Factors affecting image magnification in indirect ophthalmoscopy with Volk or similar lenses and a biomicroscope

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Abstract
Purpose: To explore the factors affecting the linear magnification of the intermediate fundus image during indirect ophthalmoscopy with a slit-lamp biomicroscope.

Methods: A simple paraxial model, based on a ‘reduced’ eye and a ‘thin’ ophthalmoscopy lens, is used to develop equations showing the effects of the power and ametropia of the eye, and the equivalent power and position of the ophthalmoscopy lens on fundus magnification. Predicted magnifications are compared with practical results found in earlier published experimental studies, which used Volk ophthalmoscopy lenses in conjunction with physical model eyes with adjustable levels of axial ametropia.

Results: The model’s magnification predictions, as a function of the eye’s ametropia, are in good agreement with previous experimental measurements, provided that the equivalent powers of the Volk lenses are used rather than their labelled nominal powers. Magnification values typically change by approximately ±10% over the practical range of each parameter if other parameters are held constant. In particular, normal variations in the equivalent power of the adult emmetropic eye result in magnifications which range from about 90–120% of the nominal value given for an eye power of +60.00 D. It is demonstrated that the recommended working distances for different powers of Volk ophthalmoscopy lenses approximate optimal matching between the various pupils of the eye-Volk lens-slit-lamp biomicroscope system.

Conclusions: All the parameters considered have marked effects on magnification. The magnification values quoted by manufacturers can be regarded as only approximations of those which may be found in practice. Better estimates of magnification can be obtained by inserting the appropriate parameter values into the equations derived in this paper, using, where appropriate, the equivalent power of the indirect ophthalmoscopy lens, rather than the lens’ labelled, ‘nominal’ power.

Keywords: ametropia, equivalent power, indirect ophthalmoscopy, magnification, Volk lenses, working distance
INTRODUCTION

Following Helmholtz’s development and popularisation of his direct, hand-held ophthalmoscope, indirect examination became a standard part of ophthalmic examinations. Later, factors such as the direct ophthalmoscope’s limited field-of-view (especially for patients with high myopia), the need to work very close to the patient and the associated discomfort for both the patient (bright light) and the examiner (postural and dexterity considerations), resulted in the increasing adoption of indirect ophthalmoscopy. This technique uses a hand-held, positive ophthalmoscopy (condensing) lens to form an intermediate image, which is then viewed either directly by the observer or through some form of monocular or binocular eyepiece system. All these variants offer several advantages over the direct method, including a larger field-of-view and a comfortable working distance between the patient and examiner. The binocular system has the additional merit of giving the observer a stereoscopic view.

The rise of contact lens-related practice led to an increase in the widespread availability of binocular slit-lamp microscopes (biomicroscopes) in the clinical setting. These devices can act as an alternative eyepiece system to view the image formed by various types of ophthalmoscopy lens. Early developments in such lenses are well described by Hruby. One of the first, designed by Koepe, was a lens that would come into direct contact with the patient’s cornea. The back-surface curvature of Koepe’s lens was designed to match that of the anterior corneal surface, while the front surface was flat, so that the whole refractive power of the eye was approximately neutralised. This allowed the fundus to be viewed directly by the microscope. Hruby himself introduced the idea of using a high-powered negative lens (−55 dioptres [D]) held close to the cornea to produce an erect intermediate image which lay internal to the eye but within the focusing range of the observing microscope. Numerous other positive and negative lens designs, some with multiple components and offering wider fields-of-view, have followed.

One important aspect of any design of ophthalmoscopy lens is the transverse linear magnification of the retina that it provides in the intermediate image (see h, in Figure 1), since this affects judgements of the lateral dimensions of any observed retinal lesions or other fundus features. The final magnification is, of course, further increased by that of the eyepiece system or biomicroscope which is used to observe this intermediate image.

Manufacturers usually specify a nominal magnification that a particular ophthalmoscopy lens provides. In practice, however, it is clear that factors such as the distance at which the lens is positioned with respect to the cornea (d, in metres [m]), and, since the eye forms an integral part of the overall ophthalmoscopic optical system, the power and ametropia of the patient’s eye must also have some influence on the actual magnification achieved. Using model and cadaver eyes, various authors have demonstrated that this is indeed the case.

Since the human eye and ophthalmoscopy lens are both ‘thick’, possibly aspheric lens systems, whose optical axes may be laterally separated and tilted with respect to one another, sophisticated optical modelling is ideally required for accurate determination of the impact of changes in various parameters on magnification. However, useful insights into the effects of patient ametropia and lens position on magnification can be gained by using paraxial optics, with the patient’s eye modelled as a ‘reduced’ eye with a single refracting surface, the ophthalmoscopy lens as a ‘thin’ lens and the optical elements having a common optical axis. Henson and Rabbetts presented partial analyses of this type, but their derived magnifications are the angular magnification for an observer directly viewing the image of the fundus of an emmetropic eye formed by the ophthalmoscopy lens under specified conditions, rather than the transverse magnification between the fundus and the image plane for the ophthalmoscopy lens. Raasch used a more sophisticated schematic eye model to consider the effects of a patient’s ametropia or aphakia on the angular magnification experienced during visual observations with various types of direct and indirect ophthalmoscopy.

During quantitative indirect ophthalmoscopy using a biomicroscope, it is the linear size of the image of any retinal detail formed by the ophthalmoscopy lens that is usually measured. The clinical significance of measurements of
this type can only be assessed properly if the magnification of this image is accurately known. A more extensive analysis of the factors affecting the transverse magnification in such images is now presented, using a paraxial reduced-eye model and a ‘thin’ lens approximation for the ophthalmoscopy lens, as described earlier. Factors such as the power of the patient’s eye and its spherical ametropia, together with the power and longitudinal position of the ophthalmoscopy lens from the cornea, are included as variables. The results of this modelling are then compared with previously published practical measurements of the magnification of some popular Volk ophthalmoscopy lenses. Finally, further examples are given of the effects of changes in individual variables. Our analysis shows the substantial impact that these variables can have on magnification and helps to clarify some of the properties of the Volk lenses studied.

