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RESEARCH ARTICLE

Compact Models of Presbyopia Accommodative Errors for Wearable Adaptive-Optics Vision Correction Devices

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ABSTRACT Subjective accommodation refractive error models are essential for implementing adaptive vision correction devices that utilize varifocal optics. This article describes compact empirical models of subjective accommodative refractive errors in subjects with advanced presbyopia. The models are based on measurements of subjective refractive errors from fifteen presbyopes over the age of 45 using commercially available focus-tunable eyeglasses under three different illumination conditions over a 3.08D accommodation stimulus range. The resulting average residual root-mean-squared (RMS) error values for the best fitting 8-parameter model was 0.25D compared to an average RMS error of 0.4D for the conventional DDF and HHG models. The RMS error for the best-fitting model is below the average refractive error of the human eye.

INDEX TERMS Presbyopia, adaptive optics, vision correction, smart eyewear, autofocusing eyeglasses.

I. INTRODUCTION

Presbyopia is an age-related refractive disorder caused by the loss of focal tunability of the eye crystalline lens. The tunability loss manifests as an inability to focus on nearby objects and an overall object distance-dependent defocusing error. According to recent surveys [1], [2], more than a quarter of the world's population is affected by presbyopia and this number is slated to increase with an improvement in the life expectancy of the global population.

Current presbyopia mitigation devices such as multifocal and progressive lenses do not restore normal vision. Instead, these devices fragment the field of view into zones of different but fixed object distance. Therefore, these mitigation devices reduce the in-focus field-of-view resulting in major reductions in the visual acuity when averaged over the entire view field.

Fundamentally the restoration of normal vision in presbyopes requires not fixed but variable/ tunable focus lenses that can adjust their optical power and cancel out the presbyope refractive error at all object distances. Recent advances in adaptive optical systems [3]–[6] suggest that adaptive smart eyewear systems can be implemented in lightweight form and contact lens configurations [7], [8]. Autofocusing eyewear like smart-eyeglasses for example can provide automatic accommodation correction and potentially restore near-normal vision in presbyopes [5], [9]–[13]. Such devices hold great promise as they can potentially restore normal vision and improve the lives of billions of people affected by presbyopia.

The restoration of normal vision with these devices essentially requires detailed knowledge of a particular presbyope accommodative response (AR) curve. The accommodative

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response curve is defined as the change in the optical power of the crystalline lens a function of the object distance. The object distance is specified by the accommodation stimulus (AS) parameter, which is the reciprocal of the object distance [14]. The difference between the AS and the AR is often termed as accommodative lead (positive difference, over-accommodation) and accommodative lag (negative difference, under-accommodation) or, more generally, accommodative error ($AE = AS - AR$). This curve must be measured for each individual and stored in one's autofocusing device. For all practical purposes, the empirical model must be compact and developed using minimum number of measurements and fitting parameters, thereby ensuring low test costs.

A. OBJECTIVE AND SUBJECTIVE AR

The accommodative response can be measured in several ways. Objective AR is measured optically using, for example, an autorefractor instrument. However, the objective AR [15]–[28] does not necessarily represent the presbyope's defocus perception as it does not take into account the effects of the eye depth-of-focus under different lighting and pupil aperture conditions, the static refractive errors like myopia, hyperopia and astigmatism or any other higher order aberrations. In this scenario, measurement of the *subjective* accommodation response (AR), i.e., the refractive correction required by an individual to *perceive* the clearest possible focus at an object plane, provides better information about real-world performance of the presbyopic eye. The *subjective* accommodation error (AE) thus represents the optical correction which the autofocusing device would need to provide the clearest vision in presbyopes.

In practical terms, to realize such correction, a compact software mathematical model of the subjective AE would need to be programmed into the autofocusing device to produce the appropriate optical power as a function of the object distance and illumination I such that $AE = f(AS, I)$ as shown in Fig. 1. At this juncture it is important to clarify that this function is a purely empirical model extracted from patient data, and it does not directly represent the underlying phenomena responsible for presbyopia.

While this correction concept can, in principle, be performed for any presbyope, the procedure requires detailed measurement of the individual's subjective AE as a function of the object distance and lighting conditions.

B. CONVENTIONAL MODELS OF OBJECTIVE AR

Two theoretical models of *objective AR* viz., Donders-Duane-Fincham (DDF) model and the Helmholtz-Hess-Gullstrand (HHG) model have been discussed extensively in literature so far, which originate from the two major variants of the lenticular theory [29], [30]. In the HHG theory, the loss in accommodation is completely attributed only to the morphological changes in the human eye's crystalline lens capsule. This model shows a near-ideal response to stimulus over the manifest zone and then a sharp transition into a hard saturation region, where the lens can no longer accommodate. The AR curve corresponding to HHG theory is shown in Fig. 2(a). In contrast, DDF theory attributes loss in accommodation with age to changes in only the bio-mechanical properties of ciliary muscle which controls the crystalline lens. The DDF model exhibits a linear response between the objective AR and the applied stimulus, with the slope of the line reducing progressively with age, until it is nearly flat and stagnated for advanced presbyopes. Fig. 2(b) shows the corresponding accommodation characteristics of DDF model.

