"Optics for Clinicians" by Melvin L. Rubin, MD, MS 25th Anniversary Edition 1993

the eye. This tilt is a compromise position to allow the visual axis to strike the lens perpendicularly during most visual tasks.

The normal eye constantly varies the position of its visual axis, typically between straight-ahead-gaze and downgaze for reading; (upgaze is quite unusual). If the visual axis is to strike the corrective lens as perpendicularly as possible in the *reading* position (in an effort to minimize the aberration just discussed), the tilt of that corrective lens downward, away from strict verticality, would have to be about 15°. However, since those same lenses also have to serve for *distance* seeing, a tilt compromise is usually made at about 7½°. This is called "pantoscopic" tilt and is built into most spectacle frames so that the lenses are closer to the face at their bottom edges, about 7½° off verticality; but too much such pantoscopic tilt will engender significant radial astigmatism, especially with highly ametropic corrections as already detailed.

"Corrected Curve" Lenses

Currently available spectacle lenses are designed to reduce as much as practical the radial astigmatism they generate. By properly choosing the front and back surface curvature for each distinct power desired (or, more practically, for each *range* of powers), this astigmatism can be effectively controlled. The "corrected curve" lenses manufactured by competing industrial giants do attempt to approach the proper shape. Punktal®, Orthogon®, Tillyer® — all are practical *compromises* to theoretical ideals to control radial (marginal) astigmatism and curvature of the image field.

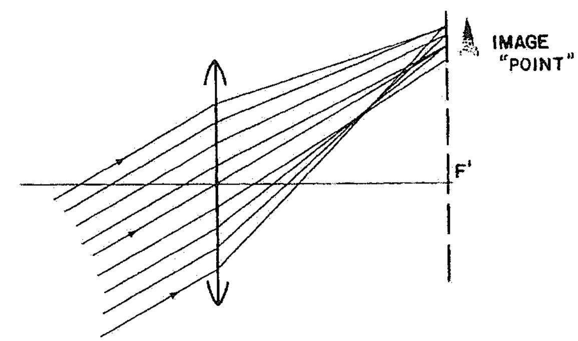
Coma

For us, this is the least important aberration. However, it also enters into retinal imagery considerations with off-axis objects and will tend to destroy the sharpness of details. It has a completely different optical basis than does radial astigmatism.

Coma is probably most easily defined as spherical aberration which occurs for off-axis object points. When you consider *axial* image "points" (or circles, in the presence of spherical aberration),

the distribution of light in those blur circles is symmetrical and the "blur circles" are round. For off-axis single object points, on the other hand, coma creates an *asymmetrical image* (as shown), that spreads out to look like a comet, even though it still is supposed to represent a *single* object point.

COMA



Coma is due to unequal magnification by different zones of the lens and can be reduced or eliminated in critical applications, such as fine camera and microscope lenses, by proper lens design and appropriate positioning of the aperture "stops".

PRACTICAL INSTRUMENTS

We are well nigh bringing this opus to a close. Before doing so I feel compelled to elaborate on the optical principles behind the use of some instruments so useful that I cannot conceive of any study of *practical* optics leaving them out. They are the *indirect ophthalmoscope* (personal bias causes me to mention this first), the *retinoscope,* the *lensometer,* and the *keratometer*. Sure, I know I've neglected others. (Some have been mentioned earlier.) These four, however, occupy so important a role in your daily clinical lives that your intellectual curiosity should prompt you to discover how they work. Any others I must leave to you.

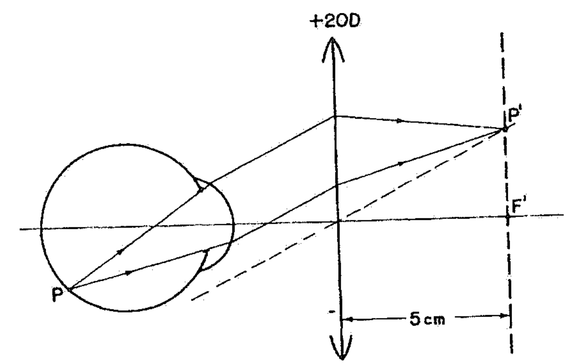
The Indirect Ophthalmoscope*

You are familiar, I am sure, with the principles of a slide projector. An intense source of light illuminates a transparent slide from behind and makes that slide a luminous object on its own. That luminous object emits light which is picked up by the projector lens. This lens is situated so that the object is just beyond its primary focal plane; then the final image can be projected onto a screen. The key here is the *luminous* object.

An "opaque" projector — the kind you use to project images of sheets of paper or pages of a book onto a screen — also works by making the opaque object material "luminous", in this case by shining an intense light onto it; the object *reflects* that light energy instead of transmitting it, as does the slide. Because the illumination system here is not quite as efficient as that for the slide projector, and also because the material to be projected is usually bigger, the illumination chamber of this projector is large and the projection lens diameter is quite wide to allow it to gather as much of the reflected light as possible so that a bright image can be placed on the screen.

Now what has all this to do with the indirect ophthalmoscope? Everything. For an observer to see the fundus in *any* kind of ophthalmoscopy, it must be illuminated. With the direct ophthalmoscope, you are simply using "flashlight" illumination to examine the innards of a dark cave. In *indirect* ophthalmoscopy you also must make the fundus a luminous object. Here, however, the eye acts like a miniature opaque projector and projects that fundus image out into space. This is accomplished by shining an intense source of light through the pupil (the only way to get the illumination in); the fundus image is then projected out.

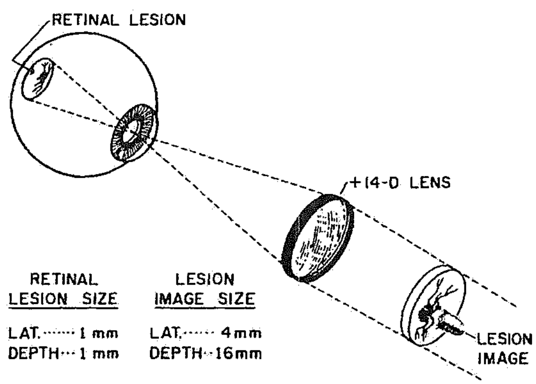
An emmetropic eye with an illuminated fundus would project the fundus image to infinity — the position conjugate to the retina. However, this distance is too impractical. (For one reason, by the time the image reached infinity, there would be so little light energy there that it would be too feeble to be seen.) If instead, we capture some of those parallel light bundles stemming from the eye with a hand-held condensing lens (say, one of + 20 D), a new luminous image will be created at *its* secondary focal plane. (Notice that this will *not* depend on how far the lens is from the eye!) While there is no screen stationed at the F' plane to intercept the rays, a real fundus image *is* there nonetheless; it is called an *aerial* image, and is, of course, inverted when compared to the orientation of the fundus itself.



