

# Experimental Simulation of Simultaneous Vision

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**PURPOSE.** To present and validate a prototype of an optical instrument that allows experimental simulation of pure bifocal vision. To evaluate the influence of different power additions on image contrast and visual acuity.

**METHODS.** The instrument provides the eye with two superimposed images, aligned and with the same magnification, but with different defocus states. Subjects looking through the instrument are able to experience pure simultaneous vision, with adjustable refractive correction and addition power. The instrument is used to investigate the impact of the amount of addition of an ideal bifocal simultaneous vision correction, both on image contrast and on visual performance. The instrument is validated through computer simulations of the letter contrast and by equivalent optical experiments with an artificial eye (camera). Visual acuity (VA) was measured in four subjects (age:  $34.3 \pm 3.4$  years; spherical error:  $-2.1 \pm 2.7$  diopters [D]) for low and high contrast letters and different amounts of addition.

**RESULTS.** The largest degradation in contrast and visual acuity ( $\sim 25\%$ ) occurred for additions around  $\pm 2$  D, while additions of  $\pm 4$  D produced lower degradation (14%). Low additions (1–2 D) result in lower VA than high additions (3–4 D).

**CONCLUSIONS.** A simultaneous vision instrument is an excellent tool to simulate bifocal vision and to gain understanding of multifocal solutions for presbyopia. Simultaneous vision induces a pattern of visual performance degradation, which is well predicted by the degradation found in image quality. Neural effects, claimed to be crucial in the patients' tolerance of simultaneous vision, can be therefore compared with pure optical effects. (*Invest Ophthalmol Vis Sci.* 2013;54:415–422) DOI:10.1167/iovs.12-11219

Although none of the current available solutions for presbyopia (the age-related loss of crystalline lens accommodation) restores the full dynamic capability of the young eye to change its refractive power upon an accommodative stimulus, there are multiple treatments that attempt to provide functionality for both near and far vision to the presbyopic

patient. Current techniques for the correction of presbyopia are based on one of four principles: alternating vision, monovision, simultaneous vision, and (in a very early stage) accommodating IOLs. Some of the optical corrections available relying on alternating vision are progressive lenses (where changes in gaze or head position allow selection of the zone of the spectacle used to view near or far objects)<sup>1</sup> or translating contact lenses (where the lens, typically gas permeable, moves upwards on the eye during downgaze while viewing close).<sup>2</sup> In monovision, one eye is corrected for distance and the other eye is corrected for near vision. Monovision solutions are commonly applied in the form of corneal, intraocular lens, or contact lens treatments.<sup>3</sup> An increasingly popular approach to the correction of presbyopia relies on simultaneous vision designs where the eye is corrected for distance and near vision simultaneously.<sup>4,5</sup> Bifocal solutions generally come in the form of refractive contact lenses, and diffractive or refractive intraocular lenses.

Simultaneous vision represents a new visual experience in which a sharp image is superimposed to a blurred replica of the same image, thus reducing the overall contrast. The intended optical effect of the correction (driven by its design) is combined with the particular aberration pattern of the eye, so a given bifocal design does not produce the same optical through-focus energy distributions in all eyes. In current designs, near addition values typically range from 1 to 4 diopters (D).<sup>6</sup>

Not all patients tolerate the contrast reduction induced by simultaneous vision. It is often argued that only those patients learning how to process the image, by ignoring or suppressing the out-of-focus image components, adapt to simultaneous vision solutions. These mechanisms for adaptation remain to be elucidated, but are supposed to rely on specialized and sophisticated (while automatic) neural processes that restore the physically degraded images to overcome their optical quality limitations.

To date, it is not clear if the lack of adaptation to multifocal vision has a physical or a neural origin, or whether they are combined. A better understanding of optical and visual interactions in simultaneous vision bifocal corrections is critical to improve lens prescription.

To date, most studies of the outcomes of bifocal corrections compare visual clinical outcomes in patients prescribed with lenses available in the market.<sup>7–10</sup> Also, systematic evaluations of many of the available lenses are normally performed on bench (lacking from the optical, and of course, the neural complexity of a patient).<sup>11–13</sup>

Adaptive optics (AO) systems are useful to manipulate the aberrations to which a subject is exposed.<sup>14–19</sup> Many of the laboratory-based AO visual simulators rely on deformable mirrors that can only modify relatively low amounts of high-order aberrations (i.e., spherical aberration), limiting the ability to introduce phase patterns with discontinuities or reproduce bifocal patterns. AO systems based on spatial light modulators<sup>20,21</sup> show much higher spatial resolution, allowing in principle steep slope changes in the wavefront, although they may be subject to additional limitations (i.e., chromatic, phase