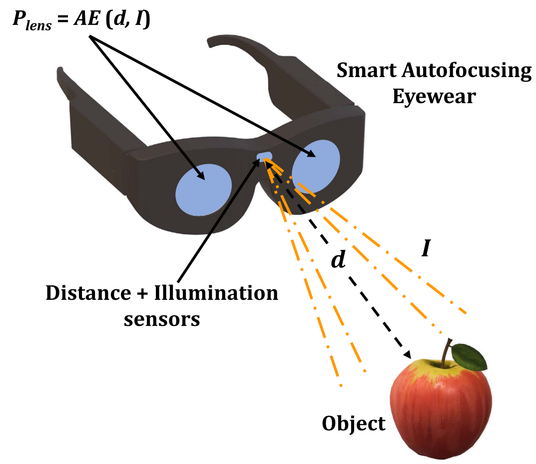

FIGURE 1. Schematic of an accommodation correction system. The accommodation error model is programmed into the correction device. The variable-focus lenses vary their optical power in real-time based on the programmed accommodation error (AE) model for the individual wearing the correction device. Variables d and I represent the object distance and illumination, respectively.

To our knowledge, no studies have been conducted to date to model the subjective AE of presbyopes at various object distances and under various illumination levels.

The objective of this paper is the development of individual-specific compact empirical models of subjective AE as functions of object distance and illumination, for use in autofocusing eyewear to provide adequate vision correction for presbyopes. The models are constructed from subjective AE data collected from a group of advanced presbyopes in a registered clinical study. For each of these presbyopes the model is fitted to the presbyope data using as few parameters as possible. We also compare the RMS fitting errors of these models in relation to accommodation restoration using smart autofocusing eyewear technologies.

C. IMPACT OF DEFOCUS AE ON VISUAL ACUITY

Visual Acuity (VA) is a unit less number that quantifies the clarity of vision and rates the ability of the human visual system to recognize/ resolve small details with precision. It is measured in terms of the logarithm of the minimum angle of resolution (MAR) or logMARs [31]–[35]. Standard vision is defined as logMAR of 0.0 (Snellen 20/20), and it helps to define normal vision. Positive logMAR values indicate vision loss, while negative values denote normal or better visual acuity. It is well known that the visual acuity deteriorates in presence of defocus errors [36]. A simple empirical model,

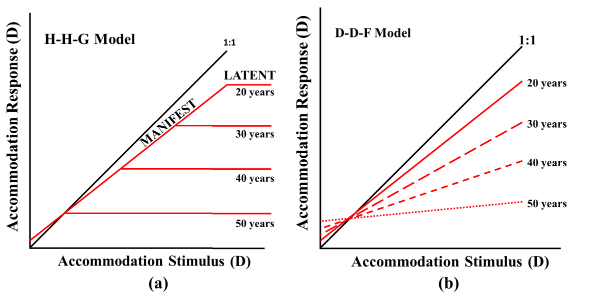

FIGURE 2. Comparison of accommodation response characteristics from the (a) Helmholtz-Hess-Gullstrand and the (b) Duane-Donders-Fincham models [29]. The black line is the ideal accommodation response. Neither model fits actual individual AD characteristics adequately for performing a good AD correction with adaptive smart corrective eyewear.

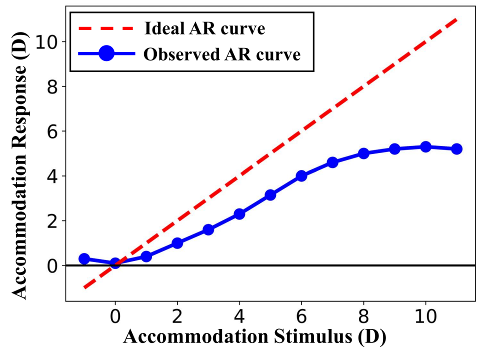

FIGURE 3. Example of typical subjective accommodation response curve for the human eye. The curve progressively flattens for higher powers with age [29], [38].

suggested by Blendowske [37], relates the defocus error and deterioration in the visual acuity as:

$$\frac{V}{V_{bc}} = \frac{1}{1 + D^2} \tag{1}$$

where V is the deteriorated decimal visual acuity. V_{bc} is the best corrected decimal visual acuity and D is the defocus error in diopters. For example, a defocus error of 1.0D reduces the visual acuity by 50%, thereby degrading the visual acuity by 0.3 logMAR. Any model used for restoration of the subject-specific AE curve must ideally be sufficiently accurate to minimize the reduction of visual acuity at all object distances and under all illumination conditions.

In practical terms, patient-specific AE models with low RMS errors will have to be programmed into autofocusing eyewear devices to avoid over- or under-compensation and avoid related deterioration in the vision of the presbyope. We have shown the implementation of one of the models in a recently published article which demonstrates the design and implementation of a pair of smart autofocusing, liquid-lens eyeglasses to mitigate presbyopia [13].

II. COMPACT MODEL DEVELOPMENT

Every individual's eye behavior is characterized by their $AR(AS)$. Ideally, the AR curve should be a line of unity slope, but as presbyopia progresses, the curve flattens to lower optical powers [14], [29], [38]–[42] as shown in Fig. 3.

In this article, we seek to develop compact empirical equations that fit the subjective AR curve for a particular eye of an individual with the smallest number of parameters. Several types of models were fitted to patient data based on the underlying physics and pure empirical observations of the patient data. These models are discussed in the sections below.