**METHODS**

**Basic components of the optical model**

Figure 1 displays the basic optical set-up. The patient’s eye, with relaxed accommodation, is modelled as a reduced eye with its principal points (P and P’) and entrance pupil (E) at the cornea. It has an axial length of \( k' \) (in m), corresponding to a dioptric length of \( K' \) (in D), calculated as \( n'/k' \), where \( n' \) is the homogenous refractive index of the media. Following convention, \( K' \) is taken as being positive. The power of the single refracting surface of the eye is \( F_E \) (in D) and the eye’s spherical ametropia is \( K \) (i.e., the power of the correction required for distance vision: positive for a hyperope, negative for a myope, in D). Applying the traditional vergence formula to the corneal refraction gives, \(-K' + F_E = -K\), such that:

\[-K' = -(F_E + K) \quad (1)\]

It is assumed that the ophthalmoscopy lens is thin and has a power, \( F_O \) (in D). It is placed at a distance, \( d \) (in m), away from the cornea. Note that since both principal points of the reduced eye lie at the corneal vertex and those of the thin lens coincide at its optical centre, \( d \) is the distance between these pairs of principal points. If the predictions of the model are to be compared with results obtained from human eyes combined with actual ‘thick’ lenses, then the value of \( d \) used in the latter situation must be the distance between the first principal points of the eye (located in the anterior chamber, approximately 1.8 mm behind the cornea) and thick ophthalmoscopy lens, rather than being equated to the manufacturers’ working distance between the adjacent surfaces of the cornea and ophthalmoscopy lens.

**Transverse magnification of the image formed by the ophthalmoscopy lens**

Referring to Figure 1, the overall magnification, \( M \), is the product of the magnifications produced first by the eye \( (M_1) \) and then by the ophthalmoscopy lens \( (M_2) \), such that:

\[ M = M_1 \times M_2 = \left( \frac{L_1}{L_1'} \right) \times \left( \frac{L_2}{L_2'} \right) \]

where \( L_1 \) and \( L_1' \) are, respectively, the object and image vergences for the cornea (treating the fundus as the object) and the corresponding quantities with a subscripted ‘2’ refer to the refraction at the ophthalmoscopy lens.

Evidently, when the retina is imaged via refraction by the cornea, \( L_1 = -K' \) and \( L_1' = -K \), therefore:

\[ M = \frac{L_2}{L_2'} \]
The initial image distance from the cornea, $l_1'$, is $-1/K$, such that its distance from the ophthalmoscopy lens, $l_2$, is $([-1/K] - d)$, which can be re-written as:

$$l_2 = \frac{-(1 + dK)}{K}$$

Hence, the object vergence for the ophthalmoscopy lens, $L_2$, is:

$$L_2 = \frac{-K}{(1 + dK)}$$

The image vergence, $L_2'$, for this lens would be:

$$L_2' = \frac{-F_0}{(1 + dK)}$$

which can be re-expressed as:

$$L_2' = \frac{\left(\frac{K'}{K} - \frac{F_0}{(1 + dK)}\right)}{(1 + dK)}$$

The magnification of the ophthalmoscopy lens, $M_2$, is therefore:

$$M_2 = \frac{L_2}{L_2'} = \frac{-K}{\left(\frac{K'}{K} + \frac{F_0}{(1 + dK)}\right)}$$

Hence, the overall magnification is:

$$M = M_1 \times M_2 = \left(\frac{K'}{K}\right) \cdot \left(\frac{-K}{\left(\frac{K'}{K} + \frac{F_0}{(1 + dK)}\right)}\right)$$

which simplifies further to:

$$M = \frac{-K'}{\left(\frac{K'}{K} + \frac{F_0}{(1 + dK)}\right)}$$

Ansari-Shahrezaei et al.\textsuperscript{14} quote an essentially similar equation, but without derivation.

Substituting Equation 1 into the numerator facilitates the elimination of $K'$ whilst simultaneously introducing $F_E$:

$$M = \frac{-(F_E + K)}{\left(\frac{K'}{K} + \frac{F_0}{(1 + dK)}\right)}$$

which, conveniently, can be re-expressed as:

$$M = \frac{-F_E \left(1 + \frac{K}{F_E}\right)}{F_0 \left(1 + dK - \frac{1}{F_0}\right)}$$

Factorising the above expression for the term $\frac{F_E}{F_0}$, and for $K$ in the denominator produces:

$$M = \frac{-\left(F_E \left(1 + \frac{K}{F_E}\right)\right)}{F_0 \left(1 + (K \left(d - \frac{1}{F_0}\right))\right)}$$

Note that Equation 4 is non-linear; but, if relatively modest levels of ametropia ($-10 \text{ D} < K < +10 \text{ D}$) are present, $F_E = +60.00 \text{ D}$, and the ophthalmoscopy lens is of a relatively high power ($F_O > +40 \text{ D}$), then it is valid to expand the numerator and denominator of the second bracketed term using the binomial expansion and to retain only the first-order terms of the expansion. This leads to the approximation:

$$M = \frac{-\left(F_E \left(1 + \frac{K}{F_E}\right)\right)}{F_0 \left(1 + (K \left(d - \frac{1}{F_0}\right))\right)}$$

which further approximates to:

$$M = \frac{-\left(F_E \left(1 + \frac{K}{F_E}\right)\right)}{(1 + (K \left(d - \frac{1}{F_0}\right)))}$$

Equation 5 suggests that the dependence of magnification ($M$) on the patient's ametropia ($K$) is almost linear over a range of the order of 20.00 D, centred on emmetropia. If the eye is emmetropic ($K = 0.00 \text{ D}$), then Equation 4 and Equation 5 both collapse to simply:

$$M = \frac{-\left(F_E \right)}{F_0}$$

Authors sometimes quote Equation 6 as representing the 'magnification' of an ophthalmoscopy lens, often assuming a value of $+60.00 \text{ D}$ for the power of the emmetropic eye;\textsuperscript{11,12} however, the qualifying comments, that the equation is only valid if the eye is emmetropic, and when equivalent powers of the emmetropic eye and ophthalmoscopy lens are used, are often forgotten.

