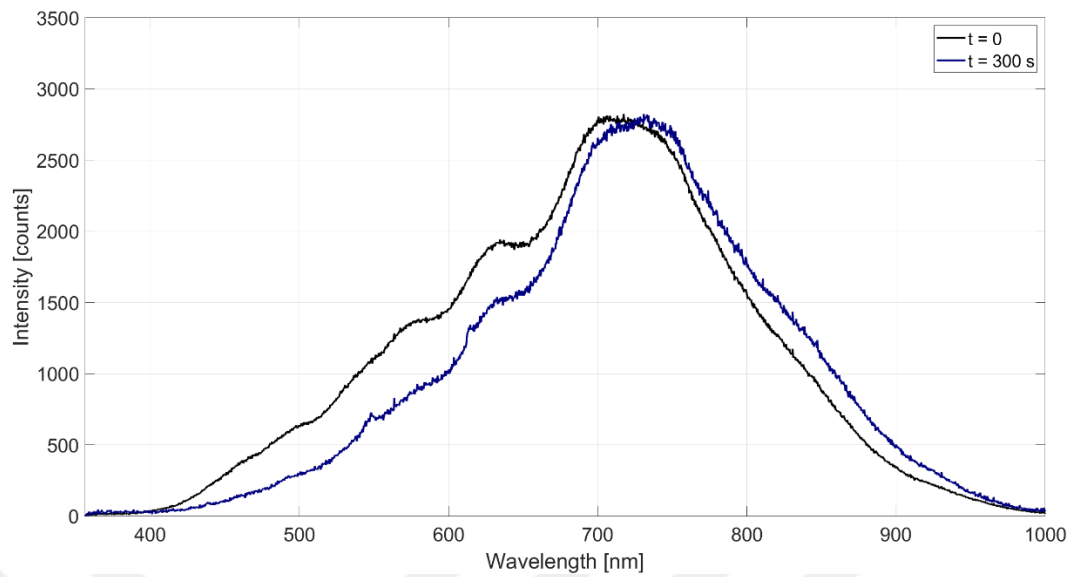


Figure 3.15: Spectroscopy measurements of cataractous IOL with two different levels of opacification. The spectroscopy measurement at opacification corresponding to 300 s is normalized to 0 s measurements for comparison.

On the final part of the opacification process characterization, rather than a measurement, we conducted color histogram analysis on captures taken at different opacification levels shown in Figure 3.16. Three captures at 0, 120, and 240 seconds of opacification were analyzed to see how the color intensities were distributed. One general conclusion was that, as the opacification increased, the pixels shifted towards lower intensity values. A weighted average of each color's percentage for each capture was also calculated. As seen, as the opacification increased, pixels tended to become more red and less blue, as expected. The percentage increase in green was negligible compared to the other colors. The pixels especially peaked around a pixel intensity of 100 at 240 seconds of soaking, losing nearly half of the initial intensity at colors blue and green and losing relatively less intensity in red color.

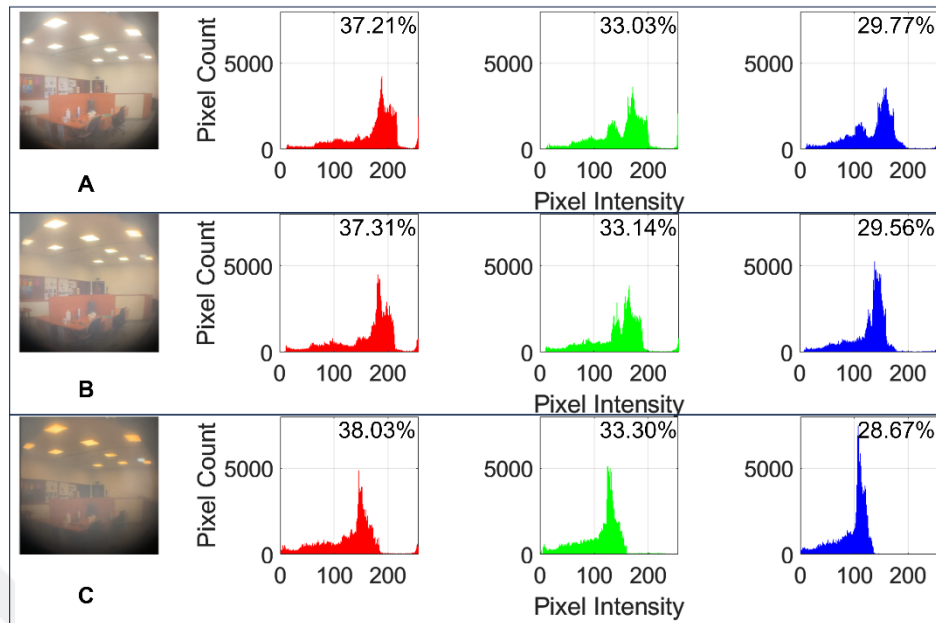


Figure 3.16: Color histogram results of images captured by IOLs with different amounts of opacification.

3.4 Discussion

With this study, we investigated the possibility of characterizing various effects of IOLs on vision. We saw that we could successfully characterize different IOL types and get expected characterizations from our opacification process. However, some limitations still existed with this approach.

The use of scientific cameras had negative effects on finding the actual PSF of the IOLs in a human eye. A camera has a very limited dynamic range compared to the human eye and is not able to capture all the details. In this study, the captures for characterization were taken in 12-bit raw images to minimize this limitation. This was enough for the purpose of estimating the visual results, but for a more accurate characterization, a wavefront sensor instead of a camera could be used with the same methods [Epitropoulos et al., 2018]. Regardless, the PSF functions obtained were like those obtained in previous studies [Pieh et al., 2002; Epitropoulos et al., 2018].