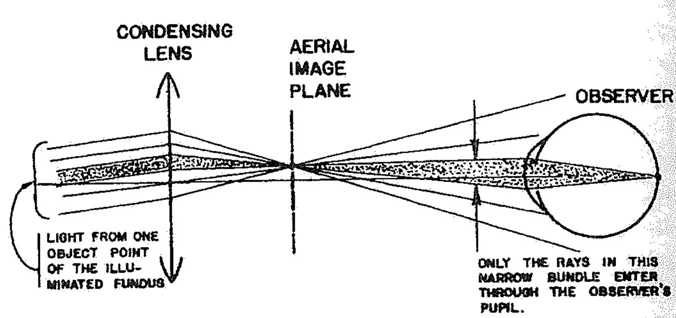
That aerial image has "thickness" representing any "thickness" present in the fundus itself; any excavation of the nerve head or elevation of a choroidal lesion will be manifested in the "depth" of the image. The *longer* the focal length of the condensing lens, the farther from the lens the image will be and, therefore, the more its *lateral* magnification. You should recall that the *axial* magnification of the image (that is, the "thickness" of the image compared to the "thickness" of the object) varies as the *square* of the lateral magnification afforded by the condensing lens. So, the axial image dimensions are *especially* thickened by the *lower* powered (longer focal length) condensing lenses, which provide a very "depthy" aerial image, and it is partially on the basis of this image depth that the observer's stereoscopic view of the fundus is determined.

* Condensed from, Rubin, M. L.: "Optics of Indirect Ophthalmoscopy" Survey of Ophthalmology, 9:449-464, October, 1964.

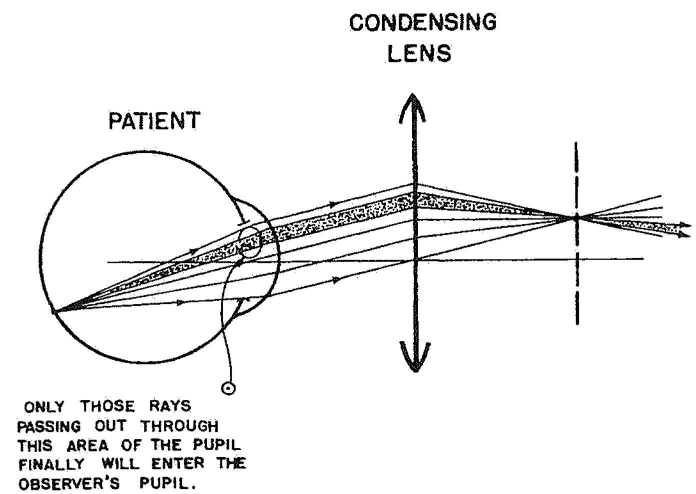
The diagram below shows the retina imaged by a + 14 D lens and the magnifications attained:



Fine. An image is there in space. But if someone is to see that image, light from it must reach the observer's eye and get through his pupil. But, only a small quantity of the light in that real aerial image can enter the observer's eye. (That tiny portion is shown in the shaded area of the next two diagrams.)



That light which does enter the observer's eye (if it is traced backwards), must originally have come through a comparably small area of the *patient's* pupil as shown below:



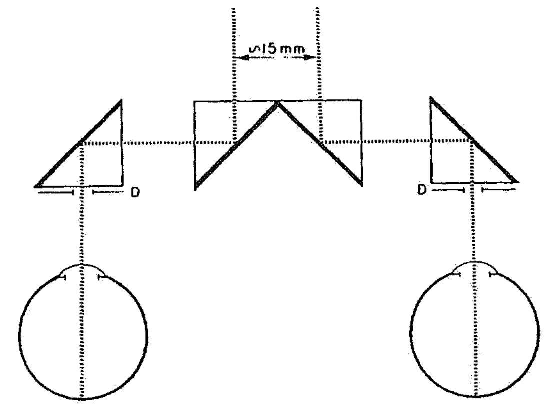
The further back from the lens the *observer's* eye is positioned, the smaller will be that area of the patient's pupil required to allow this visualization to occur. (So, if the observer moves backwards, he can more readily visualize a fundus through a small pupil.) It should take no great insight to see that the observer's pupil and that tiny area of the patient's pupil are conjugate; that is, the spot in the patient's pupil *is* the image of the observer's pupil by the condensing lens. (I call this object-image relationship a "conjugacy of pupils".)

So, although the condensing lens *could* be positioned *anywhere* in front of the patient's eye if its *only* job was to form an aerial image of the fundus (the image would always be at its F' plane if the patient's eye were emmetropic), the fact that that same lens must also image the observer's pupil into the patient's pupillary area *fixes* the position of that lens. If the observing eye approaches, in order to maintain this "conjugacy of pupils" the hand-held lens must be pulled toward the observer somewhat. An added benefit of moving closer is that the observer gets closer to the aerial image and obtains the reward of a larger image of it on his own retina.

As if that hand-held lens didn't have enough to do; don't forget we must somehow illuminate the patient's fundus. That same lens must also project *into* the patient's eye the image of the illumination source, *and* it must do so through an area of the patient's pupil which is not "occupied". As just shown, that image light which comes out of the pupil and heads for the observer's pupil has its own reserved area. This area must not be interfered with by the incoming illuminating beam. (This is another of Gullstrand's innovations — that is, he stressed the necessity of keeping the incoming and outgoing beams separated to keep glare at a minimum for the observer.)

So far, well and good. We have now shown the principles of monocular indirect ophthalmoscopy, where the observer uses only one eye. But one of the great advantages of examining a fundus by this indirect method is that it affords us the means of having *stereoscopic* visualization. For this to occur, *both* the observer's pupils must receive light from the "thick" aerial image, and the only way this can occur is if *both* his pupils are imaged into the patient's pupil. For all this to happen within the plane of the patient's pupil, it should be obvious that that pupil must be as dilated as possible.