### Position of image formed by ophthalmoscopy lens

From a practical point of view, the distance of the image from the ophthalmoscopy lens, $l_2' (= 1/L_2')$, is of some importance. For example, the range of usable values of $l_2'$ is constrained by the geometry of the head- and chin-rest and viewing biomicroscope, since the fundus image must lie within the focusing range of this instrument. A more important constraint is that optimal conditions for the illumination and the greatest field-of-view of the fundus image are obtained when the eye's pupil is imaged in the region of the entrance pupils of the viewing slit-lamp's biomicroscope (also known as pupil matching). The exit pupil of the illuminating system should also be imaged at the eye's pupil.\textsuperscript{5} These aspects will be discussed later, in more detail. As $l_2' = 1/L_2'$, simple inversion of Equation 2 gives:

$$l_2' = \frac{(1 + dK)}{\left(\frac{K}{F_E} + \left(F_0 \left(1 + dK\right)\right)\right)}$$

Note that the image lies at infinity when the denominator of Equation 7 has a value of zero. i.e., when
\((\frac{-K}{d} + (F_O(1+dK))) = 0\). Rearranging this formula, for the term \(F_O\) results in the general effectivity equation:

\[
F_O = \frac{K}{(1+dK)}
\]  
(8)

**RESULTS**

**Comparison with experimental data obtained with a model eye**

Although, as noted earlier, the equations provided above involve major simplifications, we believe they show that magnification with ophthalmoscopy lenses varies substantially with factors such as lens position \(d\) and the patient’s ametropia \(K\). Before this is explored in detail, it is helpful to demonstrate that the magnification equations derived above are consistent with available experimental evidence.

Predicted magnification values can be compared with those measured experimentally by Ansari-Shahrezaei et al.\(^\text{14}\) for various labelled powers of ‘classic’ Volk lenses (i.e., +60 D, +78 D and +90 D). These lenses (Volk Optical, volk.com) are typically symmetrical in their design, with deeply-aspheric surfaces.\(^\text{22,23}\) In their studies, Ansari-Shahrezaei et al.\(^\text{14}\) used a sophisticated physical model of the eye, similar to those described by both Rassow\(^\text{24}\) and Rudnicka and colleagues.\(^\text{25}\) Ansari-Shahrezaei et al.\(^\text{14}\) water-filled model eye had a power of +59.40 D. The axial length could be accurately adjusted to produce different values of axial ametropia.\(^\text{25}\) A further study of the +90 D lens by the same group\(^\text{15}\) produced essentially the same results as their preceding work.\(^\text{14}\) Magnification was determined by matching the dimensions of the image of a scale on the retina to the width of the slit-lamp beam. The variation in transverse magnification of the fundus image with ametropia and other factors does not affect the dimensions of any slit image which might be compared with that of a fundus feature to determine its dimensions.\(^\text{14,15}\) This is because the slit-imaging light makes a double passage through the eye/ophthalmoscopy lens system, whereas the imaging beam for the fundus only passes through once. The outgoing light from the slit projection system places a first outgoing slit image, of known dimensions, at the viewing microscope’s plane of focus. If the reflected light from the fundus is also in focus in this plane, then the projected slit is next re-imaged onto the fundus at magnification \(1/M\), where \(M\) is the magnification between the fundus and microscope’s plane of focus. A further magnification of \(M\) occurs between this retinal slit image and the final outgoing reflected image in the microscope’s plane of focus. The overall magnification between the outgoing and incoming slit images is then \(1/M \times M = +1\times\), and the reflected slit image in this plane is the same size as the original outgoing projected image. Thus, variations in the magnification of fundus features with variations in ametropia and/or the ophthalmoscopy lens’ distance from the cornea are not compensated for by corresponding variations in the magnification of the slit image.

The distances of the ophthalmoscopy lenses from the model eye’s cornea were not specified by Ansari-Shahrezaei et al.\(^\text{14}\) however, we assume that they were close to those recommended by the lens’ manufacturers. For initial comparison of the experimental results with our theoretical predictions, a value of \(d = 1/F_O\) was used, so that the expected magnifications were derived from a simplified version of Equation 4:

\[
M = -\left(\frac{F_E}{F_O}\right) \cdot \left(1 + \frac{K}{F_E}\right)
\]  
(9)

Equation 9 is a straight line with a slope of \((-1/F_O)\) magnification units/D of ametropia. The values used for \(F_O\) were the manufacturer’s labelled ‘nominal’ powers and \(F_E\) was taken as +60.00 D. The results of the comparison are presented in Figure 2.

Equation 9 successfully predicts the general trends of the results, but the rates of change of magnification with ametropia differ between the experimental and predicted cases for each lens power, and the predicted magnitudes are always too low. A particularly important case is the magnification for an emmetropic eye \((K = 0.00 D)\). The various relevant values are shown in rows 2 to 4 of Table 1. Although the experimental values (row 3) agree with those specified by the manufacturer (row 2), the theoretically-predicted values \((M = -[F_E/F_O])\), row 4) do not. These predicted values are obviously systematically incorrect.

Several factors, such as the possible use of slightly different values for the lens’ position \(d\), differences in the definition of the ametropia (such as referring refractive error to the spectacle plane rather than to the corneal vertex), use of slightly different values for \(F_E\) and wider-field distortion effects in the ophthalmoscopic image which invalidate the paraxial approximations used for our predictions, could all produce small differences between the experimental and predicted results shown in Figure 2 and Table 1.