A. SIGMOID MODEL

The typical subjective AR curve displays a saturation type of response with flat regions at low and high accommodation stimuli and an approximately uniform slope in between (Fig. 3). Empirical data [14], [29], [39]–[41], [43]–[48] of AR in the literature exhibits a typical "S"-shaped or sigmoid curve that can be modelled using the sigmoid function family [49]. The simplest continuous static sigmoid function [50] is the shifted logistics equation

$$AR_{sig} = \frac{a}{1 + \left(e^{-k\cdot[AS-b]}\right)} - d \tag{2}$$

where a, b, d and k are fitting parameters. a is the maximum subjective accommodative amplitude, k represents the range of AS for which the AR exhibits a linear response and b represents the shift along the horizontal axis. The parameter d represents any vertical shift in the subjective AR due to the presence of other refractive errors.

B. PIECEWISE LINEAR MODEL

The subjective AR in Fig. 3 can also be modeled with simple piecewise linear models with one or two breakpoints, using the Heaviside step function $\boldsymbol{H}(x)$. We can define a ramp function $\boldsymbol{R}$

$$\boldsymbol{R}(x, S, \alpha) \equiv (x - \alpha) \cdot S \cdot \boldsymbol{H}(x - \alpha) \tag{3}$$

where S is the slope of the ramp function. Using these definitions, we can approximate the curve with one breakpoint piecewise linear model

$$AR_{PWL1} = \boldsymbol{R}(AS, k_1, 0) + \boldsymbol{R}(AS, k_2, x_0) + f \tag{4}$$

where, k_1, k_2, are slopes of the two segments and are fitting parameters and x_0 is the location of the breakpoint. Similarly, a two- breakpoint piecewise linear model can be constructed as:

$$\begin{aligned} AR_{PWL2} = {} & \boldsymbol{R}(AS, k_1, 0) + \boldsymbol{R}(AS, k_2, x_0) \\ & + \boldsymbol{R}(AS, k_3, x_1) + f \end{aligned} \tag{5}$$

where, k_1, k_2, k_3 are the slopes the three segments, respectively and x_0, x_1 are the locations of the two breakpoints and f is an initial offset. We choose to limit the number of breakpoints used in the piecewise linear models to a maximum of two since any more breakpoints would increase the number of fitting parameters beyond the number of observations.

C. ILLUMINATION DEPENDENCE

The eye pupil changes diameter with different illumination levels to control the light intensity at the retina. At high illumination the pupil constricts to a smaller diameter. This increases the depth-of-focus of the visual system and improves vision. Thus, the subjective *AR* will correspondingly vary under different illumination levels. Recent studies have shown that the subjective *AR* of presbyopes bears a linear relation with the logarithm of luminance levels [51] and the slope of the *AR* curve decreases with reduction in object luminance [52]. Since luminance is proportional to illumination (in lux) we modify the sigmoid model of Eq. (2) as

$$AR_{sig}\left(AS, I\right) = \left(\frac{a}{1+\left(e^{-k\cdot[AS-b]}\right)} - d\right)\cdot \log\left(I\right) \tag{6}$$

where I is the illumination in lux. Similarly, the illumination dependence is included in the piecewise models as a multiplicative factor.

$$\begin{aligned} AR_{PWL1,I} &= AR_{PWL1}\cdot \log\left(I\right), \\ AR_{PWL2,I} &= AR_{PWL2}\cdot \log\left(I\right) \end{aligned} \tag{7}$$

Since the range of the illumination parameter in our recorded observations was narrow and approximately within the same orders of magnitude (75–800 lux), we also utilized a simple exponential function to model the illuminance dependence on the subjective *AR*. Accordingly, using Eq. (2), (4) and (5) the sigmoid and piecewise linear models can be expressed as:

$$AR_{sig}\left(AS, I, \beta\right) = \left(\frac{a}{1+\left(e^{-k\cdot[AS-b]}\right)} - d\right)\cdot I^{\beta} \tag{8}$$

and the piecewise approximations

$$AR_{PWL1,I} = AR_{PWL1}\cdot I^{\beta}, AR_{PWL2,I} = AR_{PWL2}\cdot I^{\beta} \tag{9}$$

where β is an empirical exponent.

In addition, we also utilized a piecewise linear model with one breakpoint with slopes and breakpoints that are linear functions of the illumination

$$f_I = f_{00} + f_{01}\cdot I, \tag{10}$$

$$x_I = x_{00} + x_{01}\cdot I, \tag{11}$$

$$k_{I1} = k_{10} + k_{11}\cdot I, \tag{12}$$

and,

$$k_{I2} = k_{20} + k_{21}\cdot I \tag{13}$$

where $f_{00}, f_{01}, x_{00}, x_{01}, k_{10}, k_{11}, k_{20}$ and k_{21} are fitting parameters. The linear dependence of parameters appears to fit the characteristics of the patient data and it is purely empirical.

III. METHODS

A. STUDY DESIGN

Many researchers have explicitly studied objective *AR* or *AE* in order to analyze the effects of age on the morphological changes observed within the eye and how the *AR* changes with age [15], [16], [24]–[28], [40], [43]–[48], [53], [54]. Such objective *AR* and *AE* are measured using sophisticated, computer controlled optometers, retinoscopes and aberrometers. In these studies, the pupil diameter is tightly controlled with the help of phenylephrine [55] in order to decouple the effects of the depth of field on the *AR/AE* measurements. Therefore, such measurements cannot be directly or easily converted to subjective *AR/AE* [15], [16], [19]–[23], [27], [56], [57] for practical use in autofocusing eyewear.