However, we are presented with an optical problem. The observer's interpupillary distance is roughly 65 mm. This separation must be reduced if we are to squeeze the images of *both* observer's pupils into a patient's pupil which is only 6 or 7 mm in diameter. If the *observer* were far enough away (say 100 cm or more), the image of *both* his pupils *could* project into the patient's, but the observer's arms are not usually that long! So, we turn to another option and that is to optically reduce the observer's interpupillary distance with mirrors (actually, totally internal-reflecting *prisms*). These are placed in such a way so as to narrow the interpupillary separation from 65 to about 15 mm.

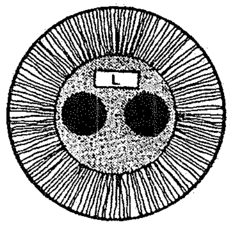


The beauty of this device (pictured above) is that the outer mirrors are made movable and can be adjusted to suit any individual observer's p.d. The actual, "optical" separation of the pupils, however, is *fixed* by the non-movable inner mirrors at about ¼ the *normal's* p.d.

There is also about 2 D of plus power built into the eyepieces to relieve any strain on the observer's accommodation (and concomitantly, on his accommodative-convergence) as he attempts to view the aerial image.

(Incidentally, the narrowed instrument p.d. reduces the image disparity available for the two eyes for *all* objects: So, as you look about through the eyepieces, space seems "flattened" — your stereopsis for everything is reduced; you may fumble and find it very difficult to pick up the traction sutures around the extraocular muscles during a retinal operation.)

So with the narrowed p.d., both pupils *can* be squeezed optically into the patient's pupil. The illuminating light can enter through a separate pupillary area.



Voila! — a full, stereoscopic "Technicolor" view of the patient's fundus pops into view and seems to "fill" the condensing lens.

The extent (field of view) of the patient's fundus which is visible by this method is great — roughly 25° versus only 8° with the direct ophthalmoscope — almost 10 times as large an area becomes visible at one time. For a given power of condensing lens, the extent of field visible is directly proportional to the *diameter* of the lens. A 20 D lens of *large* diameter affords a greater field of view than one of smaller diameter. (Here, I'd like to "plug" the new Nikon 20 D aspheric lens which affords a beautiful, crisp aerial image with the same magnification (linear and axial) as that provided by a standard 20 D lens; but in addition, the diameter of that lens is large enough to provide the same wide field of view as that given by a standard *30 D* lens!)

Light Reflexes

As we have pointed out, ideally, if a reflex-free field of view is to be obtained by an examiner, the illuminating and observation systems must be kept completely separated as the light rays course in and out — through the observing lens, the patient's cornea, pupil, and crystalline lens. If the image of the illuminating system is projected by the condensing lens into the patient's pupil through an area *away* from the position where the images of the observer's two pupils are imaged (last figure), the entering and outgoing light paths *are* kept separate and no corneal or crystalline lenticular reflexes would be visible. However, things are not always ideal. The light source

utilized in the modern indirect ophthalmoscope headpiece is too broad to be focused easily in this manner, though it *is* possible — especially with the new MIRA* "small pupil" ophthalmoscope. The latter is a very flexible instrument. It is designed with enough adjustments to allow not only compensation for the observer's p.d., but also for the diameter of the *patient's* pupil; that is, the "instrument" p.d. can also be varied as can the position and size of the light source. All this allows examination of a fundus through a very small pupil with even some stereopsis possible. Furthermore, the "instrument" p.d. can be *widened* to take advantage of a fully dilated pupil to provide a greater stereoscopic effect than can the average indirect ophthalmoscope.

But, in the average clinical situation with a standard indirect ophthalmoscope, a completely reflex-free condition is unlikely. Careful observation by an examiner will reveal that there are three bright and annoying reflexes seen when he tries to visualize the fundus through the hand-held condensing lens; these are most troublesome to the beginning ophthalmoscopist. These three are the light reflections which occur from the anterior and posterior surfaces of the *condensing lens* plus the one from the anterior corneal surface. Since the condenser is a biconvex lens, the anterior convex surface forms a *virtual* reflected image of the illuminating headlamp, and the posterior surface, being concave to the illuminating source, forms a *real* image. These images therefore will move in opposite directions if the lens is tilted. The higher the dioptric power of the condensing lens, the more will be the tilting necessary to move the reflexes away from the center of the field of view. The small, bright *corneal* reflex moves only slightly with the lateral motion of the condensing lens, but it can be moved sufficiently so that the specific area being examined is not obscured. All this is usually discovered empirically by the clinician, who learns quickly to maneuver about even though these reflexes are present.

* Medical Instrument Research Associates, Boston

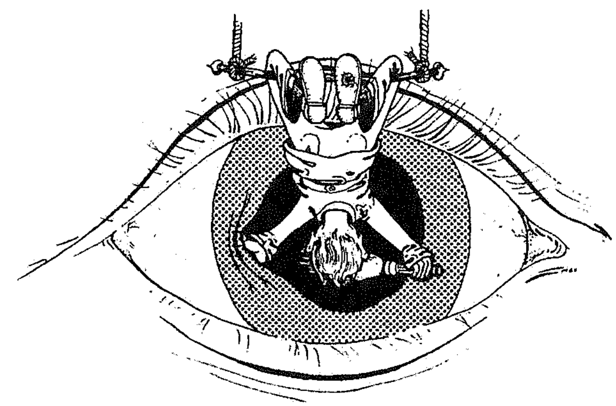
Image "Brilliancy"

A widely dilated pupil facilitates the *admission* of light in any type of ophthalmoscopy. However, as you have seen, not all this light is used in the formation of the visible image. When one views an illuminated fundus with a direct ophthalmoscope, the diameter of the *emergent* beam through the dilated pupil is "stopped down" by the narrow aperture in the ophthalmoscope head (about 3 mm), but this is close to the patient's pupil, so a relatively high proportion of emergent light enters into the image, though, of course, *some* light is "wasted". In indirect ophthalmoscopy much more of the light is lost, since the only rays emerging from the patient's pupil which are *really* effective in the formation of the visible image are those which come through the tiny pupillary areas which contain the images of the observer's pupils. Thus, as long as the patient's pupil is sufficiently dilated to allow the unhindered entry of the illumination beam and the free emergence of the image-forming beams, having the pupil still more dilated can *not* further increase the brightness of the aerial image (except when you are looking at the fundus periphery, where the dilated pupil reduces vignetting). What the dilated pupil *does* accomplish, however, is it allows more leeway in the positioning of the observer's eyes, the viewing lens, and the illumination source, thus simplifying the entire viewing procedure.