We believe, however, that the major factor responsible for the relatively large discrepancies observed is in the way that the manufacturer assigns the labelled ‘nominal’ powers to the Volk lenses studied. In his 1986 patent, Volk states, “… the nominal power of each lens being the sum of the dioptric power of its two identical surfaces …”. Thus the lenses with nominal powers of +60 D, +78 D and +90 D have identical pairs of surfaces with powers of +30 D, +39 D and +45 D, respectively. Since each lens has a thickness of several millimetres (mm), and the paraxial equivalent power is given by the traditional ‘thick’ lens formula:

\[
F_O = F_1 + F_2 - \left(\frac{t}{n}\right) \times F_1 \times F_2
\]  
(10)

where \(F_1\) and \(F_2\) are the two surface powers (in D), \(t\) is the axial thickness (in m), and \(n\) is the lens’ refractive index, then the
equivalent lens powers, which must be used in Equations 4, 6 and 9 etc., will always be smaller than the corresponding ‘nominal’ powers.

Fortunately, it is possible to estimate the Volk equivalent powers from the limited data provided by the manufacturer. Using their magnification values, the equivalent power of each lens can be deduced on the assumption that the magnification quoted by the manufacturer is that for an emmetropic eye, whose \( F_E = +60.00 \) D, and is derived from Equation 6. Thus, by rearranging this equation, we expect that the equivalent power for the ophthalmoscopy lens should be given as \( F_O = -\left( \frac{F_E}{M} \right) \), where \( F_E = +60.00 \) D. The fifth row of Table 1 gives the results of this procedure. As expected, for these thick, bi-convex lenses the equivalent powers are all smaller than the corresponding labelled nominal powers, which merely represent \( F_1 + F_2 \) (i.e., the nominal powers assume that the lens is ‘thin’ in its form).

An independent check on the correctness of one of these deduced powers is possible. Although few data on the optical parameters of Volk lenses have been published, in his 1994 patent Volk does quote some values for a ‘classic’ lens of nominal power = +90 D. Volk states that the apical radii of the two surfaces are both 11.622 mm, \( t = 8.763 \) mm and \( n = 1.523 \). Using \( F = \frac{(n – 1)t}{r} \) for the surface powers in air, we find \( F_1 = F_2 = +45 \) D, totalling to the labelled +90 D as expected, and that the equivalent power (using Equation 10) \( F_O = +78.35 \) D. The first principal (P) and nodal (N) points are each located within the lens, at a distance of \( \frac{t}{n}F_2/F_O = +3.305 \) mm away from the lens’ front surface. Considering that the manufacturer’s magnification value for this lens (−0.76x) is only quoted to two significant figures, the equivalent power of this lens, as derived from these parameter values (+78.35 D), agrees well with that derived from the magnification value in Table 1 (+78.9 D).

**Table 1** Magnification produced by three Volk lenses of different labelled powers (+60 D, +78 D and +90 D) with an emmetropic eye

| Labelled ‘nominal’ Volk lens power (D) | +60 | +78 | +90 |
|--------------------------------------|-----|-----|-----|
| Manufacturer’s claimed magnification (assuming \( F_E = +60.00 \) D) | −1.15x | −0.93x | −0.76x |
| Experimental magnification (assumes \( F_E = +59.40 \) D) | −1.12x | −0.93x | −0.74x |
| Predicted magnification using nominal lens powers given in row 1 (assumes \( F_E = +59.40 \) D) | −0.99x | −0.76x | −0.66x |
| Equivalent power required to give the manufacturer’s claimed magnification value (D) (assumes \( F_E = +60.00 \) D) | +52.2 | +64.5 | +78.9 |
| Second equivalent focal length, \( f_{O'} \) (mm) | +19.2 | +15.5 | +12.7 |

Note: The penultimate row presents the Volk lens’ equivalent powers necessary to give the manufacturer’s magnification value by rearranging Equation 6, to: \( F_O = -\left( \frac{F_E}{M} \right) \), where \( F_E = +59.40 \) D.

**Figure 2** Image magnification as a function of axial ametropia. Symbols represent the experimental values found in the model eye-based study of Ansari-Shahrezaei and co-workers. Lines represent the corresponding predictions using Equation 9 (i.e., assuming that \( d = 1/F_O \)), when the values of the powers of the Volk lenses are the ‘nominal’ values supplied by the manufacturer (i.e., +60 D, +78 D and +90 D), and \( F_E = +59.40 \) D for the model eye.
Volk lens magnifications as a function of axial ametropia, then the revised comparison of experimental versus predicted magnification results, presented in Figure 3, is obtained. In this analysis, it has been assumed that the value of $d$ used in Equation 4 is the sum of the manufacturer’s ‘recommended’ working distance and the distances of the relevant principal planes from the cornea and ophthalmoscopy lens’ surfaces (i.e., $d$ would = 19.8 mm, 15.2 mm and 11.9 mm, for the +60 D, +78 D and +90 D Volk lenses, respectively). Finally, $F_E$ was fixed at +59.40 D.14

Clearly, while some discrepancies remain between the experimental and predicted values, these are generally quite small. If proportional differences are considered, these exceed 7.5% only for high magnitudes of ametropia (≥10 D), as shown in Figure 4. It seems reasonable to attribute at least part of these remaining discrepancies, which appear to vary systematically rather than randomly, to the uncertainties in the lens position used in the experimental investigation by Ansari-Shahrezaei et al.14 and the other factors mentioned earlier.