An additional noteworthy limitation encountered in our study pertained to the saturation levels observed in the initial image captures, which were subjected to PSF convolutions. Despite the digital single-lens reflex camera employed possessing a dynamic range of 16-bit, the acquisition of completely unsaturated nighttime scenes was

unattainable due to the persisting scientific research problem regarding the capture of unsaturated nighttime images with lights [Karadeniz et al., 2021] and 16-bit being still very limited compared to the dynamic range of the human eye [Hoefflinger, 2007]. This inherent limitation led to preexisting glares present in the original captures prior to convolution, thereby inadequately capturing the comprehensive photic phenomena induced by IOLs in human vision.

Regardless of these limitations, characterization of IOLs through an artificial eye model is a promising method for assessing IOL performance. In the next chapter, we utilize these characterizations to show subjects how they would see the world after IOL implantation of their choice.

In addition to IOL characterization, we tried to mimic the progression of cataracts on the human lens and how vision degrades correspondingly. A simple method utilizing acetone and IOL to create a scattering surface (opaqueness on IOL) was developed for this purpose. Previous investigations replicated this opacification physically or digitally, using virtual reality, augmented reality, or optical setups. However, this opacification has never been replicated as progress in virtual simulations, and the single physical example was an irreversible process that would not be recreated on the same substance.

The advantage of this method over other methods is that it is a reversible process, so the opacification can be reversed at any time to repeat and observe the effects on vision. Furthermore, it clearly shows the progression of the opacification of the lens on vision. However, this process wears out the IOL as it is repeated, so the repetition limit is up to 20 times before the IOL starts to break apart. Another advantage of this method is that by partially applying the acetone to the opaque surface of the IOL, a cataract-like pattern can be created to simulate an even more realistic cataract case. Depending on how long the acetone is applied to the surface, while some areas have dense cataracts, others can have mild or no cataracts. Moreover, applying this method to selected surfaces of the IOL makes it possible to mimic different cataract or opacification cases and their progression.

All analyses showed the expected results. While the IOL without cataracts could show satisfactory visual performance by itself, as it was soaked in water after the acetone treatment and the opacification on its surfaces increased, there was a significant downgrade in its performance. The IOL choice was monofocal, which does not perfectly represent the human crystalline lens in multi-depth scenarios. However, the

primary concern of this study was to observe and understand the scattering and performance decrease overall, so it was not a significant issue.

Another unique point of this method was recreating the opacification process in a patterned sense. While there could be several options, such as masking the surface and treating only selected parts with acetone, cleaning the already opacified surface areas with alcohol was more practical. The impracticality of the masking lay in the small surface area, which would require very delicate work, causing our method to stray away from being simple.

We must mention one major limitation of our method. Since the chemicals are only applied to the surface, and the opacification occurs on the surface, it does not physically represent natural cataracts, which occur inside the crystalline lens rather than just on its surface. This limitation did not affect the results much since the analyses showed that our method could successfully mimic cataract progression when the resulting visual qualities were observed.

An integral aspect of our study pertains to its potential clinical implications. With cataracts and PCO typically occurring slowly and patients often becoming aware of their condition only in severe cases, our approach holds promise as an educational tool within the realm of ophthalmology. This method could facilitate early diagnoses by demonstrating the progressive impact of opacification on vision, allowing for timely interventions that prevent a significant decline in patients' quality of life.

Chapter 4:

**HOLOGRAPHIC DISPLAY BASED INTRAOCULAR LENS
SIMULATOR****4.1 Introduction**

There are some IOL simulators to show vision through IOLs to cataract patients prior to surgery. However, they do not work well for patients with moderate or dense cataractous lenses due to impaired vision and heavy scattering. One such case is an IOL simulator that uses a biconcave lens back-to-back with an IOL placed in a thin wet chamber. The compact lens assembly has a limited field-of-view but allows patients to see the world through a multifocal intraocular lens [Na et al., 2022]. Similarly, the Rassow telescope follows the same architecture as the intraocular lens simulator but utilizes a convex lens [Wahl et al., 2019]. VirtIOL (10Lens S.L.U., Barcelona, Spain) is another device for simulating vision through intraocular lenses [Studený et al., 2018]. It allows the subject to obtain the defocus curves of patients as if they were wearing the IOL of their choice. SimVis (2EYEVISION, Madrid, Spain), which prescribes multifocal corrections by utilizing the temporal multiplexing of the lens power [Akondi et al., 2018]. VAO (VOPTICA, Madrid, Spain) is another device with integrated adaptive optics for simulating multifocal IOLs with various measurements and functionalities such as ocular aberrations, subjective refraction, and visual tests [Hervella et al., 2019].

In this study, we combine a holographic vision simulator developed in our group and physical IOL measurements using an artificial eye model. Our approach enables us to preoperatively estimate the photic phenomena and side effects that patients might experience after IOL implantation and characterize these visual functions using the artificial eye model. To validate our approach, we tested and compared visual acuity, contrast sensitivity, and halo severity through the artificial eye model and simulated the results using phase holograms. Our study suggests that our approach has the potential to provide patients with a more realistic and personalized understanding of the effects of cataract surgery and IOL implantation on their vision.