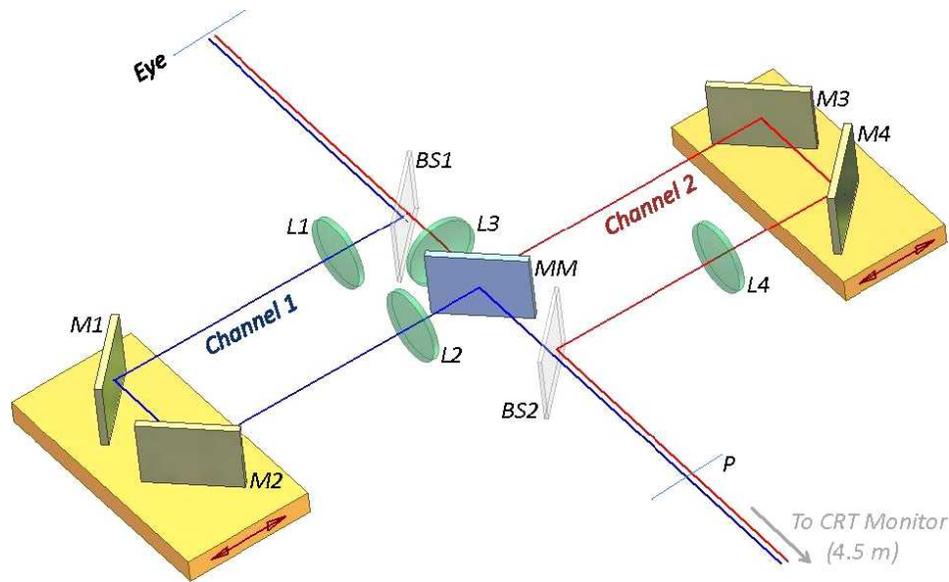
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**FIGURE 1.** Schematic diagram of the simultaneous vision system. Light from a CRT monitor is separated into two channels by means of a beamsplitter ( $BS_2$ ) and recombined by means of a double mirror (MM) and a second beamsplitter ( $BS_1$ ). Each channel consists of an independent Badal optometer (composed by two 150-mm focal length lenses, and two mirrors mounted on a motorized moving stage). Channel 1 (blue line) is typically focused at far (subject's distance prescription) and channel 2 (red line) moves to simulate near additions. An artificial pupil (P) limits the natural pupil size (4 mm in this study). The two channels (illustrated with red and blue lines slightly separated) are perfectly coincident in the real setup between the monitor and  $BS_2$ , and between  $BS_2$  and the eye.

wrapping and diffraction effects),<sup>22,23</sup> and are still relatively pricey.

In this study, we present a new instrument<sup>24</sup> that allows the experimental simulation of a pure simultaneous vision correction by combining two channels, one focused at near and the other focused at far. The system allows for experimental simulation of idealized bifocal corrections (i.e. isolated from factors inherent to the specifics of real corrections), such as lens flexure and fitting in a bifocal contact lens, tilt or decentration of a bifocal IOL, yet in the presence of the subject's own aberrations and neural response. The system allows investigation of fundamental questions associated with simultaneous vision, with a relevant practical interest.

In particular, this study addresses the impact of the amount of addition power on image quality and on visual performance with a simultaneous vision correction in an experimental setting. The additions used in this study ranged from 0 to 4 D, within the range of additions generally available in commercially available bifocal designs. To our knowledge, this is the first systematic study of the impact of the amount of addition on contrast degradation and visual acuity (VA). Using the new developed simultaneous vision experimental simulator, the study will respond to the following specific questions: What is more deleterious for vision: a large addition which creates a low contrast largely defocused retinal image superimposed to a sharp image, or a low addition which introduces a smaller amount of blur but with the two images (one sharp and one slightly blurred) rather similar in contrast? Is there an optimal (or a particularly suboptimal) addition? If found, are those specific to the subject, or relatively similar across different subjects?

## METHODS

### Optical System

A compact instrument has been developed with two collinear channels that allow simultaneous projection of two overlapping images on the

retina. A schematic diagram of the instrument is shown in Figure 1. The distances in the diagram are drawn to scale. The size of the set up is 325 × 275 mm. Two independent motorized Badal optometers allow control of the defocus level in each channel. Typically, one of the channels (channel 1, blue line, Fig. 1) is focused at far (i.e., compensating the subject's distance refraction), and the other (channel 2, red line, Fig. 1) is focused at near (i.e., simulating a near addition in a bifocal prescription). The focal length of the four Badal optometer lenses is 150 mm; therefore, a displacement of 10 mm produces a refraction/vergence change of 0.88 D in each channel. The two channels are combined by means of a thin double-mirrored plate. An artificial diaphragm placed at a conjugate pupil plane (P) allows us to limit the subject's pupil size to 4 mm. Visual stimuli were presented on a gamma-corrected CRT monitor (Mitsubishi Diamond Pro 2070; Mitsubishi Electric, Rydalmere, Australia) located 4.5 m from the conjugated pupil plane and controlled by a VSG card (Cambridge Research Systems, Rochester, UK). The mean luminance of the monitor is 100 cd/m<sup>2</sup> providing each channel with an effective luminance of 25 cd/m<sup>2</sup>.