### Influence of individual variables on magnification

Having demonstrated the broad validity of our model, it is helpful in relation to the interpretation of clinical findings of the size of fundus features to consider the way in which changes in the values of individual parameters influence the image magnification. Evidently, numerous combinations of the various relevant parameters are possible, potentially leading to a wide range of magnification values. To illustrate the magnitude of the effects which might be expected, Figures 5–8 give examples of the results, as derived from Equation 4, of varying specific parameters, while the other parameters are given fixed, typical values. To allow comparison with possible clinical results, the equivalent powers of the ophthalmoscopy lenses used in several of the calculations are +52.2 D, +64.5 D and +78.9 D, i.e., they match those of the three Volk lenses studied in the previous section (whose nominal powers were +60 D, +78 D and +90 D, respectively).
Spherical ametropia \((K)\)

Since, in our simple model, emmetropia only demands equality between the dioptric length of the eye, \(K'\), and its power, \(F_E\), any specific level of ametropia might arise from an ‘abnormal’ value of either or both of these variables. For our present purposes we will initially assume that the ‘normal’ default values are both +60.00 D; that ‘axial’ myopia results from \(K' \neq +60.00\) D due to changes in axial length, \(k'\), combined with a normal +60.00 D value for \(F_E\); and that ‘refractive’ ametropia \((F_E \neq +60.00\) D, \(K' = +60.00\) D) is due to changes in corneal curvature. Refractive ametropia due to a change in refractive index (which affects both \(K'\) and \(F_E\)) is not considered here.

Note again from Equation 4 that for the case where the eye is emmetropic \((K = 0.00\) D), the overall magnification is simply \(-\left(\frac{F_E}{F_O}\right)\). In the examples shown in Figure 5, the effects of the patient’s ametropia on magnification are shown with each of the three ‘classic’ Volk lenses under the alternative assumptions of ‘refractive’ and ‘axial’ origins for the ametropia. Equation 4 has been used together with the same \(d\) values as described earlier (i.e., \(d = 19.8\) mm, 15.2 mm and 11.9 mm for the +60 D, +78 D and +90 D Volk lenses, respectively). It can be seen that, although in the axial case the magnitude of the magnification is greater in hyperopic eyes and smaller in myopic eyes, changing the degree of refractive ametropia has substantially less effect. For the parameter combinations illustrated, magnification...
levels for higher levels of axial ametropia (|K| ≈ 10 D) may differ by more than 15% from the emmetropic values.

Equivalent power of the ophthalmoscopy lens (FO)

Figure 6 illustrates that, as is well known, the magnitude of the magnification diminishes as the power of the positive ophthalmoscopy lens increases, being approximately inversely proportional to this power (see Equation 4). We have provided the results for emmetropia and two levels of axial spherical ametropia (±5.00 D). It is assumed that the power of the eye, $F_E$, remains constant at +60.00 D, such that the ametropia arises as a result of changes in the axial length, $k'$. All lenses are positioned so that $d = 1/F_O$, facilitating the use of Equation 9. The inverted image is always largest in the hyperopic case and smallest in the myopic case.

In the case of refractive ametropia, when $d = 1/F_O$, Equation 4 still simplifies to Equation 9. Furthermore, the magnification can be expressed as:

$$M = -\left(\frac{60.00 - K}{F_O}\right) \left(1 + \frac{K}{60.00 - K}\right)$$

because $F_E = K' - K$, where the dioptric length $K' (= n'/k')$ remains unchanged at +60.00 D.

Upon further simplification, this yields a magnification of $M = -(60.00 D/F_O)$, which is independent of the value of the eye’s refractive ametropia (K) and effectively equals that of the +60.00 D emmetropic eye.

FIGURE 6 Effects of changing the positive equivalent power of the ophthalmoscopy lens on image magnification. It is assumed that $F_E = +60.00$ D and that $d = 1/F_O$. The three curves correspond to differing levels of ‘axial’ refractive error, of −5.00 D, 0.00 D and +5.00 D. In the case of refractive ametropia the changes are always the same as in the emmetropic case.

FIGURE 7 Image magnification as a function of the distance, $d$ (in mm), from the corneal vertex for three Volk ophthalmoscopy lenses of nominal powers +60 D, +78 D and +90 D. For each lens, results for eyes with −5.00 D, 0.00 D and +5.00 D of axial refractive errors are presented. In each case, the eye is assumed to have a power of $F_E = +60.00$ D.
Lens’ position from the eye (d)

In the simple reduced-eye model, if the eye is ametropic and the ophthalmoscopy lens is in contact with the cornea (i.e., d = 0 m), then the magnification, from Equation 4, becomes:

\[ M = - \left( \frac{F_E}{F_O} \right) \cdot \left( 1 + \frac{K}{F_E} \right) \left( 1 + \frac{1}{F_O} \right) \]

If it is assumed that K is small, in comparison to both \( F_E \) and \( F_O \), then we can approximate that the overall magnification is:

\[ M = - \left( \frac{F_E}{F_O} \right) \cdot \left( 1 + \frac{K}{F_E} \right) \]

which further simplifies to:

\[ M = - \left( \frac{F_E}{F_O} \right) \cdot \left( 1 + \frac{K}{F_E} + \frac{1}{F_O} \right) \]

It can be seen that, for both ‘refractive’ and ‘axial’ ametropia, if both \( F_E \) and \( F_O \) are positive (i.e., when a Volk-type lens is used rather than a negative Hruby lens), this means that the image is larger if \( K \) is positive (hyperopia) and smaller if \( K \) is negative (myopia). A positive value for \( F_O \) also implies that the image produced by the ophthalmoscopy lens will be inverted.

As noted earlier, a special case occurs when the first equivalent focal point of the ophthalmoscopy lens coincides with the principal plane at the cornea, here, Equation 4 will reduce to Equation 9, which is a straight line with a slope of \(-1/F_O\) magnification units/D of ametropia.

Figure 7 illustrates the general case using Equation 4 and a fixed \( F_E \) of +60.00 D. Results are presented for the three ‘classic’ Volk lenses of nominal powers +60 D, +78 D and +90 D; but, again, their equivalent powers have been used rather than the nominal powers. For each lens power, the cases of −5.00 D, 0.00 D and + 5.00 D of axial refractive error have been plotted.