So, if it's not the dilated pupil, what is it that makes the image seen in indirect ophthalmoscopy so bright? You should give most of the credit for the bright image and the good visibility to both the truly intense light source available in the modern instruments plus the large field of view which provides a "bird's eye view" of a large fundus area. Thus, an indirect ophthalmoscope seems to "cut through" the opacities in the optical media of the eye examined and affords the opportunity to see a fundus which may be hidden from the view of an observer who attempts to use a direct ophthalmoscope.

This is the story of indirect ophthalmoscopy and is complete enough for this time and place. You should now understand the basic

operations involved and should picture to yourself the following cartoon summary of the optics of your trusty instrument:



The Retinoscope

Principles

The retinoscope provides us with an opportunity to determine the refractive error of an eye objectively. For an anesthetized infant, for example, from whom we get no subjective response to "Which is better, one or two?", the retinoscope affords us the only practical way to measure the refractive state.

The retinoscope is a small projector which emits a *spot* or *streak*-like image of the lamp filament itself. (The "streak" scope is now the most popular design, and deservedly so since it makes the determination of astigmatism so much simpler). This "projector" is aimed at a patient's eye.

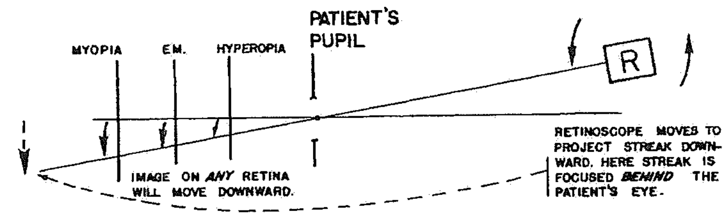
Most commercially available scopes provide for a variable focus of the filament image. The retinoscopist can project that image onto a wall any distance away and focus it sharply, from infinity to any place behind or in front of the patient's eye; that is, he can vary the vergence of the focused image. In fact, the image can even be made virtual with *divergent* light; that is, it can optically be positioned so that it seems to arise from *behind* the observer.

If the projected light emitted from the retinoscope is made *either* divergent or parallel or convergent to a point somewhere *behind* the patient, the optical effect as far as the patient's eye is concerned is the same; and this is called the "plano-mirror" effect. If, on the other hand, the projected filament image is located somewhere between the observer's eye and the patient's eye (the only space left!), the retinoscope is said to have a "concave-mirror" effect.

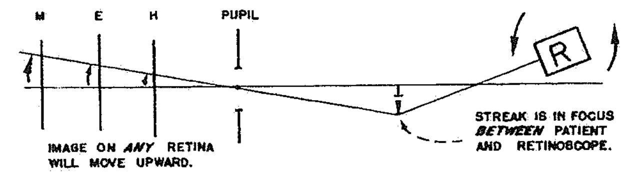
This flexible focusing arrangement allows a swift change from "plano-mirror" to "concave-mirror" effect, but should only be considered an instrumental refinement; it is *not* absolutely necessary to the performance of clinical retinoscopy. Actually, the retinoscope's focus may be *pre-set* to yield either a "plano-mirror" *or* "concave-mirror" effect. In fact, even an ophthalmoscope can be used to function as a "plano-mirror" retinoscope!

The reason that actual focusing of the streak is not a critical requirement of retinoscopy is that the patient's eye is not meant to accommodate on that streak image anyway; the blurry image of the light streak which does fall on the patient's retina simply becomes a new, tiny *source* which can be thought of as "originating" there. The motion of that source on the retina is directly dependent on the retinoscope's movement. When the retinoscope is set for a "plano-mirror" effect, the direction of motion of the light source on the *retina* will always (*regardless* of the type or amount of refractive error of the patient's eye) be in the *same* direction as the motion of the retinoscope light as it sweeps across the patient's face; that is, any movement of retinoscope's image will impart a "like" motion to the retinal image (see figure A below). In contrast, the retinoscope with a "concave mirror" effect (which images the filament *between* the observer and the patient) will create a motion of the "new source" on the *patient's* retina which is always in the opposite direction to the motion of the retinoscopic streak. (See figure B below).

A. PLANO-MIRROR EFFECT



B. CONCAVE MIRROR EFFECT

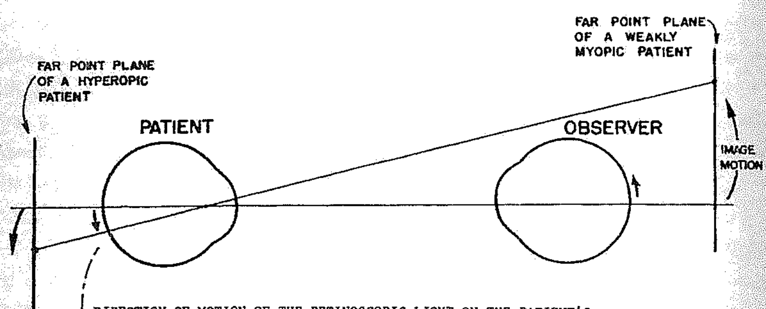


The relative motions we have just discussed are between the image of the retinoscope light and the patient's retinal image.

After the (streak) image is projected into the patient's eye, that eye will treat the bright but blurry image as a new source which now originates at the patient's retina. The dioptric power of his eye projects it back out toward the observer. With no spectacle lens present in front of the patient's eye, the projected image from the patient's retina *must* lie at the far point plane of that eye, since the retina and the far point plane are conjugate.

The observer watches this light through the peephole ("stop") of his retinoscope. The light he sees coming through the patient's pupil is actually coming from (or aiming towards) the far point of that eye. But it still *seems* to be located within the patient's pupil.

Depending on where the patient's far point plane is relative to the observer, the motion of that image projected from the patient's eye will either appear to the observer to move in the same ("with") direction or in the opposite ("against") direction as the motion of his retinoscope light.