For the purposes of the current study, channel 1 was used to correct distance refraction, providing a sharp image in best focus. Channel 2 was moved to create superimposed hyperopic or myopic defocused images, while keeping channel 1 fixed. As a result of the Badal optometer configurations, all powers refer to the pupil plane, not the spectacle plane. Figure 2 compares the simultaneous vision as achieved with, for example, a diffractive bifocal intraocular or contact lens (*top panels*) with that produced with the simultaneous vision simulator (*lower panels*). For far vision, the bifocal lens produces a sharp image of the far object, superimposed to a defocused near vision image (Fig. 2A). For near vision, the bifocal lens produces a sharp image of the near object, superimposed to a defocused far vision image (Fig. 2B). Conversely, the simultaneous vision simulator produces a myopic defocus (positive dioptric correction, which mimics a near addition) by channel 2, and a far sharp image in channel 1, allowing testing of the impact of a near addition on far vision (Fig. 2C). Also, a hyperopic defocus (negative dioptric correction) in channel 2 allows testing the impact of a defocused far image (Fig. 2D). For the purposes of this study, best focus in either channel is referred to 0 D, and the

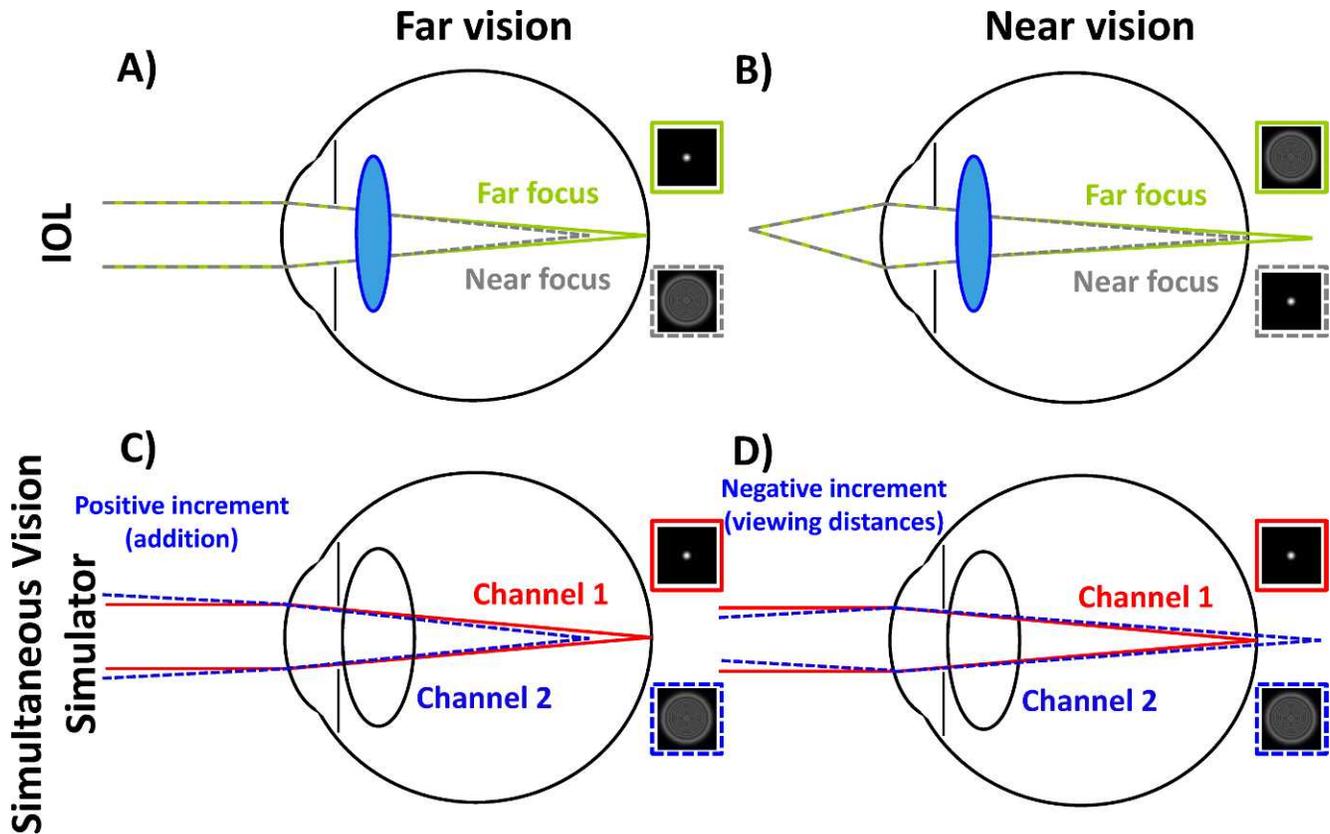


FIGURE 2. Schematic diagram of the near and far vision conditions produced by a bifocal intraocular (A, B) and those simulated in our study (C, D). PSFs and surrounding boxes represent the image projected in the retina by the rays with the corresponding color/line style.

addition is therefore defined as the refraction difference between channel 1 and channel 2.