Figure 7 shows that, for any positively-powered ophthalmoscopy lens, increasing the distance of the lens from the eye has no effect if the eye is emmetropic, but this decreases the magnitude of the magnification for the axial hyperope, yet increases it for the axial myope. A similar behaviour was observed in Ansari-Shahrezee et al.’s experimental model-eye studies with Volk lenses. The magnification provided by an ophthalmoscopy lens is typically found to be independent of the degree of axial ametropia (and thus equivalent to that of an emmetrope) at a specific position, which, as can be seen from Equation 4 and Figure 7, is achieved when \( d = 1/F_E + 1/F_O \). In ametropic eyes, the departure of the magnification, away from its fixed emmetropic value, varies with the distance of the ophthalmoscopy lens from this position.

Equivalent power of the eye (\( F_E \))

So far in evaluating our equations, it has often been assumed, following many earlier authors, that the ‘normal’ overall power of the eye, \( F_E \), is +60.00 D. A major problem affecting the prediction of magnification in clinical practice is that usually we do not know the value of \( F_E \) for any real eye with a given level of ametropia. In reality, emmetropic eyes and those at any level of ametropia can be found within a range of values of \( F_E \) as a result of their biometric characteristics.
parameters being appropriately in scale with one another (in the simple reduced-eye model, emmetropia requires only that \( K' = F_E \)). The total range of \( F_E \) found in adult human emmetropic eyes extends over approximately 15 D. Axial lengths, \( x \), in human emmetropes are found to vary over a range from about 21 mm to 26 mm, corresponding to a value of \( F_E \) between about +70.00 D and +55.00 D. In the case of both real and reduced emmetropic eyes \( (K = 0.00 \text{ D}) \), the predicted magnification is simply \( -\frac{K}{F_E} \), so that for any given ophthalmoscopy lens, the magnification is smaller for longer, lower-powered eyes and higher for shorter, higher-powered eyes. This means that if, for example, a +78 D Volk ophthalmoscopy lens (with an equivalent power of \( F_O = +64.5 \text{ D} \)) was used on an adult emmetropic eye, then the magnification might lie anywhere between approximately \(-0.85x\) and \(-1.09x\) (note again that the manufacturer's figure for magnification, \(-0.93x\), appears to be derived assuming an \( F_E \) value of +60.00 D).

The relative departure in magnification from its nominal value, as specified by the manufacturer, depends upon the power of the ophthalmoscopy lens used and may approach around 20%, as shown in Table 3. Since a similar spread in the values of axial length occurs at any specific level of human ametropia (although the mean length will vary systematically with the value of the ametropia), a similar uncertainty in the predicted level of magnification occurs with ametropes. If a more precise estimate of magnification is required for a particular adult human eye, whose ametropia is known, additional information must first be obtained to allow its equivalent power to be estimated. This might be achieved with adequate accuracy by using the methods devised by Bennett et al. for determining the sizes of retinal features in fundus photography with a retinal camera. In this method, ocular powers are estimated on the basis of an individual patient’s measured corneal radius, axial length and ametropia. Bennett gives an appropriate computing scheme to determine the equivalent power of the eye from these values with an accuracy which is adequate for clinical purposes. A simpler approximation, based on a more sophisticated schematic eye model, requires only the eye’s ametropia \( (K, \text{ in D}) \) and a measurement of its axial length, \( x \) (expressed in mm), together with the assumption that the second principal point \( (P') \) of the eye is located 1.82 mm behind the corneal apex. The equivalent power of the ametropic eye \( (F_EQ, \text{ in D}) \) is then approximated by:

\[
F_EQ = \left( \frac{1336}{x - 1.82} \right) - K
\]

Power estimates obtained using the simpler method differ by only about \(-0.4 \pm 0.7\%\) (mean \pm standard deviation) from those found by the more elaborate approach. Note that the approximation effectively replaces \( K' \) of the original reduced-eye model with \((x - 1.82)\) mm.

### Image position as a function of the power of the ophthalmoscopy lens

It was shown in Figure 7 that the overall magnification of the fundus image varies with the position of the ophthalmoscopy lens \( (d) \) and the ametropia \( (K) \) of the observed eye. However, as noted earlier, although the magnified image may, in principle, be observable using a range of lens positions, there are advantages in placing the lens at a specific distance from the eye. This is primarily because if it is to act as a condensing lens and provide the optimal field-of-view and maximal image luminance, it must ideally image the exit pupil of the observed eye close to the entrance pupil of the biomicroscope (better known as pupil matching), and the illumination system's exit pupil onto the eye's pupil. In our reduced-eye model the entrance pupil, \( E \), is situated at the cornea (Figure 1). The requirement that the ophthalmoscopy lens has a condensing role implies that the object distance \( d \) (i.e., the distance of the eye's pupil from the ophthalmoscopy lens) is chosen such that the eye pupil's image lies at a distance from the lens equal to the sum of the distance of the fundus image from the lens and the working distance of the biomicroscope. Although the latter varies somewhat between different makes of slit-lamp biomicroscope, a working distance value of 100 mm is often assumed.

Evidently, the requirement that the image distance for the eye’s pupil satisfies these conditions implies an optimal value of \( d \), the eye pupil-to-ophthalmoscopy lens distance in our model. When a real eye and a ‘thick’ ophthalmoscopy lens are used, the corresponding recommended working

| Nominal power of Volk lens (D) | +60   | +78   | +90   |
|-------------------------------|------|------|------|
| Equivalent power of Volk lens (D) | +52.2 | +64.5 | +78.9 |
| Manufacturer’s claimed nominal magnification for a presumed emmetropic eye with \( F_E = +60.00 \text{ D} \) | \(-1.15x\) | \(-0.93x\) | \(-0.76x\) |
| Magnification for \( F_E = +55.00 \text{ D} \) | \(-1.05x\) (91%) | \(-0.85x\) (91%) | \(-0.70x\) (92%) |
| Magnification for \( F_E = +70.00 \text{ D} \) | \(-1.34x\) (117%) | \(-1.09x\) (117%) | \(-0.89x\) (117%) |
distances can be derived by subtracting, from \( d \), the distances of the entrance pupil of the human eye to the anterior corneal surface (about 3.1 mm) and the distance of the first principal point (P) from the ophthalmoscopy lens’ front surface (as presented in Table 2 for three ‘classic’ Volk lenses).