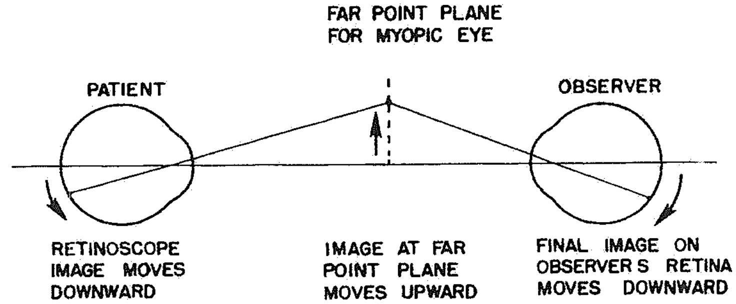


DIRECTION OF MOTION OF THE RETINOSCOPIC LIGHT ON THE <u>PATIENT'S</u> RETINA. THIS MOTION IS PROJECTED TO THE PATIENT'S FAR POINT PLANE -- A HYPEROPIC AND A WEAKLY MYOPIC ONE ARE SHOWN. THE MOTION "SEEN" BY THE OBSERVER'S <u>RETINA</u> IS INDICATED.

Study the diagram above:

If the patient's far point plane is located anywhere *except* between himself and the observer (where it *would* be in moderate to severe myopia), a downward motion of the image on the *patient's* retina will always produce a light motion in the *upward* direction on the *observer's* retina; and, naturally, any image movement which moves upward on the observer's retina is interpreted by his own consciousness as *downward* movement. Thus, with the retinoscope streak moving downward, he would perceive a "with" motion — a downward image movement coupled with a downward retinoscope movement.

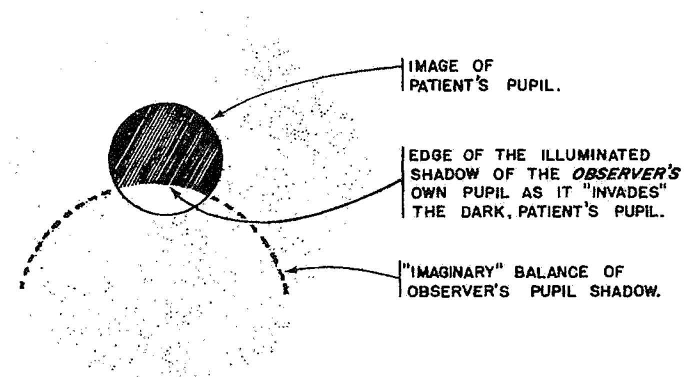
But, if the far point plane (and, therefore, the specific location of the projected image from the patient's retina) is *between* the patient and the observer, then, as the light on the patient's retina moves down, the image *projection* (in the far point plane) moves upward (as shown below).



Since here, this image is in *front* of the observer's eye, its upward movement will cause a downward movement of the final image on the *observer's* retina. His brain will interpret this as *upward* and, therefore, "against" the downward motion of the retinoscope light.

The observer sees all this light-image motion as taking place within that tiny area on his own retina that contains his image of the patient's pupil. For you optical purists, look at a sketch of what *is* on the observer's retina:

ON THE <u>OBSERVER'S</u> RETINA



The image of the *un*illuminated patient's pupil is there as a dark spot. As the light emanating from the patient's far point plane is projected towards the observer's eye, it casts a shadow of the *observer's own pupillary margin* onto his own retina. The part of this latter shadow that corresponds to the *observer's* central pupillary *opening* (shown above) is obviously *illuminated* by the entering light and is not dark; however, the invading light is limited to that area shown because of the particular aperture "stops" of this optical system. What can be said is that the closer to the observer's eye the patient's luminous retinal image is located, the larger and brighter will be the illuminated shadow and the quicker it will move. In summary, the edge of the *light* reflex seems to invade the dark area which is the image of the patient's pupil; in reality, the *border* of this light represents the shadowy margin of the observer's own pupil (or, his *retinoscope's* aperture "stop", whichever is the smaller).

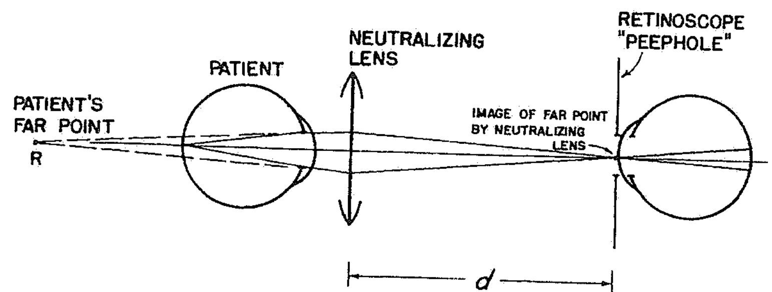
So, in retinoscopy, (actually "*skia*scopy", since one is viewing *shadow* movements), the observer's first task is to compare the motion of his retinoscope light across the patient's face to that of the light emanating from the patient's eye. The motion will be "with" or "against". His next step is to attempt to influence the motion he observes by placing appropriate lenses in front of the patient's eye — plus lenses are used to counter any observed "with" motion; and minus lenses, the "against". As these lenses are added, the observer continues to watch their effect on the image motion through the retinoscope "peep hole". His goal is to "neutralize" this observed motion with the lenses. At "neutrality", a small change in the added lens power will cause a prompt reversal of the direction of image motion.

Specific Optics

As stated, when starting retinoscopy with *no* lenses in front of the patient, the image of the retina is projected to the far point plane of that eye. As lenses are systematically added, that retinal image is, in turn imaged elsewhere, plus and minus lenses moving that image as you'd expect. At "neutrality" the projected image from the patient's retina is at the "peep hole" aperture of the retinoscope (or the observer's pupil). When the image is immediately in front

of it, a rapid "against" motion is seen; when just behind, a fleeting "with" motion is evident. But *at* "neutrality", the image of the patient's retina is imaged precisely at the retinoscope, *wherever* that instrument happens to be located. *This* is the key principle of retinoscopy. At the end-point (neutrality), the patient's retina is made conjugate to the retinoscope. Do not miss this forest for all the trees (the "trees" being the analyses of the various image motions), even though the specific trees and leaves are interesting.