### Simulations

The optical degradation produced by pure bifocal vision was computationally simulated. These simulations provide a reference for the subsequent experimental testing of the system by means of an artificial imaging system, to establish predictions on pure optical bases. Defocused images were generated using standard Fourier optics techniques.<sup>25</sup> The point spread function (PSF) for the corresponding levels of defocus was generated, and the simulated image targets (of different sizes and contrast) were obtained by convolution of the original targets with the corresponding PSFs using custom-developed routines written in a numerical computing environment (MATLAB; MathWorks, Natick, MA). Pure bifocal images were simulated by adding two images, one in focus and the other out of focus (i.e., the addition of two images, one in focus and the other defocused by 4 D, will represent a bifocal simultaneous vision image with an addition of 4 D). The superimposed images were normalized dividing by 2. For the purposes of this simulation, one of the added images was always in focus while the other varied in defocus from  $-4$  to  $4$  D in  $0.1$ -D steps. Simulations were performed for different letter sizes (5–50 arcmin) and 10 levels of contrast (white background, and letter luminance level ranging from 0–0.9 times the white level). The Michelson contrast (MC) inside the E-letter of the resulting superimposed images was then calculated.

### Experimental Measurements on an Imaging System

The contrast loss in the simultaneous images was experimentally measured through the system, to evaluate the pure optical degradation.

These experiments also served to test the system alignment and configuration (in the absence of the subject's aberrations) and were compared with computer simulations (Fig. 3) and to visual performance in subjects (Fig. 5). Images through both channels were projected on an artificial imaging system consisting on a scientific CCD camera (Retiga 1300, 16 bits; Qimaging, Surrey, BC, Canada) and a 100-mm  $f/3.5$  camera lens (Cosina, Nakano, Nagano Prefecture, Japan). The stimuli (presented on the CRT monitor and projected on the artificial imaging system's CCD through both channels) were Snellen E targets, similar to those in the computer simulations. Channel 1 was focused at far, and additions were achieved by moving the focus of channel 2 from  $-4$  to  $4$  D in  $0.1$ -D steps. Targets of different sizes (5–50 arcmin) and contrasts (black background, and white letters ranging from 0–0.9 times the white level of the monitor) were used. Monofocal control conditions were also tested, with high contrast (white on black) letters. Contrast degradation was estimated by computing the Michelson contrast inside the letter. The acquisition process was automatically controlled with custom software written in C++, and contrast calculations were obtained with custom routines programmed in a numerical computing environment (MathWorks).

### Subjects

Four subjects aged 28 to 42 years ( $34.3 \pm 3.4$  years) participated in the study. All subjects were experienced observers, trained in different psychophysical tasks. Subjects S1 and S2 were emmetropes, and subjects S3 and S4 were myopes requiring corrections of  $-3$  and  $-5.5$  D, respectively. No subject had clinically relevant astigmatism. Both myopic subjects performed the experiments wearing their usual monofocal contact lenses correcting their far vision. Monochromatic (720 nm) high order aberrations (HOA) were measured using a custom Hartmann-Shack aberrometer.<sup>19</sup> Root mean square wavefront error

(RMS) for HOA (4 mm) was 0.17, 0.27, 0.20, 0.22, for subjects S1 to S4, respectively. All subjects had undergone an ophthalmological evaluation before performing the experiments. Accommodation was paralyzed to simulate presbyopia (and to dilate the pupil) with periodic instillation of drops of 1% tropicamide.<sup>26</sup> Subjects signed a consent form approved by the institutional review boards after they had been informed on the nature of the study and possible consequences. All protocols met the tenets of the Declaration of Helsinki.

### Experimental Measurements in Subjects

While keeping channel 1 adjusted for subjective best focus at far, VA was measured for different amounts of defocus increments (i.e., additions) in channel 2 ranging from  $-4$  to  $4$  D. Measurements were performed in 0.5-D steps between  $-2$  and  $+2$  D of addition, and in 1-D steps between  $\pm 2$  and  $\pm 4$  D. Measurements with positive defocus increments in channel 2 represent far vision with different near additions superimposed, while measurements with negative defocus increments in channel 2 represent near vision in focus, in the presence of a defocused far image. A control condition was also tested by blocking channel 1, therefore testing vision in a monofocal condition, from  $-2$  D to  $2$  D (0.5-D steps) in channel 2. The subject adjusted his/her best focus while looking at an empty black square (45 arcmin, also used as fixation stimuli in between VA letters) in monofocal conditions (through one channel at a time).

VA measurements were performed for high-contrast (HC, MC = 1) and low-contrast (LC, MC = 0.33) targets, with white backgrounds. VA was measured using tumbling Snellen E letters in a four alternative forced choice procedure (4AFC) programmed in the Psychophysics Toolbox in the computing environment (MathWorks).<sup>27</sup> The procedure was followed until 16 reversals were performed, or 50 letters were presented. The average of the six last reversals was taken as the subject's VA for that condition. A total of 39 measurements of VA were performed, 15 for each one of the bifocal conditions (HC, LC) and nine for the monofocal condition. The three series of measurements (HC bifocal, LC bifocal, and monofocal) were randomized between subjects. Measurements in one subject lasted typically between 4 and 5 hours, and were performed in 1 day. Subjects were given breaks at their request.