4.2 *Methods*

4.2.1 *Holographic Display*

Holographic displays offer unique capabilities for direct view displays as well as virtual and augmented reality applications. Holography is the only technology that can create virtual images with all the 3D depth cues and create programmable exit pupil patterns. Our group exploited those properties to develop a cataract simulator for patients mainly mimicking the performance of monofocal lenses [Kavakli et al., 2021].

In this study, we use a benchtop holographic display setup to show subjects computer generated holograms of visual stimuli. After characterizing the IOLs, target image was first convoluted using the measured PSF for each IOL at different wavelengths. We used the following target images for different tests: Sinewave grating contrast chart, ETDRS chart, and real-life photographs. Then, we computed the phase holograms using CGH algorithms. Following the content creation, CGH calculations are done through focus plane discretization, object wave computation, and 3D picture reconstruction. Once the content is discretized into different focus planes based on depth map values, the object wave on the scene plane is computed using Fresnel Space Propagation with respect to the optical parameters of the display setup. As the final step, the complex-valued object wave is computed. The final product is encoded CGH, replicating the target content. We utilized a holographic display setup discussed below to show it to patients.

Figure 4.1 shows the optical architecture consisting of a point light source, a phase only SLM (Holoeye, Berlin, Germany), and optical components. A 125 mm focus convex lens (Thorlabs, Newton, NJ) collimates the light coming from the point light source, which then illuminates the SLM. The operating wavelengths are 470 nm for blue, 530 nm for green, and 632 nm for red light. The SLM modulates the light with a resolution of FHD (1920x1080 pixels) and a pixel pitch of 8 micrometers. The propagated light beam then goes through a 4f-lens system consisting of two lenses (Thorlabs, Newton, NJ) of 100 mm and 50 mm focal length, respectively, and an aperture. Finally, the modulated and magnified image reaches the patient's eye.

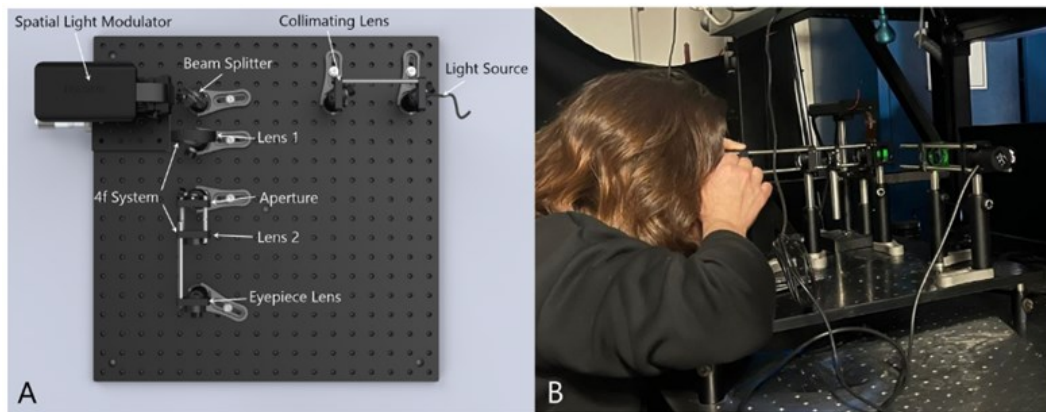


Figure 4.1: The visual acuity and contrast sensitivity tests were conducted on a proof-of-concept benchtop setup. (A) 3D rendering and optical components of the setup . (B) A real photograph of the holographic setup during the user tests.

4.2.2 Patient Tests

The study was conducted in accordance with the institutional guidelines and was approved by the Koç University Ethics Committee (Starting from 23.12.2022, 2019.400.IRB2.121) on human subject research, following the principles of the Declaration of Helsinki. The purpose and methodology of the study were described to the subjects prior to testing, and informed permission was acquired. A total of 10 subjects (ages 23-65) were recruited from personnel and students at Koç University, with care taken to ensure a diverse sample for unbiased results.

The holographic display simulator was used in the test setup. To minimize bias, the contents for the test were randomized for a double-blind experiment. The following tests were conducted: defocus curve, contrast sensitivity and two-point resolution limit. Subjects were also shown light sources for them to experience halos and glare.

In contrast to the artificial eye setup, where physical IOLs were changed for testing, in the holographic setup, the effects were applied to the holographic content. Moreover, IOL effects applied to real-world scenes were displayed as holograms to subjects and they were asked to rate their satisfaction. The details of the tests are as described below.

Visual Acuity and Defocus Curve Tests

We first conducted a visual acuity test using the holographic setup to simulate IOL effects. Subjects viewed an ETDRS chart displayed as holographic content with correct size. Visual acuity was calculated according to standard logMAR rules. To simulate different defocus levels, we applied IOL effects as PSFs to the holographic content, equivalent to placing a spherical trial lens with power ranging from +1.0 to -5.0 D in

front of the artificial eye model, with 1.0 D increments. For each IOL and defocus level, we applied the corresponding PSF to the holographic content. To verify the accuracy of our holographic visual acuity measurements, we also measured the best corrected visual acuity of three subjects and found that the results matched with their holographic results.