FIGURE 4. A presbyope undertaking a visual task at 1m chart distance.

The focus of this study is to develop *AE* models which can be directly used in smart autofocusing eyewear algorithms for personalized corrections. In order to simulate the real-world conditions in which such algorithms and systems will be used, we measured the subjective accommodative errors which the presbyopic subjects exhibited during their visual task at different distances as described in the subsections below. The methodology used in this study resembles the "push-up" method of analyzing accommodation amplitude, where the target or an eye chart is moved progressively closer to the subjects' eyes till a blurry image is reported [24], [26]. In our method, the distance is fixed at a few locations away from the observer and the image at each location is made subjectively as clear as possible utilizing a set of commercially available tunable power eyeglasses which cancel the distance-dependent subjective refractive error. Additionally, we utilize calibrated ETDRS charts for various chart distances in which the angular detail of the optotypes is maintained [31]–[35]. The details of the experimental method used is described in the sections below.

B. HUMAN STUDY PARTICIPANTS AND EXCLUSION CRITERIA

Human study approval was acquired from the University of Utah Institutional Review Board (IRB 00114415), and experiments were performed according to the ethical standards laid down in the Declaration of Helsinki, 1964. A total of 15 advanced presbyopia subjects ages 45-68 years, with a mean age of 54.6 (S.D. = 6.8) years, were recruited from a population of patients from the University of Utah Moran Eye Center and associated clinics. Since emmetropes and early presbyopes generally exhibit a non-zero objective accommodation amplitude, they were not considered for this study. All subjects provided informed, signed consent before entry into the study. A record on clinical trials performed in this study has been registered with ClinicalTrials.gov [58] (NCT03911596). Individuals with astigmatism > 1.0D, artificial intraocular lenses, or those having any ocular pathology that would inhibit accommodation of their natural lenses were excluded. The recruited subjects had prescriptions between -1.5 D and $+2.5$ D and were correctable to logMAR 0.0 (Snellen 20/20).

C. STUDY CONDITIONS

Testing was carried out in a light proofed optometry exam room at the University of Utah Moran Eye Center. Chart retro-illumination was not used. In order to explore the effects of illumination on the accommodative insufficiency, we conducted the study under 3 chart illumination levels- 75 lux to simulate dark conditions, 500 lux to simulate normal indoor lighting conditions and 800 lux to simulate outdoor conditions [59]. Chart illumination was controlled using an LED studio lighting system which also provided diffused lighting for the exam room. The correlated color temperature of the lighting system was fixed at 5000 K. Chart illumination was kept constant while measuring the accommodation insufficiency. Fig. 4 shows a subject undertaking a visual task.

D. STUDY PROCEDURE

The visual task consisted of reading optotypes on 7 different Early Treatment Diabetic Retinopathy Study (ETDRS) charts calibrated for 7 distances (4 m, 2 m, 1 m, 70 cm, 50 cm, 40 cm and 30 cm) under 3 chart illumination conditions (75 lux, 500 lux and 800 lux). We utilized a commercially available manually-tunable variable-focus Adlens Hemisphere eyeglasses to assess the subjective AE. The standard deviation of the eyeglasses optical power was 0.03 D. Subjects were assisted by the study staff in manually tuning the eyeglasses until they could correctly identify, to the best of their ability, the optotypes corresponding to logMAR 0.0 (Snellen 20/20) line on the ETDRS charts. Each lens in the eyeglasses was monocularly tuned before patients undertook the visual task, binocularly, under every test distance and illumination condition. Once optimum subjective refraction was reached, the visual acuity of the subjects was measured in logMAR. This was done to verify if the variable-focus eyeglasses provided the optimum correction necessary for subjects during all the visual tasks. The eyeglasses were then taken from the subjects and their optical power was measured using a Thorlabs WFS150C-AR Shack-Hartmann wavefront sensor. Subjects who were unable to completely identify the optotypes corresponding to logMAR 0.0 line, were instead asked to identify the optotypes on the previous line (logMAR 0.1) and their visual acuity was recorded accordingly. Those who could correctly identify the optotypes corresponding to the logMAR 0.0 line were then asked to identify optotypes on the logMAR -0.1 line and their visual acuity was recorded accordingly. Calibrated ETDRS charts, which were placed at distances of 1 m, 70 cm, 50 cm and 30 cm, were carefully designed and printed on high quality optical white paper. Subjects were tested under chart illumination levels of 500 lx, 800 lx and 75 lx, sequentially. The study was paused after changes in the illumination levels till the subjects were comfortable with their visual experience. The study sessions lasted between 1.5–3 hours in duration for each subject.