"NEUTRALITY" END POINT



As shown above, then, at "neutrality" we will have found the neutralizing lens power necessary to image the retina at the retinoscope. But, that isn't enough; what we *really* want to know is the *"corrective"* lenses for this patient's eye, (that is, what the refractive *error* is). For this we must image the retina not at the retinoscope, but further back behind the observer, at *infinity*. This will always require some additional minus lens power, the exact amount of which would depend on how far away the retinoscope is from the patient's eye. If this latter distance is d (called "the working distance"), the lens always has to be of power $\frac{1}{d}$ to accomplish its task. If $d = 50$ cm, a -2 D lens must be added to the lenses already in place at "neutrality" to yield the full "corrective" lens for the patient's eye.

For an astigmatic eye, each major meridian must be neutralized separately, and this may be done with spheres or cylinders. Though either plus or minus cylinders *can* be used for retinoscopy, *plus* cylinder neutralization (with the use of a *"plano-mirror"* retinoscope effect) is much to be preferred since it makes the observer's task easier. (For all practical purposes, it is much simpler to recognize and neutralize "with" motion than "against", and *plus* cylinders force you to work with "with" motion.)

The fine points of practical "streak" retinoscopy are presented by Jack Copeland ("Mr. Retinoscopy" himself) in his "how-to-do-it" manual.*

Monocular Diplopia

This might seem to be an unusual spot for this subject, but you will soon see why it is placed here.

THE WOMAN WHO SAW TOO MUCH**

A 50-year-old secretary appeared in my office with the following story: a few months ago she had begun to see that the lines she was typing were definitely "double." This symptom gradually worsened, so that now everything she looked at was tainted; that is, all subjects appeared as if they had a "will-o-the-wisp" quality, with a faint, offset, secondary image superimposed. Moreover, this montage was seen only with her left eye! I smugly asked how she was so sure that her right eye contributed nothing to the diplopia and she embarrassed me with the quick retort, "I lost that eye in a childhood accident." I tried to recover my shattered composure with an inane, "That sure is a beautiful prosthesis; just where did you have it made?"

Using a *single* Snellen target line, I found that her left eye showed a best-corrected visual acuity of 20/20 with her present glasses. During this test, she mentioned spontaneously, "There are *two* lines visible; the lower one is clear but the upper one is blurred." Moreover, I was amazed to find that the blurred line *could* be sharpened to 20/20 too, with a $-3 +1.50 \times 43°$ correction over her previous prescription. (This additional lens, of course, blurred the previously clear line.) Diagnostician that I am, I fully anticipated finding some slit-lamp evidence of an early cataract, but the examination with a slit revealed no sign — not even a hint that I could detect — of any lenticular (or corneal) bubbles or opacities!

On fundus examination, as I peered through her own eye's optical system with my direct ophthalmoscope, I had another surprise; I too saw double — a blurred *and* a clear image of each single retinal vessel! At least the existence of a bona fide monocular diplopia was confirmed. By using yet another objective test for a quick confirmation of the optical error, I convinced myself that she indeed did have an early cataract.

Following my examination, confidently, I sat down with the patient and analyzed her clinical problem. Since she is still engaged in her vocation without encountering too much difficulty, she could continue to do so as long as she can, especially since she has but one eye. The latter finding should certainly flash the caution signal and lead a surgeon to take a very conservative approach. However, when the diplopia does become too great a burden for her daily routine, one can try a simple trick which occasionally temporizes; prescribe a trial of pinhole lenses. *(See last sentence, page 190.)* If this doesn't work (or her visual acuity drops to a distinctly poor level for her everyday needs), only then will I recommend sharpening the keratome and extirpating the culprit of her difficulties!

Before I let *you* in on the objective test that I used to detect that there was indeed a real optical error present, let's discuss monocular diplopia for a few moments.

"Double-vision" is a most peculiar symptom, especially when it occurs with only one eye doing the seeing! This is monocular diplopia or polyopia. Sometimes patients describe a complaint which initially seems like polyopia; on careful questioning, however, the symptom really will be found to be a *distortion*, where "straight" lines appear kinked or crooked. Such patients may have a preretinal membrane with retinal wrinkles ("pucker"), submacular fluid, or choroidal elevations. They do not exhibit true diplopia. Real monocular diplopia has a number of causes but most have a purely *optical* rather than a *neural* basis. (Occasionally following squint surgery, during a period of spontaneous conversion from anomalous to normal retinal correspondence, there will be a few days of *"neural"* monocular diplopia. However, generally, the diplopia arises from a problem within the eye's refractive components.)

The commonest cause of monocular diplopia is probably *physiological.* Any "sharp" individual may notice this, wonder about it, and call it to your attention, though it usually will not bother him greatly. This type is due to a malalignment of the cornea and the lens so that their optical axes do not coincide — the lens being tilted somewhat. Remember that the human lens is not the simple structure that we usually describe, but a truly complex one composed of a number of layered, curved surfaces separated by different indices of refraction; a tilting on its axis may allow the edge of one of the refractive junction zones within the lens to straddle the pupil and create multiple retinal images. Typically, only 2 recognizable images are formed — the bright, primary one and a very faint and "ghostlike" secondary. This latter image is usually about $4—6^{\Delta}$ offset superiorly from the main one. (Critical observation of a bright point or horizontal line target on a dark background will demonstrate this to you.)

Obviously, this same cause and symptom can be exaggerated if the lens is loose or partially dislocated, say, after trauma. In this instance, if the lens is also knocked slightly *backward* from its normal position, in addition to the possible diplopia, some *hyperopia* is induced too; If, instead, the lens falls somewhat *forward*, a myopic error will supervene. Furthermore, with either direction of lens shift, the more the lens is tilted, the greater the astigmatic error created (radial astigmatism).

Another extremely common cause of monocular diplopia is *early* cataract, (as in the case reported above). That diagnosis sounds as if it were a cinch to make, but apparently it is not since it is so often missed. Of course, when the lens changes are advanced, it is not difficult to associate the cause (cata-

* OPTEC Inc., 3543 N. Kenton Avenue, Chicago

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ract) and effect (diplopia). But very tiny cortical "bubbles" or a very slight increase in the lens' nuclear density can also cause a remarkable amount of diplopia and yet you might see nothing unusual during your slit-lamp examination. So, what *can* help you determine what's wrong? It is your trusty *retinoscopic reflex* which serves as a sleuthing beacon. It can provide the tell-tale evidence of an early cataract, or for that matter, *any* optical cause of the monocular diplopia. The reflex will show distortion much earlier and more predictably than will any other method of detection. So, use it wisely and learn to recognize small irregularities which even your "hotshot" colleagues might overlook.