Contrast Sensitivity Tests

The holographic setup was also used to conduct the contrast sensitivity test. In this test, no physical IOL or screen was used. Instead, holographic content was presented to the patient as if they were viewing a screen placed 3 meters away. The holographic content contained the effects of IOL and was propagated at the correct magnification so that subjects perceived the content as if it was 3 meters away from them. Sine-wave gratings with spatial frequencies 3, 6, 9, 12 and 18 cpd were shown to subjects as holographic contents. Each spatial frequency content had 5 samples of contrast thresholds, starting from higher contrasts, and gradually decreasing to lower contrasts calculated by Michelson's formula [Kukkonen et al., 1993]. Subjects were asked to tell the orientation of the gratings, which were either slightly tilted 15 degrees in a clockwise or counterclockwise direction, or not tilted (vertical straight lines). Each answer was recorded, and the minimum contrast threshold was found for each spatial frequency. Contrast sensitivity was calculated by taking the reciprocal of the contrast thresholds. This test is conducted for each IOL, and three different graphs are obtained. The contrast threshold levels were chosen according to the routine contrast sensitivity tests done in Koç University Ophthalmology Clinics.

Photic Phenomena and Two-Point Resolution Tests

The halo test conducted with subjects had two parts. In the first part, we showed the subjects holograms of PSF-applied content for each IOL consisting of two-night scenes and a daylight scene in traffic. Since most dissatisfaction due to halos and glares occurred in traffic and car lights, we decided it was crucial to patients these types of scenes. After showing the holographic contents, we asked subjects to rate their satisfaction with such a vision on a scale of 1 to 5, where one meant very dissatisfactory and five meant very satisfactory.

In the second part of the test, we aimed to objectively understand and quantify this photic phenomenon. Quantifying the halos and glare occurring due to the multifocality of IOLs is still an ongoing research topic. Although several attempts have been made

lately to find an objective metric for halo severity, their validations were still highly subjective since they relied on feedback from patients [Ukai et al., 2021; Carson et al., 2018]. To address the subjectivity of this quantification, we decided to apply a two-point resolution test to subjects through the artificial eye model. By relating the minimum distance of two light sources the subjects could distinguish to our halo characterization, we investigated the efficiency of IOLs. The metrics we focused on in our characterization were the distance of peaks of the halos from the source and the magnitude of these peaks relative to the center peak.

The test setup for the two-point resolution test consisted of the artificial eye model, a point light source, and a mirror. The point light source and the mirror were placed 1 meter from the artificial eye model. We positioned the mirror so that subjects could see two light sources when they looked through the artificial eye model. The mirror was initially placed 20 mm away from the light source, so the distance between the two sources was 40 mm for subjects. We systematically decreased the distance between the light source and the mirror with a step size of 0.01 mm until the subjects could no longer distinguish two sources. This distance was noted, and the test was repeated for each IOL.

4.2.3 Statistical Analysis

We used the Kolmogorov-Smirnov test to analyze patient information (age, spherical equivalent, cylindrical error) and test results (contrast sensitivity, 2-point resolution limit, vision satisfaction). If the data was normally distributed, the mean $\pm$ standard deviation was used, and data was compared in pairs with paired t-test. If the data was not normally distributed, the median was used (interquartile range) and data was compared in pairs with Wilcoxon signed rank test.

We conducted the Kolmogorov-Smirnov test to compare the distant visual acuity of different IOLs at vergences 0D, -2D and -3D. The test showed that none of the data was normally distributed at 0D vergence ($P = 0.0039$, $P = 0.0039$, and $P = 0.0031$ for monofocal, bifocal, and trifocal IOLs respectively). Similarly, the data was not normally distributed, at vergences -2D or -3D ($P = 0.0016$, $P = 0.0006$, $P = 0.0025$ for -2D and $P = 0.0004$, $P = 0.0025$, $P = 0.0025$ for monofocal, bifocal, trifocal IOLs respectively). Therefore, we used the Wilcoxon signed rank test to compare data between different

IOL pairs. We used MATLAB R2022a for statistical analysis, and a significant level of 5% was considered.

4.3 Results

4.3.1 Applied Results in Target Images and Holography

The contrast decrease that was characterized for each IOL was applied to sine-wave gratings that corresponds to their respective spatial frequencies. An example of the contrast decrease applied to 3 cpd content is shown in Figure 4.2. The first column represents the target content for the holography for each IOL. The second column is the real captures of holograms generated with the targets in the first column. The last column is for validation purposes, allowing us to understand if the contrast decreases match the expected results. We initially generated holograms of the same content without any contrast decrease and captured them with the artificial eye model to determine how much a physical IOL decreased the contrast of the hologram.

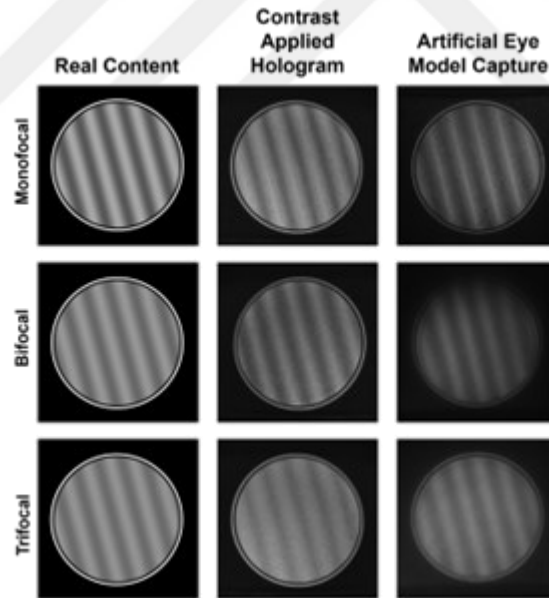


Figure 4.2: Contrast sensitivity sinewave grating test charts with 3 cpd spatial frequency. Column1: Prepared content to mimic the contrast decrease of each IOL. Column2: Corresponding captured images in the holographic setup. Column 3: Standard test patterns (with 100% modulation) imaged through the artificial eye model using each IOL. Comparing the images in Column2 and Column3, we can claim virtual simulators using holography closely mimic the contrast reduction of the physical IOLs.