E. MODEL FITTING

The subjective AE models were surface-fitted to the recorded subjective AE data for each eye of every subject with the Python 3.8 SciPy 1 optimization library [60]. We utilized the Levenberg-Marquardt nonlinear least squares method for surface-fitting Eq. (6)–(9) to the AE data [61]–[63] using the $k's$, $x's$ and f as fitting parameters. The independent variables-AS and the illumination were not bounded during the surface-fitting process as the simple Levenberg-Marquardt method does not handle bounds on the independent variables [63], [64]. The surface-fitting process produced a separate characteristic equation and a surface for every eye of every subject. The surface-fits were analyzed and compared with each other using the R^2 metric (goodness of fit/ coefficient of determination) and the residual RMS errors exhibited by the models. It is important to note that these models are going to be used for personalized corrections using autofocusing eyewear and hence the fitting parameters extracted from the curve fitting process would vary from eye to eye.

IV. RESULTS AND DISCUSSION

A. ACCOMMODATION ERRORS

The observed subjective AEs of the subjects were plotted against the AS. The recorded data has been made available in Data File 1. Figs. 5 – 7 show the observed AE for 15 subjects (30 eyes) under chart illumination levels 75, 500 and 800 lux, respectively. The solid lines with circular markers represent the right eye and the dashed lines with square markers represent the left eye. The size of the markers is proportional to the standard deviation of the measurement instrument. The markers are 5x the standard deviation for better visibility.

It can be clearly seen from the plots that the subjective AE kept increasing with increasing stimulus, under all illumination conditions. All subjects required higher positive additions to perform near vision tasks at AS of 2.5 D and 3.33 D, corresponding to chart distances of 40 cm and 30 cm, respectively, when compared to far vision tasks at AS of 0.25 D and 0.5 D corresponding to chart distances of 4m and 2m, respectively. This trend is consistent with the existing theories of presbyopia [29] and subjective/ objective AR data reported throughout literature [14], [29], [39]–[42], [44]–[48], [54] where the AR reduces or stagnates irrespective of the increase in the stimulus and causes AE to increase with increasing stimulus. It can also be seen from Figs. 5 – 7 that the recorded AE in the left and the right eyes of all subjects are very similar under all illumination conditions, except for subject #5 whose right AE is significantly higher compared to their left AE. This could be attributed to the presence of a high-power static refractive error in one of their eyes.

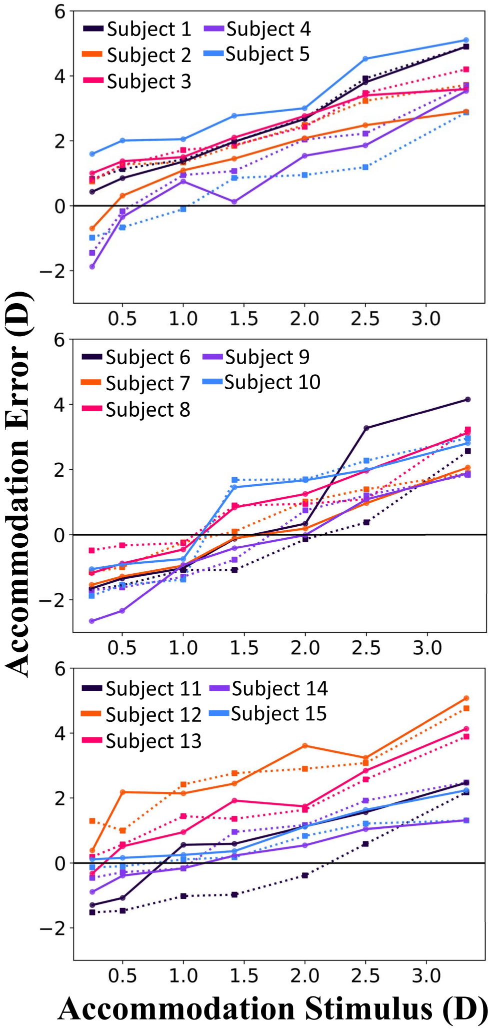

FIGURE 5. Measured accommodation error for 15 subjects (30 eyes) under 75 lux chart illumination. The markers show the recorded data from the subjects. The solid lines with circular markers and the dashed lines with square markers represent subjective AE in the right and left eye, respectively. The size of the markers is 5 times the standard deviation of the manually tunable eyeglasses, for better visibility.

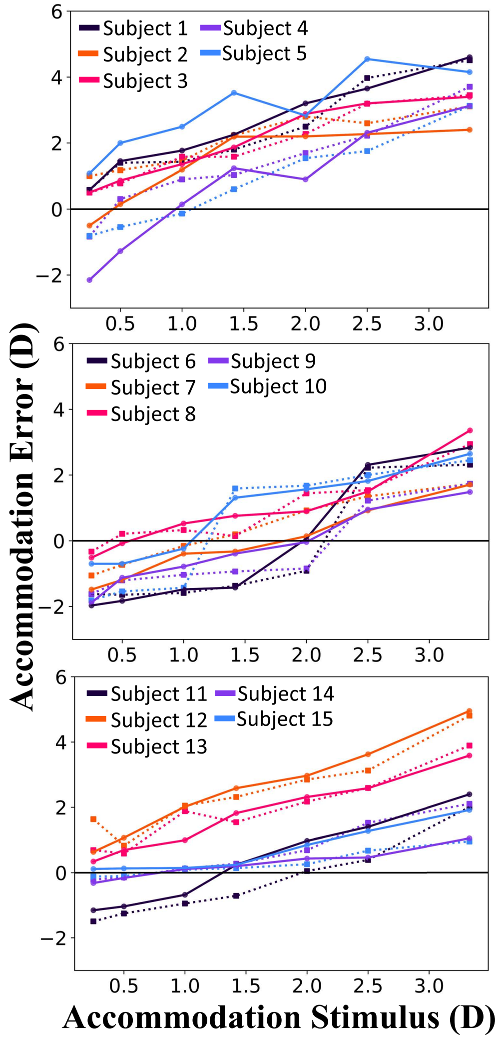

FIGURE 6. Measured accommodation errors for 15 subjects (30 eyes) under 500 lux chart illumination. The markers show the recorded data from the subjects. The solid lines with circular markers and the dashed lines with square markers represent subjective AE in the right and left eye, respectively. The size of the markers is 5 times the standard deviation of the manually tunable eyeglasses, for better visibility.