## RESULTS

### Image Contrast with Simultaneous Vision from Simulations and Experimental Measurements

Measurements of image contrast degradation on a diffraction-limited imaging system acting as an artificial eye allowed us to test purely optical factors, in the absence of aberrations or neural factors. Figure 3 shows the normalized contrast of the targets imaged on the CCD of the artificial imaging system through both channels simultaneously, as a function of defocus in channel 2 (*blue lines*), as well as the contrast of the computer simulated targets (*black lines*). The curves shown in each panel represent data for targets of different sizes (equivalent to decimal VA ranging from 0-1). *Blue dashed lines* represent the standard deviation of normalized contrast values obtained for different initial contrasts on the images captured on the CCD ( $\pm 0.01$  on average). As expected, the curves from both simulations and experiments on the artificial imaging system were highly symmetric. The slight asymmetry observed in the experimental curves may arise from some minor aberrations in the experimental system.

As the relative impact of the optical aberrations of the system varied across letter sizes, the experimental values of contrast shown in Figure 3 were divided by a factor (ranging from 0.67 for a letter size equivalent to VA = 1.0-0.98 for a letter size equivalent to VA = 0.1) to match the contrast in

monofocal conditions (addition = 0 in the simulation and experiment). While for the largest letter tested (VA = 0.1) the experiment and simulations yielded similar contrast degradation, discrepancies in the absolute contrast degradation increased when decreasing the letter size, likely as a result of the contrast loss introduced by residual aberrations in the system.

In both experimental and simulated bifocal images, the contrast loss varied with the amount of addition. The maximum contrast was obtained in all cases for monofocal vision (zero addition). The minimum contrast was obtained for values of addition ranging between 0.5 and 2 D depending on the letter size, while contrast increased for the largest amounts of addition. The contrast had a notch of maximum degradation (23% with respect to the target contrast, on average across letter sizes) in the 0.5- to 2-D addition ranges (depending on the letter size), while contrast degradation was less than 15% (on average) in the 2.5- to 4-D addition range.

In all cases, contrast decreases rapidly when adding defocus (addition) in channel 2. After reaching a peak in degradation, the image contrast is partially recovered, as it increases with higher defocus values. This analysis allows estimating sets of additions producing the largest contrast degradation for each letter size, as shown in Figure 4 for simulations and experiments. The range of additions that produced the largest image degradation followed a similar trend in simulations and experiments: between 1.8 D (for the largest letter tested: 50 arcmin, VA = 0.1) and 0.3 D (for letters <10 arcmin, or VA > 0.5) for the experimental images, and between 2.1 and 0.3 D for the simulated images.

### Simultaneous Vision in Subjects

Figure 5 shows the individual measurements of VA for the four subjects of the study in three different conditions: HC and LC VA for bifocal vision and different additions (for channel 1 focused at far, and at different focus positions in channel 2); and HC VA for monofocal vision (blocking channel 1, and for different focus positions in channel 2). The 0- to +4-D addition range (*shaded green*) represents bifocal vision with far vision in focus and different near additions. On the other hand, the  $-4$ - to 0-D addition range (*shaded blue*) represents bifocal vision, with near vision in focus and the different additions representing different viewing distances. In this case, the superimposed blurred image is focused behind the retina. The monofocal condition represents a standard through-focus VA curve, for reference. Unlike the data obtained in the imaging system acting as an artificial eye, in general, the curves are less symmetric for positive and negative defocus, likely due to the presence of aberrations in the eye. Performance with LC stimuli tends to parallel, in most subjects, that for HC stimuli. Monofocal VA decreases steadily with defocus, as expected. In bifocal vision, VA decreases rapidly for small additions, typically reaches a minimum, and improves for larger additions. Bifocal VA in focus (with different superimposed additions) largely exceeds monofocal vision in the presence of equivalent amounts of defocus for most of the defocus range in all subjects.

Figure 6 shows average VA curves versus addition (for simultaneous vision) or defocus (for monofocal vision), averaged across subjects. At 0 D, monofocal and bifocal VAs are very close (monofocal slightly higher for S1, bifocal slightly higher for S2, and similar for S3 and S4)—despite the luminance in the bifocal condition being double (as it combines light from two channels)—to that of the monofocal condition. Differences in VA associated to this luminance difference are very minor, and likely within the experimental error (note, for example, that for the identical condition—