The contrast values calculated using Michelson's formula are given in Table 4.1. The results are very close to each other, but we see a slight underestimation in simulated

holographic contrast captures compared to the captures of standard test contents through the artificial eye model. After some discussion with ophthalmologists, it was concluded that this difference was not significant for clinical applications.

Table 4.1: Contrasts in target contents and holographic results through different IOLs. Column 1 shows the contrast percentage in prepared content to mimic IOLs in holography. Column 2 shows the contrasts of corresponding captured images in holographic setup. In column 3, contrasts of captures of the standard tests pattern through the artificial eye model are given to verify simulation results.

IOL Type	Prepared Content Contrast (%)	Captured Image Contrast (%)	Holography through Artificial Eye Contrast (%)
Monofocal IOL	32.4	32.0	33.1
Bifocal IOL	20.2	20.4	20.9
Trifocal IOL	16.0	15.8	16.0

IOL, Intraocular lens.

Sinewave grating contents were captured at various spatial frequencies to determine if the generated holograms were correctly showing contrast reduction (Figure 4.3). All holograms were captured using a scientific camera, the same as the characterization setup. The first column is the hologram captures generated with a target content corresponding to the standard test. The rest of the columns show real hologram captures which had simulated contrast reduction in their target contents, respectively for monofocal, bifocal, and trifocal IOLs.

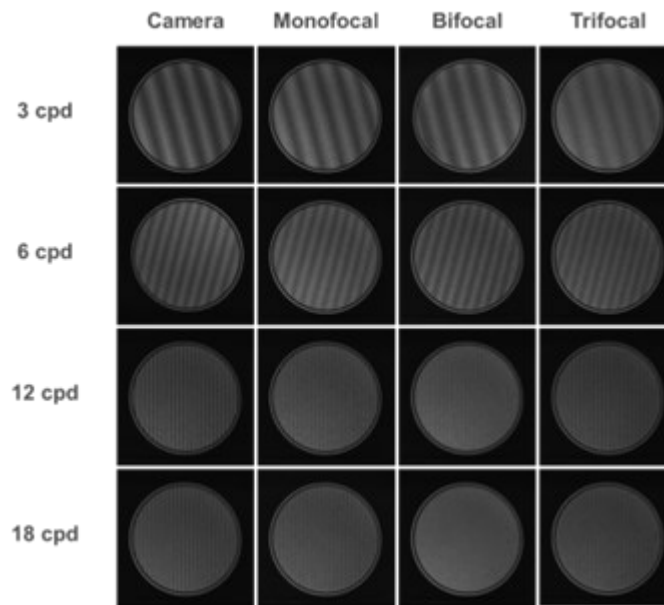


Figure 4.3: Captured images in the holographic setup using different spatial frequency fringe-based holograms computed to mimic contrast levels of different IOLs. 3 cpd row is the same as 1st column of Fig. 9. Column 1 corresponds to using a camera lens (not an IOL). Columns 2 through 4 correspond to holograms with reduced contrast corresponding to different IOLs. Looking at the contrast reductions, we can confirm that results match with contrast characterization made with the artificial eye model for all spatial frequencies.

Calculated contrasts of the captures in Figure 10 are given in Table 4.2. Monofocal IOL had 5, 23, 47, and 52 percent contrast decrease at 3, 6, 12, and 18 cpd spatial frequencies respectively compared to the expected contrast from the test. In the same contents, bifocal IOL had 38, 47, 71 and 79 percent contrast decrease and trifocal IOL had 35, 49, 67, and 77 percent contrast decrease. Compared to the characterization done with the artificial eye model setup (Figure 3.10) for real scenes, we observe a slight but clinically insignificant underestimation in contrast decrease calculated through holography simulations.

Table 4.2: Contrasts of IOL simulated contents captured in holographic setup at different spatial frequencies based on Figure 10 and their contrast reduction relative to standard test chart holograms.

Spatial Frequency (cpd)	Real Contrast (%)	Monofocal Contrast (%)	Monofocal Contrast Reduction (%)	Bifocal Contrast (%)	Bifocal Contrast Reduction (%)	Trifocal Contrast (%)	Trifocal Contrast Reduction (%)
3	32.7	31.0	5	20.2	38	21.2	35
6	24.3	18.9	23	12.9	47	12.4	49
12	39.0	20.9	47	11.2	71	12.9	67
18	48.5	18.9	52	10.6	79	11.5	77

Cpd, cycles per degree.

The resulting scenes obtained through convolution with the IOL PSFs found by the artificial eye model are given in Figure 4.4. In the day scene, no halo or glare are observed in any IOL, but some blur and contrast decrease are seen in bifocal and trifocal IOL compared to monofocal IOL. In both night scenes, in addition to blur and contrast decrease, bifocal and trifocal IOL PSFs add extra halos and glare, resulting in a visible dissatisfaction for the patients.



Figure 4.4: IOL PSF applied scenes through convolution using the algorithm in Figure 3.2. One daytime and two night-time scenes were used. The scenes were blurry for bifocal and trifocal IOLs, and more intense halos and glares were observed compared to monofocal IOL especially in the night-scene.

Simulation of one of the IOL PSF applied scenes is given in Figure 4.5. Although additional contrast decrease is observed in the results. The targeted effects such as blur and halos are prominent when IOLs are compared to one another.