B. AE AND ILLUMINATION

We also explored the dependence of subjective AE on the illumination levels. Fig. 8. shows the average of the absolute subjective AE for 30 eyes under 3 different illumination conditions of 75, 500 and 800 lux. The blue bars represent the left eyes while the purple bars represent the right eyes. A trend can be seen from the plots where the average subjective AE progressively decreases with illumination for most subjects. However, this trend cannot be quantified in this study for two main reasons-(a) the AE was studied under only 3 illumination levels and (b) only 15 presbyopes (30 eyes) were studied, which are insufficient to conduct any meaningful statistical analysis.

The depth of field of the human eye is dependent on the pupil diameter [65] and the pupil diameter is a function of the illumination levels. With increasing illumination levels, the pupil size decreases [66], thereby increasing the depth-of-focus of the visual system and reducing the required amplitude of accommodation to focus on an object. One of the principal reasons why presbyopes tend to squint is because doing so improves their visual acuity at the expense of much lower brightness. At lower illumination levels, the pupil size increases, reducing the depth of field and requiring more accommodation amplitude [53]. It has also been shown that accommodation depends on cone activity [67], which ceases completely under low illumination conditions. These factors can help explain why a majority of the subjects exhibited decreasing average AE with increasing illumination. Fig. 8 also shows that for few participant's eyes, the AE at 800 lux was greater than that at 500 lux. This seems counter-intuitive as accommodation error decreases under higher object illumination levels due to improved depth-of-focus [68]. However, there are some clues to explain this discrepancy. As light enters the eye, part of its energy is scattered in the crystalline lens [69] and gives rise to a phenomenon called disability glare [70], [71]. Disability

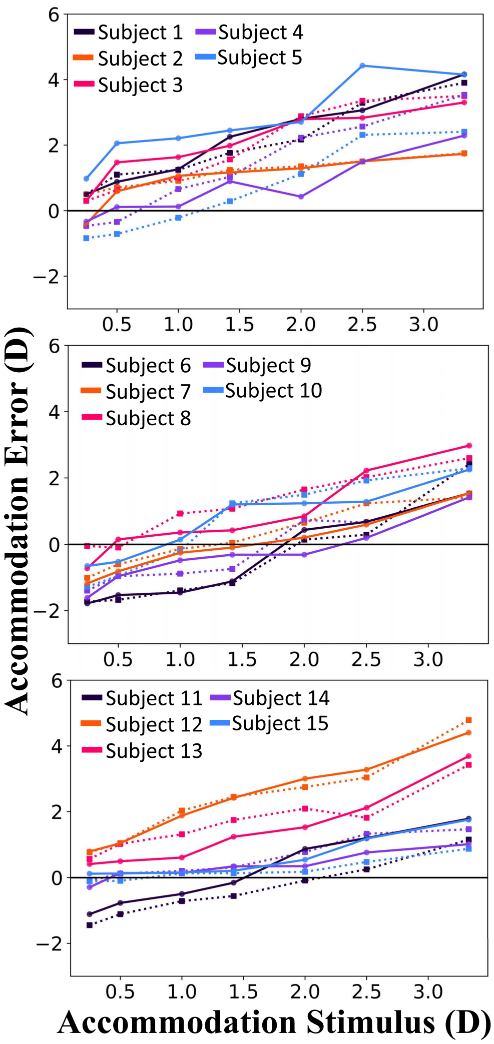

FIGURE 7. **Measured accommodation errors for 15 patients (30 eyes) under 800 lux chart illumination. The markers show the recorded data from the subjects. The solid lines with circular markers and the dashed lines with square markers represent AE in the right and left eye, respectively. The size of the markers is 5 times the standard deviation of the manually tunable eyeglasses, for better visibility.**

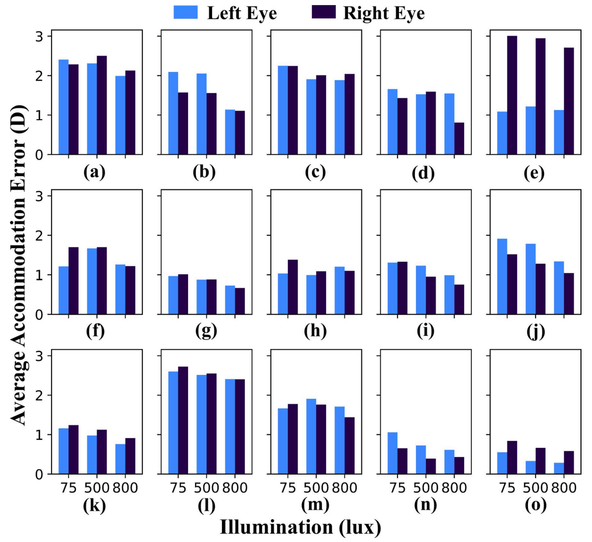

FIGURE 8. **Average accommodation errors for 15 subjects (30 eyes). A majority of subjects show decreasing AE with increasing illumination levels. Subplots (a)–(o) shows the average accommodation errors for Subjects 1-15, respectively.**