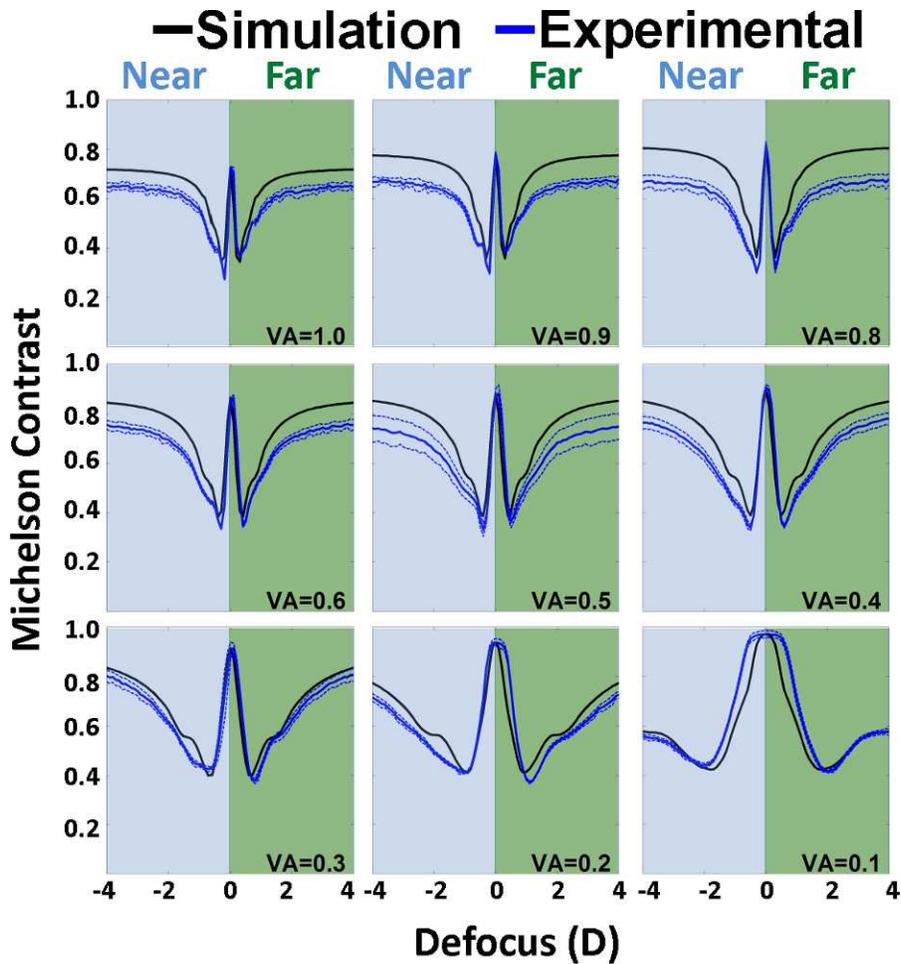


FIGURE 3. Comparison of the Michelson contrast obtained with computer simulations (black lines) and with the artificial imaging system (blue lines) as a function of the amount of addition. Each panel represents a different letter size (expressed in terms of the corresponding VA). Positive defocus (shaded green) represents far vision in focus in presence of a near defocused image (due to the addition). Negative defocus (shaded blue) represents near vision in focus (at different distances) in presence of a far defocused image. Blue dashed lines represent the standard deviation of normalized contrast values obtained for different initial contrasts on the images captured on the CCD ( $\pm 0.01$  on average). Data are for 4-mm pupils.

except for luminance—in the monofocal or bifocal measurement, the higher luminance shows higher VA in S2, lower VA in S1, and identical in S3 and S4). This is expected, as the dependence of VA on luminance for this luminance range (25–50 cd/m<sup>2</sup>) is minor, compared with that seen at lower

luminances.<sup>19</sup> Monofocal VA decreased systematically with defocus (from 1.05 at 0 D to  $0.35 \pm 0.04$  at  $\pm 2$ D, on average). For simultaneous vision, VA decreased when increasing addition with a minimum of  $0.66 \pm 0.06$  at  $1.69 \pm 0.25$  D (averaged for HC and LC across subjects), and then increased for higher additions ( $0.78 \pm 0.06$  for an addition of  $3.75 \pm 0.23$  D). While, on average, VA for monofocal vision decreased below 0.7 for defocus higher than 0.5 D, VA with bifocal corrections remained above 72% of the VA obtained under monofocal conditions for all the addition range ( $0 \pm 4$  D). Low contrast VA under simultaneous vision tended to parallel high contrast VA, with a relative average reduction of 32%.

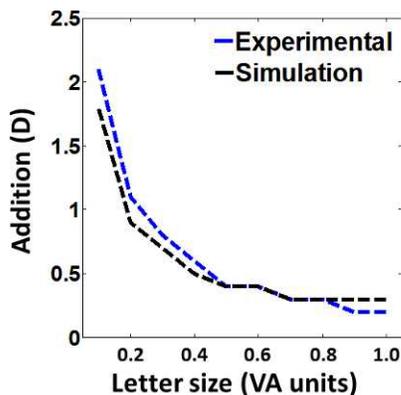


FIGURE 4. Additions that produced the maximum contrast reduction, as a function of letter size. Letter size is represented in VA units from 0.1 to 1 (equivalent to 50–5 arcmin).

### DISCUSSION

We have presented a new prototype to provide presbyopic (or simulated presbyopic) subjects with a visual experience of pure bifocal simultaneous vision.<sup>24</sup> The instrument is composed of two Badal systems, which allow providing simultaneously a correction for far vision and a near addition. Visual quality (and also optical quality) can therefore be measured simulating critical parameters of a simultaneous bifocal correction such as the amounts of near addition, as shown in this study.