Other possible etiologies for monocular diplopia are as follows: high astigmatic errors (with or without optical correction by lenses), keratoconus (even in very early stages), corneal scars, megalocornea, and spherophakia.

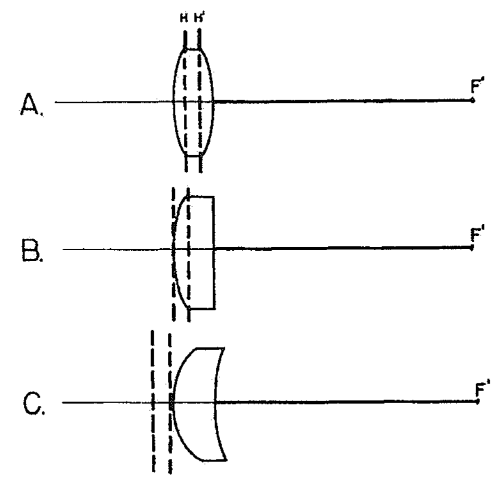
I trust you will now look less at the psyche and more critically at the retinoscopically created red reflex the very next time a patient complains of a superabundance of images! (*Article finis.*)

The Lensometer

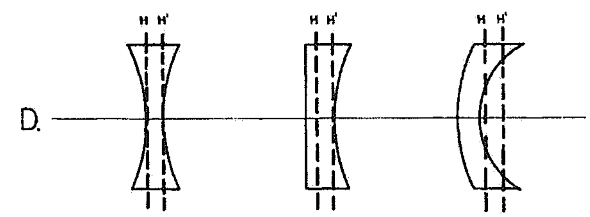
In clinical practice, somehow we have to be able to determine quickly the power, cylinder and axis of an "unknown" lens. I have already described the "hand neutralization" method and have alluded to a "computer" in the office which could take complex spherocylindrical lens combinations and produce a resultant without recourse to formulas or graphs — that "computer" is the lensometer. (This instrument of fantastic clinical utility can even measure the amount and direction of any prism which may be incorporated or induced by the lens).

The most clinically important information about any corrective lens is the distance between its back surface and its secondary focal point. This distance is the "back focal length" — its reciprocal is the *"vertex"* power (also called the "effective power").

The reason I stress the term *"back* focal length" is to contrast it with the "true" focal length, which is the distance between the secondary principal point and the secondary focal point. The position of the principal planes is determined by the "shape" of the lens (that is, the curvature of the front and back surfaces) — the more highly curved the surfaces (for *plus* lenses) the more anteriorly the planes are "shoved". (See next figure). In contrast, increasing the surface curves of a *minus* lens causes the principal planes to be pushed in the other direction — "out the back").



THE VERTEX POWERS ARE IDENTICAL; THE "TRUE" (EQUIVALENT) POWERS ARE *UNEQUAL*.



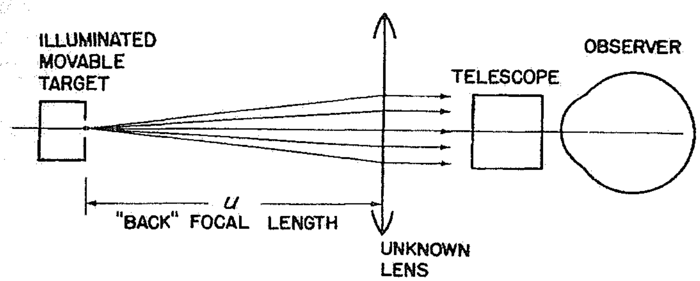
CHANGING THE SHAPE OF A ***MINUS*** LENS "SQUEEZES" THE PRINCIPAL PLANES THE OPPOSITE DIRECTION

Look at the three sample plus lenses above, A, B, and C. In each case it is the distance from H' to F' that is the *true* secondary focal length. This is longer in C than in A, and therefore, the *"true power"* would be greatest in A. Knowing the *true* power (also called the "equivalent" power) of a lens might be important in camera lens design, but for us, the important relationship is the *vertex* power. In the clinic, F' must be capable of being pinpointed quickly, and relative to the readily accessible lens surface — *not* to the intangible principal plane. This is the task of the lensometer.

In each of the first three lens examples above, the distance from the *rear surface* of the lens to F' is the same; thus, the "vertex" power of these lenses is identical — a lensometer would provide the same power reading for all.

Now, let's investigate how a lensometer *might* be constructed. The idea behind a simplified one is shown below:

A SIMPLE LENSOMETER

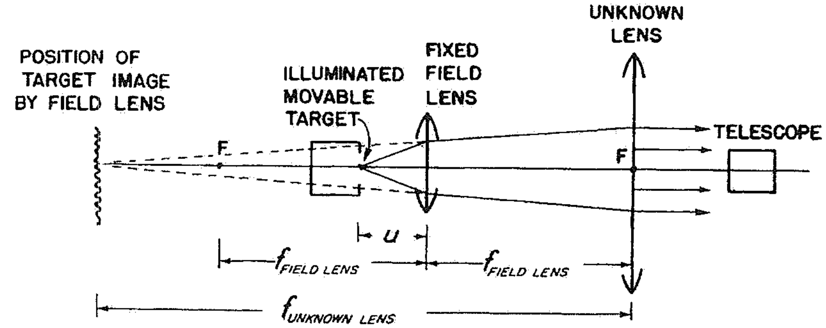


A grid-like, luminous target is moved back and forth behind an "unknown" lens until light rays leaving the lens are parallel (as detected by a small eyepiece telescope which is focused for infinity so that only parallel light bundles will appear to be clear). The target must then be located at F' of the lens. Its distance u is simply measured and converted to diopters.

Now, what's so tough about that? Nothing, really, but this method is highly impractical — the instrument would have to be too long. (Think of trying to measure a lens of 0.25 D power — you'd have to have an instrument which was 4 meters long!) Another problem with this lensometer is that the dioptric scale would not be *linear;* a given, say 1 cm, movement of the target might correspond to **1** D for a short focal length lens (say a **10** D lens), but corresponds to only .01 D for a 1 D lens. Thus, our calibration scale (in diopters) would have to expand tremendously at the lower end of the power scale.

Both of these objections are taken care of by adding another fixed lens (field lens) into the system. It is carefully positioned (see diagram below) to take advantage of a neat optical trick called Badal's principle.