Figure 4.5: Full colored simulations of IOL PSF applied scenes. We see more blur and visible halos in multifocal IOLs compared to monofocal IOL.

4.3.2 Validation Tests for Holographic Setup

Thirteen eyes from 13 subjects were included in this study. Relevant details of the subjects were given in Table 4.3. Median age of the subjects was 25.0, and gender distribution was 11 men to 2 women.

Table 4.3: Demographics and characteristics of the subjects participating in the study.

Parameters	Mean $\pm$ SD	Median (IQR)
Age, years		25.0 (6.0)
Gender, men: women	11:2	
Spherical equivalent, D	-1.4 ± 2.8	
Cylindrical, D		-0.5 (0.75)

D, diopters; SD, standard deviation; IQR, interquartile range.

Figure 4.6 shows the defocus curves of the monofocal IOL and multifocal IOLs using the IOL effect simulated holographic setup. In the case of monofocal IOL, the visual acuity showed a maximum at 0 D, and then it decreased continuously. In the case of bifocal IOL, the visual acuity showed a maximum at 0 D, then it slightly decreased and increased, having a second peak around -3 D. It then started decreasing again. Trifocal IOL also showed its maximum visual acuity at 0 D, compared to bifocal IOL, it had a smaller decrease and had similar peaks at -2 D and -3 D. Afterwards, visual acuity decreased. At 0 D, all IOLs showed similar performance with 0.04 (0.08), 0.04 (0.08), and 0.04 (0.06) logMAR visual acuities for monofocal, bifocal, and trifocal IOLs

respectively. At -2 D defocus, trifocal IOL had better visual acuity, with 0.08 (0.06) logMAR compared to the other two IOLs which had 0.26 (0.1) and 0.3 (0.02) logMAR for monofocal and bifocal respectively. At -3 D defocus, both bifocal IOL and trifocal IOL had better visual acuity, with 0.12 (0.1) and 0.08 (0.12) logMAR respectively, than monofocal IOL, with 0.46 (0.04) logMAR visual acuity. Trifocal IOL had slightly higher visual acuity than bifocal IOL. P-values found by Wilcoxon signed rank test by pairs can be found in Table A.1.

Contrast sensitivity results of sinewave grating tests of IOLs using the IOL effect simulated holographic display is given in Figure 4.6. As expected, in case of monofocal IOL, contrast sensitivity is at the higher side of the normal human limits, making a peak at 6 cpd spatial frequency. Contrast sensitivity curves of bifocal and trifocal IOLs follow a similar trend to monofocal IOL, both having a peak at 6 cpd spatial frequency. In comparison, although they still stay within the normal human limits, they show less contrast sensitivity compared to monofocal IOL. At lower spatial frequencies, 3 and 6 cpd, it is not possible to say any IOL has better performance than others (All p-values higher than 0.050 with 5% confidence interval). The difference in performance between IOLs becomes significant in higher spatial frequencies. At 12 and 18 cpd spatial frequencies, monofocal IOL, with 1.99 (0.00) and 1.52 (0.00) log contrast sensitivity had better performance than bifocal IOL, with 1.70 (1.87) and 1.18 (1.15) log contrast sensitivity ($P = 0.047$ and 0.035 , Wilcoxon signed rank test) and trifocal IOL, with 1.70 (1.69) and 1.28 (1.15) log contrast sensitivity ($P = 0.020$ and 0.042 , Wilcoxon signed rank test). Bifocal and trifocal IOL contrast sensitivities were not significantly comparable in any spatial frequencies. Detailed p-values can be found in Table A.2.