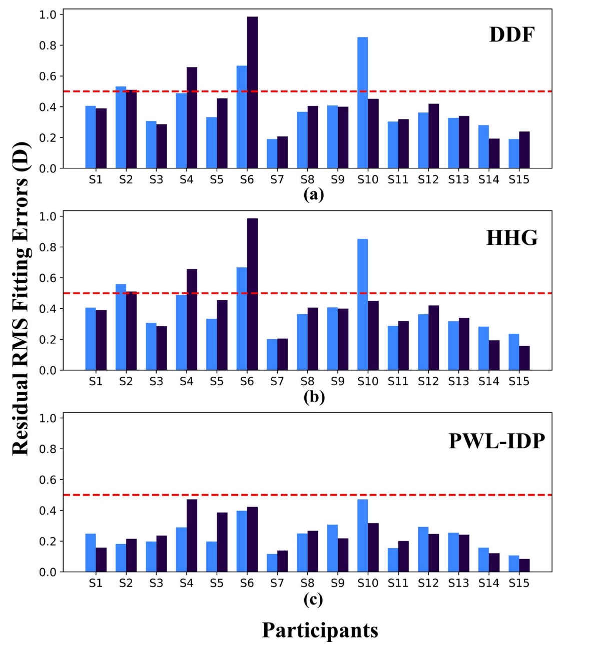

FIGURE 9. **RMS fitting errors for 15 subjects corresponding to the older (a) DDF and (b) HHG models in literature and the (c) proposed best fitting one-break PWL model with illumination dependent parameters (PWL-IL). The blue bars represent the left eye and the purple bars represent the right eye. RMS fitting errors for the one-break PWL-IDP models are significantly smaller compared to the older model.**

glare is a physiological glare which generally impairs vision, decreases visual acuity without causing any discomfort [72] and decreases contrast sensitivity [73]. Due to anatomical changes in the visual system with age [74]–[77], senior people were found to be more sensitive to glare [78]–[81]. This could possibly explain why some eyes in this study exhibit higher accommodation error amplitudes at 800 lux chart illumination compared to those at 500 lux chart illumination. Detailed analyses of subjective *AE* with respect to the illumination levels is beyond the scope of this work.

C. MODEL FITTING RESULTS AND DISCUSSION

At any given fixed power, the crystalline lens in the human eye has an average RMS wavefront error of 0.35 μm [82]. This is roughly equivalent to a natural refractive defocus of 0.3D [83]–[86] at a pupil size of 6 mm diameter, for healthy, young individuals and steadily increases with age [86], [87] which cannot be corrected by conventional eyeglasses. Therefore, in order to restore pre-presbyopic accommodation we seek subjective *AE* models with average errors < 0.3D and any model that meets this criterion is sufficient for describing the AE. To find such model we have fitted each of our proposed AE models for each of the 30 eyes considered in this study, and from the RMS errors for each fit we report the averaged RMS error for the eye set. The model fitting plots and the fitting parameters for 15 subjects (30 eyes) and all proposed models are available in the supplemental document.

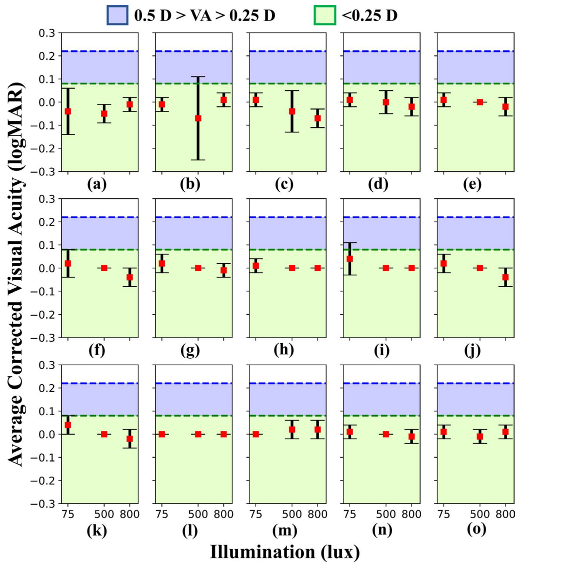

FIGURE 10. Average corrected visual acuities (VA) of 15 subjects recorded under various chart illumination conditions. The bars represent one standard deviation. Average corrected VA of all 15 subjects is less than +0.1 logMAR. The blue and green dashed lines correspond to spherical refractive errors of 0.5 D and 0.25 D, respectively. Subplots (a)–(o) shows the average corrected VA for Subjects 1-15, respectively.

TABLE 1. Average RMS fitting errors for 15 subjects (30 eyes).