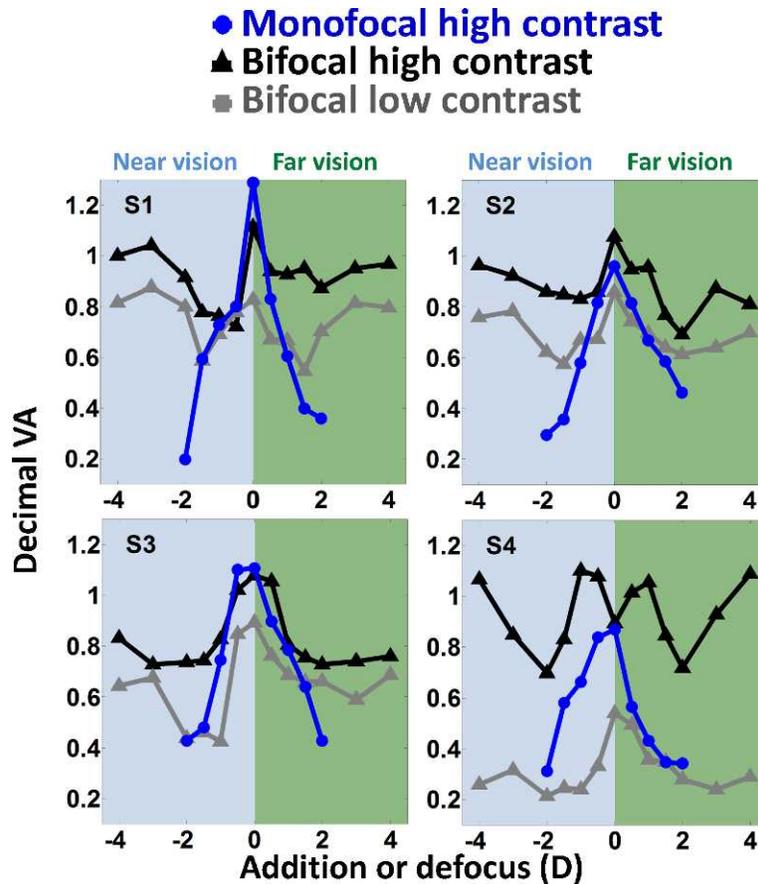


FIGURE 5. Decimal VA for different defocus increments in channel 2 (representing additions for bifocal images, and defocus for monofocal images): Bifocal HC VA curves (black line and symbols); Bifocal LC VA (gray line and symbols); monofocal through-focus HC VA (blue line and symbols). Each panel represents data for a different subject (S1 to S4). The 0 to +4 D addition range (shaded green) represents bifocal far vision with different near additions. The -4 and 0 D focus range (shaded blue) represents bifocal near vision at different distances. See text for a detailed explanation.

We have measured the impact of the presence of a near addition (of various amounts) on far VA, and alternatively the presence of a defocused far image on near VA (at various viewing distances). The experimental simulation of pure bifocal simultaneous vision on real subjects, without the limitations imposed by the practical implementation of the bifocal corrections (i.e., diffraction effects or abrupt transitions in refractive elements or chromatic effects in diffractive elements) has allowed isolation of visual degradation of simultaneous vision at different additions from other factors.

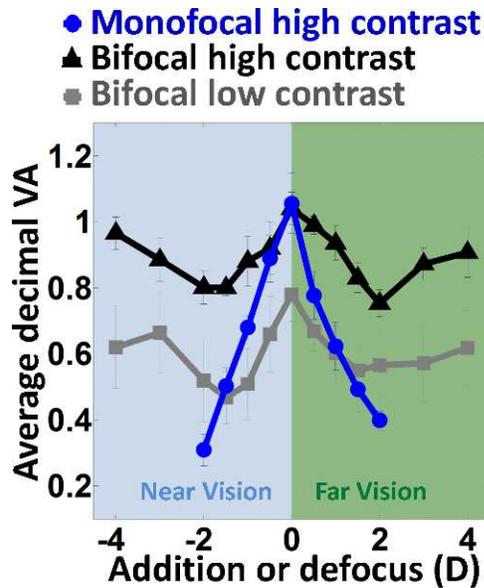
Simultaneous images (and even more in the context of this study) can be understood as the superposition of two channels, one in focus and the other one out of focus, that compete in the subject's visual system. It has often been argued that the perception of simultaneous images is driven by neural processes that are able to: first, separate both superimposed channels; and second, to suppress to some extent the blurred one while preserving the sharp one. Our study, however, points to a primary role of optical factors in visual performance with simultaneous vision.

We have found that additions in the 0.5- to 2.0-D range produce the strongest reduction of VA in subjects (and also the largest decrease of contrast in the simultaneous images), while larger additions decrease VA more moderately. Lower additions produce a closer superposition of two relatively high contrast images, resulting in a loss of resolution. Higher additions produce a higher blur in the second image, which creates a relatively gray uniform background, resulting in a loss of

contrast. Loss of resolution (as occurring with low additions) appears more deleterious to VA than loss of contrast (higher additions).