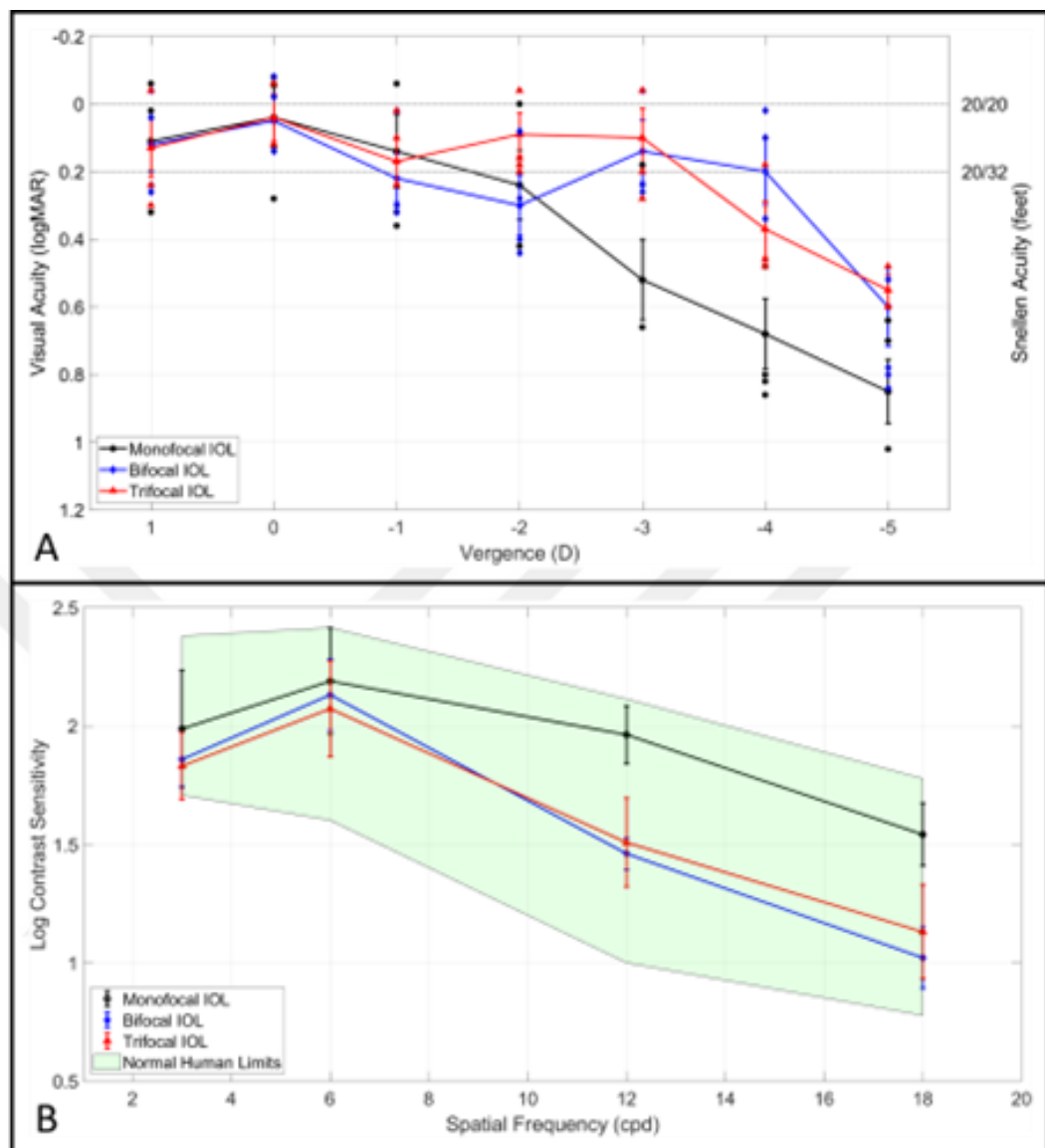


Figure 4.6: Defocus curves and sinewave grating contrast sensitivity curves of monofocal, bifocal and trifocal IOLs using the IOL effect simulated holographic setup. (A) In the case of monofocal IOL, the visual acuity showed a maximum at 0 D, and then the visual acuity decreased continuously. In the case of bifocal IOL, the visual acuity showed a maximum at 0 D, then it slightly decreased and increased, having a second peak around -3 D. It then started decreasing again. Trifocal IOL also showed its maximum visual acuity at 0 D, compared to bifocal IOL, it had a smaller decrease and had similar peaks at -2 D and -3 D. Afterwards, visual acuity decreased. (B) All curves were within normal human limits. The monofocal IOL curve had a peak sensitivity at 6 cpd then decreased with increasing spatial frequency. Bifocal and trifocal IOLs also had their peaks in 6 cpd but in less magnitude compared to monofocal IOL. Both curves followed a similar pattern to monofocal IOL with less contrast sensitivity.

Table 4.4 shows halo distance from center, halo magnitude relative to center peak, 2-point resolution test results, and satisfaction with vision. Bifocal IOL had a greater halo distance from center than trifocal IOL meanwhile trifocal IOL had a greater halo

magnitude relative to center peak. Monofocal IOL had better results in 2-point resolution limit, with 0.93 (0.07) mm distance than bifocal IOL, with 1.13 (0.06) mm distance ($P < 0.01$, Wilcoxon signed rank test), and trifocal IOL, with 1.06 (0.06) mm distance ($P < 0.01$, Wilcoxon signed rank test). Trifocal IOL also had better results than bifocal IOL ($P < 0.01$, Wilcoxon signed rank test). For satisfaction with vision, monofocal IOL again was better than other IOLs ($P < 0.01$, paired t-test, for both cases). Neither bifocal IOL nor trifocal IOL was statistically better than each other ($P = 0.096$, paired t-test).

Table 4.4: Halo quantification, 2-point resolution limit results, and satisfaction survey results of the subjects.

	Monofocal IOL	Bifocal IOL	Trifocal IOL	p-value		
				Mono & Bi	Mono & Tri	Bi & Tri
Halo distance from center, mm/cpd	-	1.60 / 313.5	1.33 / 382.9			
Halo magnitude relative to center peak (%)	-	7.6	9.4			
2-point resolution limit, mm/cpd (at 1 meter)	0.93 (0.07)* / 27.3	1.13 (0.06)* / 23.1	1.06 (0.06)* / 25.0	<0.01 [†]	<0.01 [†]	<0.01 [†]
Satisfaction with vision, 1-5	4.15 ± 0.38 [‡]	2 ± 1.22 [‡]	2.38 ± 1.19 [‡]	<0.01 [§]	<0.01 [§]	0.096 [§]

*Median (IQR); [†]Wilcoxon signed rank test; [‡]Mean ± standard deviation; [§]Paired t-test.

4.4 Discussion

To our knowledge, we are the first ones to apply the characterizations made by an artificial eye model to holographic content and show the results to subjects regardless of the severity of their cataracts. Experiencing the vision of an IOL through a holographic simulator helps subjects understand the effects of their IOL choice and show them what

to expect after surgery. The holographic display setup also has the potential to be optimized into a compact design, allowing the display to be portable and wearable if needed. The tests reported here were conducted with healthy subjects. Our results will be verified in a clinical study with patients before and after surgery by comparing the visual acuity, contrast, depth of focus, and halo and glare.