To establish the optimal value of \( d \) we must first consider the position of the fundus image as a function of \( d \). As an example, Figure 8 shows this dependence for a nominal +78 D powered Volk lens (actual equivalent power +64.5 D) with three values of ocular ametropia (\( K = -5.00 \) D, 0.00 D, +5.00 D). The image distances have been derived using Equation 7. For the emmetropic eye, the image distance is 15.5 mm for this particular lens, or, more generally, \( f_0 \). This implies that, for pupil matching, an acceptable approximation to the desired total image distance for the patient’s pupil, from the lens, is \((f_0 + 100)\) mm. Using the \( f_0 \) values from Table 1, this gives image distances of 119.2, 115.5 and 112.7 mm for the nominally-powered +60 D, +78 D and +90 D Volk lenses, respectively.

Using these ‘desirable’ image distances and the paraxial ‘thin’ lens equation, \( 1/l' - 1/l = 1/f \), the corresponding ‘desirable’ values of the object (the eye’s pupil) distance, \( d \), can be calculated. This gives values of \( d \) = 22.8 mm, 17.9 mm and 14.3 mm for the nominally-powered +60 D, +78 D and +90 D Volk lenses, respectively. To convert these values into working distances with real, ‘thick’ ophthalmoscopy lenses we must subtract the distance of the first principal points from the anterior surface of the lenses (from Table 2), and the distance of the entrance pupil of the eye located behind the anterior corneal surface (assumed to be 3.1 mm). This yields values of 14.7 mm, 9.4 mm and 8.1 mm, each only a little longer (by approximately 1 to 1.5 mm) than the manufacturer’s recommended working distance values of 13 mm, 8 mm and 7 mm for the +60 D, +78 D and +90 D lenses, respectively. These minor discrepancies may, in part, be due to the fact that the aspheric design of Volk lenses is such that the outer zones of the lens produce an image of the pupil at shorter distances than the paraxial zone.\(^{22,23}\) Thus, the manufacturer’s suggested working distances may yield a pupil image which is a better compromise between the different zonal foci rather than a paraxial focus as assumed by our model. It may also be that different assumptions are made regarding the relative position of the microscope’s objectives.

**DISCUSSION**

It appears that, provided the equivalent power of the ophthalmoscopy lens is used, our simple paraxial model and associated equations give satisfactory predictions of the changes in magnification with a model eye of known equivalent power \( F_p \) and known levels of axial ametropia. If, however, successful predictions are to be made with human eyes, it is essential that the equivalent power, as well as the ametropia of the eye, are known so that they can be properly considered when calculating the expected magnification. The equivalent ocular power can be approximated from a measurement of its axial length (see Equation 11). The requirement for a proper allowance to be made for the eye’s role as part of the image-forming system, particularly in terms of refractive error (\( K \)), will also apply to other clinical optical instruments used for observing the fundus too, such as retinal cameras and optical coherence tomographs.\(^{30-34}\)

In general, it is clear that although manufacturers may state that their ophthalmoscopy lenses have a specific magnification, these values apply only under defined, but usually unpublicised, conditions. It seems likely that these involve the use of a specific emmetropic eye model and lens position with, possibly, accurate ray-tracing through the thick lens systems of the eye and ophthalmoscopy lens without the approximations of paraxial optics. We assume the working distances that manufacturers quote for their lenses are those required for pupil matching, and that the aberrations of the thick lenses are optimised to minimise the aberrations of the fundus image when working with the chosen eye model. It would be helpful if manufacturers always specified the equivalent powers of their lenses. Although we have concentrated on higher-powered positive lenses of the type used for fundus viewing with the biomicroscope of a slit-lamp, the same basic formulae are applicable to negatively-powered lenses, such as the Hruby type, and to the lower-powered lenses used in conventional head-set-based indirect ophthalmoscopy.

**CONCLUSION**

The transverse magnification produced by any high-power ophthalmoscopy lens during indirect ophthalmoscopy, performed in conjunction with a slit-lamp biomicroscope for observing the fundus, depends upon a variety of parameters, including the equivalent power and position of the lens, together with the ametropia and equivalent power of the patient’s eye. Uncritical use of the lens manufacturer’s values for magnification may lead to errors of up to approximately 20% in the estimated dimensions of fundus features. In principle, these errors can be reduced by taking additional measurements to estimate the power of the patient’s eye, and then making use of the equations provided within this paper.