Fitting Model	# of parameters	Avg. RMS error (D)
DDF	2	0.409
HHG	3	0.408
Sigmoid- Log (**S-L**)	4	0.331
One-break PWL – Log (**1-PWL-L**)	4	0.376
Two-break PWL – Log (**2-PWL-L**)	6	0.364
Sigmoid- Exponential (**S-E**)	5	0.291
One-break PWL- Exponential (**1-PWL-E**)	5	0.294
Two-break PWL- Exponential (**2- PWL-E**)	7	0.283
One-break PWL- Illumination dependent parameters (**PWL-IDP**)	8	0.25
One-break PWL with individual illumination fits	12	0.206

Table 1 shows the averaged RMS error over the 30 eyes for each of our models. The last line shown in the table corresponds to individual fits for each illumination level.

Out of the 7 proposed models investigated, 3 models featuring an exponential dependence on illumination parameter (S-E, 1-PWL-E and 2-PWL-E) and the illumination-dependent parameters (PWL-IDP) model had RMS fitting errors smaller than the 0.3D threshold. The fits with the most fitting parameters have the lowest average RMS errors. The best fitting PWL-IDP model displayed nearly half the average RMS fitting error (0.25D) than the DDF (0.4D) and the HHG models (0.4D). Neither the DDF and HHG models feature illumination dependence.

Fig. 9 shows a comparison of the residual RMS fitting errors obtained for 15 subjects (30 eyes) after fitting the older DDF, HHG models and the one-break PWL with illumination dependent parameters (PWL-IDP). The blue bars represent the left eyes while the purple bars represent the right eyes. The red dashed line represents a 0.5D reference. Our best illumination-dependent fitting model (PWL-IDP) exhibits fitting errors less 0.3D, for 23 out of the 30 eyes (77%). The PWL-IDP model outperforms the older DDF and HHG models and, on the average, meet the acceptability threshold <0.3D for our experimental data set.

TABLE 2. Average corrected visual acuities of 15 subjects under 3 chart illumination levels.

Subject	Corrected VA (75 lux)	Corrected VA (500 lux)	Corrected VA (800 lux)
1	-0.04	-0.05	-0.01
2	0.01	-0.07	0.01
3	0.01	-0.04	-0.07
4	0.01	0.0	-0.02
5	0.01	0.0	-0.02
6	0.02	0.0	-0.04
7	0.02	0.0	-0.01
8	0.01	0.0	0.0
9	0.04	0.0	0.0
10	0.02	0.0	-0.04
11	0.04	0.0	-0.02
12	0.0	0.0	0.0
13	0.0	0.02	0.02
14	0.01	0.0	-0.01
15	0.01	-0.01	0.01

D. VISUAL ACUITY MEASUREMENTS

The results of our study show that the corrected visual acuities of all subjects were mostly equal to or better than +0.1 logMAR (Snellen 20/25) under all stimuli and illumination conditions when we use the best fitting model, i.e., the piecewise linear model with one breakpoint and illumination dependent parameters (PWL-IDP). Visual acuities recorded at different distances and chart illumination conditions for each participant have been included (Data File 2). Fig. 10 shows the average corrected visual acuities of the subjects under varying accommodation stimuli and chart illumination levels. None of the 15 subjects exhibited average corrected visual acuity worse than +0.04 logMAR.

In order to better relate an individual's visual acuity to lens prescription, visual acuity scores corresponding to defocus errors of 0.25D and 0.5D were calculated using the simple relation of Eq. (1). More accurate *VA* formulas corresponding to spherical refractive errors are available in literature [88] if the pupil size is known. The blue and green lines in Fig. 10 show these calculated visual acuity scores of 0.026 logMAR and 0.096 logMAR corresponding to spherical refractive errors of 0.25D and 0.5D, respectively. Corrected visual acuities falling within the blue region correspond to an acceptable spherical refractive error between 0.25D and 0.5D, while those falling within the green region correspond to a minimum desirable spherical refractive error of < 0.25 D in the autofocusing eyewear. In optometry, visual impairment is defined as visual acuity of worse than +0.3 logMAR (Snellen 20/40) [89]. Since the corrected visual acuities of all subjects while wearing tunable eyeglasses were better than or equal to +0.1 logMAR and the subjects responded favorably after each optical adjustment, it can be concluded the subjects in this study were adequately corrected. It can also be seen from Fig. 10 that a majority of the subjects were properly corrected for their *AE* at 500 lux chart illumination (*VA* =∼0.0 logMAR).

Table 2 shows the recorded visual acuity of the subjects, averaged over all 7 distances, under the chart illumination levels of 75, 500 and 800 lux. Overall, the corrections provided by the focus-tunable Adlens eyeglasses were able to adequately compensate the subjective accommodation errors present in the subjects' eyes, resulting in an average corrected *VA* of −0.01 logMAR for all eyes at 500 lux standard illumination condition [59].

V. CONCLUSION

We present new empirical models for presbyopes, which relate their subjective accommodative errors to the object distance and the illumination levels. Measured accommodation errors of all subjects under all illumination conditions increased with a reduction in the test-chart distance, indicating a loss in the accommodative abilities of their visual system. The proposed subjective accommodation error models were surface-fitted to the measured data of 15 subjects and the best fitting subjective accommodation error model exhibited an RMS fitting error of less than 0.3D, the average equivalent refractive error of a human eye, for a majority (77%) of the eyes under examination. These empirical AE models are sufficiently accurate for use in future personalized adaptive optics based autofocusing devices.

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