As discussed, other IOL simulators cannot achieve high visual acuity through impaired vision caused by dense cataracts. Our holographic display simulator uses a small less dense area of the cataractous lens and has been shown to be effective through dense cataracts in several patients [Kavakli et al., 2021]. In this study, we computed phase holograms using the data obtained from the physical IOLs and were able to simulate the optical performance and side effects of different types of IOLs.

One of the biggest factors in the side effects of IOLs is the pupil size and how it changes after cataract surgery [Fernandez et al., 2020]. For evaluating the visual acuity and contrast, we assumed photopic conditions and used a 3 mm pupil. On the other hand, for measuring the point light source and photic phenomena, we assumed mesopic conditions and 5 mm pupil size.

The most important limitation of this study is its focus on depths separately. In this method, contents are convolved with the PSF corresponding to the captures which are at a single depth plane. One of the noticeable aspects of multifocal IOLs is their multi-focus nature and to fully simulate a vision through a multifocal IOL, this method should be applied to contents in a multiplane manner. Physical IOL simulation using holography would be even more realistic. An optimal approach could be dividing the content into several depth planes and capturing scenes through the artificial eye model respectively at these depths. Once PSF for each depth is found, corresponding parts of the content can be separately convolved with them.

Subjects in this study were randomly chosen for unbiased results among different age groups with various accommodation abilities. We used the advantage of holographic display, sending the target image through a small section of the patient's pupil. This overcame the accommodative inabilities of the subjects, letting us determine the blur and focus of the target.

In the validation study, fully corrected visual acuity, contrast sensitivity, and 2-points resolution limits of subjects were tested with the vision simulator. The defocus curves measured using our holographic display gave similar results to the defocus curve

of patients obtained after cataract surgery with corresponding IOLs [Yamauchi et al., 2013; Alfonso et al., 2020]. Defocus curves of monofocal and bifocal IOL through holography match with the results from previous studies using IOL simulator [Na et al., 2022]. Contrast sensitivity results we obtained using vision simulator well match with the post-operative contrast studies in patients [Ravalico et al., 1993; Alfonso et al., 2020]. It was not possible to compare 2-points resolution test results to any studies before because to our knowledge this has not been done before. Regardless, the results of the test showed that photic phenomena decrease the visual acuity of the subjects. Moderate levels of photic phenomena were observed with multifocal IOLs through the artificial eye model and shown as simulations to the subjects. After seeing the photic phenomena and side effects of IOLs on our simulator, subjects were more satisfied with monofocal IOL compared to multifocal IOLs (bifocal and trifocal IOL). Our survey results also agree with previous IOL simulator studies [Na et al., 2022; Kim et al., 2020].

Chapter 5:

CONCLUSION

In conclusion, this is the first study to transform the measurement results of physical IOLs into a vision simulator using holography. Our vision simulator allows cataract patients to experience postoperative vision and photic phenomena associated with different IOL options. The test results for defocus curve, contrast sensitivity, and satisfaction matched the clinical results of patients who underwent cataract surgery and previous studies despite the discussed limitations. Using the combined methods of an artificial eye model and holographic display, regardless of the severity of their cataracts, patients could experience the side effects as well as the function of any IOLs before their surgery and know what to expect from their choice of IOL.

Moreover, our study successfully introduces a reversible and adaptable method for simulating cataract and PCO progression. The combination of our physical simulation approach and state-of-the-art objective analyses has produced compelling results that enhance our understanding of the visual implications of opacification. The insights garnered from this study will contribute to advancements in medical research and clinical practice, fostering greater awareness of opacification progression and early interventions for improved patient outcomes.

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APPENDIX A

Table A.1: Defocus visual acuity with p values

Defocus (D)	Visual Acuity (logMAR)			p-value		
	Monofocal	Bifocal	Trifocal	Mono & Bi	Mono & Tri	Bi & Tri
+1	0.1 (0.06)*	0.12 (0.08)*	0.1 (0.08)*			
0	0.04(0.08)*	0.04 (0.08)*	0.04 (0.06)*	0.594	0.762	0.344
-1	0.12 (0.02)*	0.24 (0.02)*	0.18 (0.1)*			
-2	0.26 (0.1)*	0.3 (0.02)*	0.08 (0.06)*	0.093	<0.001	<0.001
-3	0.46 (0.04)*	0.12 (0.1)*	0.08 (0.12)*	<0.001	<0.001	0.025
-4	0.56 (0.1)*	0.18 (0.14)*	0.4 (0.12)*			
-5	0.76 (0.08)*	0.6 (0.22)*	0.56 (0.08)*			

D: Diopter

LogMAR: Logarithm of the
Minimum Angle of Resolution

*Median (interquartile range)

Table A.2: Contrast sensitivity with p values

Spatial Frequency (cpd)	Contrast Sensitivity (log)			p-value		
	Monofocal	Bifocal	Trifocal	Mono & Bi	Mono & Tri	Bi & Tri
3	2.08 (1.77)*	1.79 (0.00)*	1.79 (0.00)*	0.063	0.109	0.844
6	2.20 (2.11)*	1.95 (1.85)*	2.20 (1.85)*	0.109	0.320	0.2344
12	1.99 (0.00)*	1.70 (1.87)*	1.70 (1.69)*	0.047	0.020	0.203
18	1.52 (0.00)*	1.18 (1.15)*	1.28 (1.15)*	0.035	0.042	1.000

cpd: Cycles per degree

Log: Logarithm

*Median (interquartile range)