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CONFLICT OF INTEREST

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REFERENCES
1. Pearce JM. The ophthalmoscope: Helmholtz’s Augenspiegel. Eur Neurol. 2009;61:244–9.
2. von Helmholtz HLF. [Beschreibung eines Augen-Spiegels zur Untersuchung der Netz haut im lebenden Auge], Berlin, Förster’sche Verlagsbuchhandlung. 1851. Original Monograph: Description of an ophthalmoscope for examining the retina of the living eye. See Hollenhorst, RW. (1951) for translation. Arch Ophthalmol. 1851;1:46:566–83.
3. Bishai KR. An inexpensive method of indirect ophthalmoscopy. Br J Ophthalmol. 1989;73:235–6.
4. Singh A, Cheyna K, Wilson G, Sime MJ, Hong SC. On the use of a new monocular-indirect ophthalmoscope for retinal photography in a primary care setting. N Z Med J. 2020;133:31–8.
5. Henson DB. Optometric Instrumentation. 2nd ed. London: Butterworth Heinemann; 1998. p. 301–29.
6. Hruby K. Slit lamp microscopy of the posterior section of the eye with the new preset lens. Arch Ophthalmol. 1990;43:330–6.
7. Koepppe L. Die Mikroskopie des lebenden Augen hintergrundes mit starken Vergrößerungen im fokalen Lichte der Gullstrandschen Nernstslitlampade. I. Mitteilung. Die Theorie, Apparatur und Anwendungstechnik der Slitlampenuntersuchung des Augen hintergrundes im fokalen Licht]. Graefes Arch Ophthalmol. 1918;95:282–306.
8. El-Bayadi G. New method of slit-lamp micro-ophthalmoscopy. Br J Ophthalmol. 1953;37:625–8.
9. Sharma G, Purkayastha S, Deka H, Bhattacharjee H. Commonly used diagnostic and largest lenses for retinal diseases – an overview. DOS Times. 2008;13:49–54.
10. Sneed MP, Rubinstein MP, Jacobs PM. The optics of fundus examination. Surv Ophthalmol. 1992;36:439–45.
11. Golenbrander A. Chapter 63: Principles of ophthalmoscopy. In: Jaeger EA, Tasman W, editors. Duane’s ophthalmology on CD-ROM. Vol 1. Philadelphia, PA: Lippincott & Williams; 2006.
12. Ruben S. Estimation of optic disc size using indirect biomicroscopy. Br J Ophthalmol. 1994;78:363–4.
13. Volk D. Aspheric ophthalmic lenses. Int Ophthalmol Clin. 1965:5:471–94.
14. Ansari-Shahrezaei S, Maar N, Bivovski R, Stur M. Biomicroscopic measurement of the optic disc with a high-power positive lens. Invest Ophthalmol Vis Sci. 2001;42:153–7.
15. Ansari-Shahrezaei S, Stur M. Magnification characteristic of a +90-diopter double-aspheric fundus examination lens. Invest Ophthalmol Vis Sci. 2002;43:1817–9.
16. Wells CG, Barrall JL, Martin DC. Fundus measurements with indirect ophthalmoscopy. Arch Ophthalmol. 1992;110:1303–8.
17. Mainster MA, Crossman JL, Erickson PJ, Heacock GL. Retinal laser lenses: magnification, spot size, and field of view. Br J Ophthalmol. 1990;74:177–9.
18. Atchison DA, Smith G. Optics of the human eye. London: Butterworth Heinemann; 2000. Chapter 5, pp. 39–47.
19. Rabbetts RB. Bennett & Rabbetts’ clinical visual optics. 3rd ed. London: Butterworth Heinemann; 1998. p. 301–29.
20. Raasch T. Funduscopic systems: a comparison of magnification. Am J Ophthalmol Physiol Opt. 1982;59:595–601.
21. Raasch T. Funduscopic systems: magnification in ametropia and aphasis. Am J Ophthalmol Physiol Opt. 1985;62:19–24.
22. Volk D. Indirect ophthalmoscopy lens for use with slit lamp biomicroscope (US Patent Number:4627794). United States Patent Office; 1986.
23. Volk D. Indirect ophthalmoscopy lens for use with slit lamp biomicroscope (International Patent No:WO 94/10899). World Intellectual Property Organization; 1994.
24. Rassow B. A model of Gullstrand’s “normal eye”. Ophthalmologica. 1972;164:143–8.
25. Rudnicka AR, Edgar DF, Bennett AG. Construction of a model eye and its applications. Ophthalmic Physiol Opt. 1992;12:485–90.
26. Volk D. Slit lamp lenses instructions for use (Non-contact slit lamp lenses [PDF] - in English). (2021 [updated 2021]. Available from: https://www.volk.com/pages/slit-lamp-lenses-instructions-for-use. Accessed 11 Dec 2021.
27. Benjamin B, Davey JB, Sheridan M, Sorsby A, Tanner JM. Emmetropia and its aberrations; a study in the correlation of the optical components of the eye. Spec Rep Ser Med Res Counc. 1957;11:1–69.
28. Kim HS, Yu DS, Cho HG, Moon BY, Kim SY. Comparison of predicted and measured axial length for ophthalmic lens design. PLoS One. 2019;14:e0210387. https://doi.org/10.1371/journal.pone.0210387
29. Bennett AG. A method of determining the equivalent powers of the eye and its crystalline lens without resort to phakometry. Ophthalmic Physiol Opt. 1988;8:53–9.
30. Bennett AG, Rudnicka AR, Edgar DF. Improvements on Littmann’s method of determining the size of retinal features by fundus photography. Graefes Arch Clin Exp Ophthalmol. 1994;232:361–7.
31. Garway-Heath DF, Rudnicka AR, Lowe T, Foster PJ, Fitzke FW, Hitchings RA. Measurement of optic disc size: equivalence of methods to correct for ocular magnification. Br J Ophthalmol. 1998;82:643–9.
32. Hirasawa K, Shoji N, Yoshii Y, Haraguchi S. Comparison of Kang’s and Littmann’s methods of correction for ocular magnification in circumpapillary retinal nerve fiber layer measurement. Invest Ophthalmol Vis Sci. 2014;55:8353–8.
33. Iwase A, Sekine A, Suehiro J, Tanaka K, Kawasaki Y, Kawasaki R, et al. A new method of magnification correction for accurately measuring retinal vessel calibers from fundus photographs. Invest Ophthalmol Vis Sci. 2017;58:1858–64.
34. Littmann H. Determination of the true size of an object on the fundus of the living eye. By H. Littmann from the original article, “Zur Bestimmung der wahren Grosse eines Objektes auf dem Hintergrund des lebenden Auges,” which originally appeared in Klinisches Monatsblatter fur Augenheilkunde 1982; 180:286–9. Translated by TD Williams. Optom Vis Sci. 1992;69:717–20.