The fact that VA in our four subjects shows a similar trend to that obtained in computer simulations and optical experiments on a diffraction-limited lens suggests that the decrease in performance at low and intermediate additions is not driven by neural factors as much as by optical factors, and also that it is relatively unaffected by ocular aberrations. This result is of great interest, not only to increase our understanding of how simultaneous images are perceived, but also from a clinical point of view, as it allows identifying the acceptable addition values in a bifocal correction, which, according to our results, should avoid too low near additions.

Our results are consistent with those of Sanders et al. who showed that VA in multifocal contact lens wearers decreased steadily with the amount of addition imposed (1–2.5 D) from approximately 20/16 to 20/19.<sup>28</sup> The results are in contrast with those from another clinical evaluation of visual performance with soft bifocal contact lenses that showed that the lower the addition, the higher the VA for distance viewing conditions in a wide range of contrast conditions.<sup>29</sup> Unfortunately, both groups of subjects in Sanders et al. and in Cox et al. were presbyopic (18–25 and 23–31 years, respectively) and their accommodation was not paralyzed, which made the interactions between the multifocal designs used and the subject's accommodative response unpredictable.<sup>28,29</sup>



**FIGURE 6.** Average decimal VA in the presence of addition (for bifocal vision) or defocus (through-focus monofocal vision): Bifocal HC VA (black line and symbols) and Bifocal LC VA (gray line and symbols); Monofocal HC VA (blue line and symbols). Bifocal near vision at different distances is shown shaded blue. Bifocal far vision with different additions is shown shaded green. Error bars stand for standard deviations across subjects.

Our study is, to our knowledge, the first attempt to evaluate the loss in contrast and VA produced by bifocal vision, systematically and independently from a particular lens design. Although the system uses a refractive principle, these conclusions can be applied generally into the design of diffractive IOLs. Diffractive IOLs are available commercially with different focus shifts between near and far, and their performance is independent of the pupil diameter and does not rely on specific distributions of the refractive profile across the pupil. The study presented can be directly applied to the multifocal lens (Tecnis MF IOL; Abbott Medical Optics Inc., Abbott Park, IL) that has two distinct powers and the diffractive lens splits the energy into these powers equally.<sup>30</sup> Also by adjusting the ratio of the channel intensities, IOLs such as the Acri.LISA (Carl Zeiss), which sends 65% of the energy into the distance peak and 35% to near peak, could be simulated.<sup>31</sup>

The instrument can also be extended to allow simulations of the effect of different pupil regions for near and far, therefore expanding the range of bifocal solutions that can be simulated to include specific refractive IOL designs (i.e., concentric such as the ReZoom IOL by Abbot, or asymmetric such as the MPlus IOL by Oculentis GmbH). The different patterns can be achieved by the use of a transmission spatial light modulator and the use of polarizers to split the light for far and near across different pupillary regions into each channel. In general, a systematic simulation of a multifocal correction will produce insights on the visual performance under simultaneous vision, the visual tolerance to simultaneous vision, and the mechanisms for adaptation to simultaneous vision.<sup>32,33</sup>

Bi/multifocal contact lens prescription in clinical practice normally relies on a trial-and-error procedure with different designs until (if found) a design which appears satisfactory to the patient is prescribed.<sup>34,35</sup> The use of multifocal contact lenses has considerable scope to increase, as a large majority (63%) of presbyopic contact lens wearers still use monofocal lenses.<sup>36</sup> Undoubtedly, prescription of multifocal lenses would

benefit from increased knowledge of the visual response of patients to multifocal corrections, and from screening tools for practitioners that decrease trial-and-error approaches.<sup>36,37</sup> The method described in this study could be used to screen patients suitable to receive a multifocal correction and, more importantly, to identify the optimal design parameters, to prescribe the best-suited available bifocal solution, or to customize parameters to a patient. For the screening method, based on the system described in this work, a new set of protocols, different from the extensive psychophysical measurements of the current study, and more suitable to a clinical environment, has to be developed and validated.

## CONCLUSIONS

We present a prototype of an optical system that allows simulating noninvasively bifocal corrections in subjects (or artificial imaging systems).<sup>24</sup> With this new methodology, we showed that VA and contrast were reduced (7%–41%, depending on the condition) in simultaneous vision, both for far and near. The VA decrease found in all subjects is systematically highest for typical additions used in young presbyopic patients (1.5–2 D). The trends shown in VA and contrast as a function of the induced additions are important in the design of new protocols of adaptation for presbyopic subjects requiring low additions. Those trends are similar across subjects, indicating that suboptimal near additions are relatively independent of the specific aberrations and neural factors in subjects. The simultaneous vision instrument presented in this study has proven to be an excellent tool to simulate bifocal vision and to increase our understanding of multifocal solutions for presbyopia. Future work will be directed to explore more sophisticated designs by controlling the shape and areas of the pupil that will be used for near or far vision and to expand this system to be able to measure under binocular and monovision conditions.

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