A LENSOMETER



The fixed field lens is placed so that its focal point exactly coincides with the posterior vertex of the unknown lens. (The latter lens should be clamped in place to fix its position). The observer moves the illuminated target back and forth in relation to the fixed field lens until the reticle target appears sharply in focus through the eyepiece telescope. Then, parallel light bundles must be entering his eye.

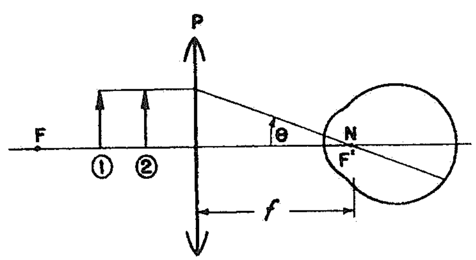
Let's examine the diagram above to see what happens when the lens is so neutralized. The target is imaged by the field lens. Since the target is closer to this lens that is its own focal point F, a *virtual* image will be created over at the left as shown. That image position must also lie in the focal plane of the "unknown" lens since we know that the *final* image rays entering the telescope are parallel. Thus, adding a field lens to our lensometer allows us to condense instrument space; the field lens optically moves the target to a given distance instead of forcing *us* to move it there physically (as was required in our original, simplified instrument).

But the field lens does more than this: with the focal point of the field lens situated exactly at the posterior vertex of the "unknown" lens, the proper optical condition is set up so that distance u (from the target to the field lens) is *directly* proportional to the power of the "unknown" lens. (This optical condition — Badal's principle — will be explained next). Because of this direct proportion, when we calibrate the lensometer scale to read directly in diopters, that scale will be perfectly linear — a given distance along it will always be equal to a fixed value of dioptric power, whether from 10 to 11 D or from 1 to 2 D — the scale will not shrink as higher powered lenses are tested, as it did in our first instrument example. The clinical lensometer is a beauty of simplistic design.

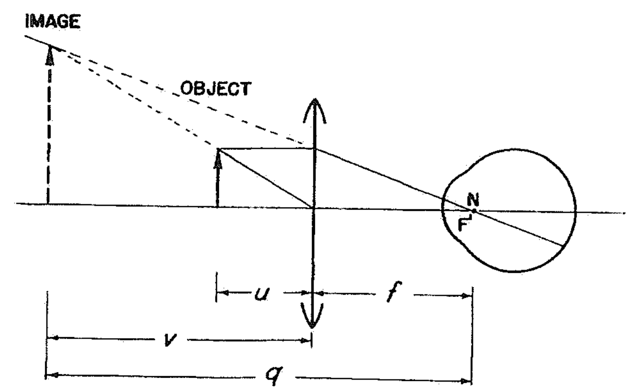
The Badal Principle

First, a little background — when you are checking a patient's near point of accommodation, you usually ask him to read some fine print on a test card. You gradually bring the card closer, approaching the eye until he admits to a blurring of the print. Though this is practical for the clinical situation, it would not be satisfactory for accurate measurements in the laboratory, since, as the card approaches the eye, the angle subtended by the print at the nodal point of the eye grows larger. You are, therefore, making it easier for a patient to see that same print and less likely for him to recognize its initial blur.

There *is* a way to vary the accommodation stimulus yet still keep the (angular) image size on the retina constant. The optical principle which allows this to happen was discovered by Badal.



Place any plus lens in front of the eye so that the lens' F' coincides with the eye's nodal point. An object is put in front of that lens, but closer than its primary focal point F. The object tip (say, an arrow with its end on the axis) will emit *one* ray which is parallel to the axis. After refraction, that ray will pass through F'. Since F' was made to coincide with the nodal point of the eye, the ray will continue on undeviated. The angle at the nodal point sets the angular size of the image seen by that eye as θ, no matter how far away that image actually is located. In the above diagram, object positions 1 and 2 both will create identical angular image sizes for the eye, although the image caused by the object at 2 will, naturally, be closer to the eye than the one formed by the object at 1. The image corresponding to the object at 2 will, therefore, stimulate accommodation to a greater extent. So you can see, with this optical system, the stimulation to accommodation varies while the angular image size remains the same.



Pertaining to lens P above, the object is at distance u and the image at distance v. Back again to $U + P = V$, but here, both U *and* V are divergent, and, therefore are negative.

$$-U + P = -V$$

$$-\frac{1}{u} + P = -\frac{1}{v}$$

When the two terms of the expression $(-\frac{1}{u} + P)$ are combined over one denominator u, we obtain

$$\frac{-1 + uP}{u};$$

So, $$-\frac{1}{v} = \frac{-1 + uP}{u}$$

Then, $$\frac{1}{v} = \frac{1 - uP}{u}$$

and $$v = \frac{u}{1 - uP}$$

Now, let's relate this information to the *eye.* The distance from the nodal point N to the image is $(v + f)$; let's rename this total distance q.

$$q = v + f$$

In this expression, substitute $\frac{u}{1 - uP}$ for v, and $\frac{1}{P}$ for f; we then arrive at

$$q = \frac{u}{1 - uP} + \frac{1}{P}$$

Combining, $$q = \frac{uP + (1 - uP)}{P(1 - uP)}.$$

In the numerator, (uP) and $(-uP)$ cancel, leaving

$$q = \frac{1}{P(1 - uP)}$$

So, the *distance* q from the eye's nodal point to the *image* is as follows:

$$q = \frac{1}{P - uP^2}$$

Then, the *divergence* Q of this image at the eye is $-\frac{1}{q}$.

$$-\frac{1}{q} = -\frac{P - uP^2}{1}$$

Therefore $Q = uP^2 - P$

Let's take a specific instance with lens P = + 5 D; when the object is 5 cm from lens P, u = .05 meters.

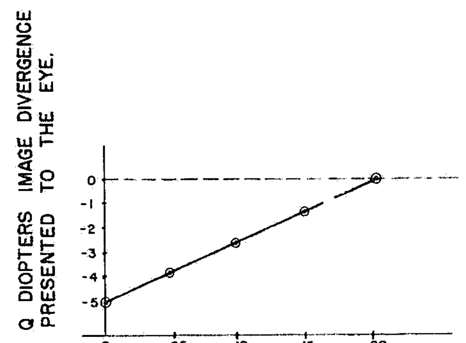
Then, $Q = .05\ (5)^2 - 5$

$Q = .05\ (25) - 5$

$Q = 1.25 - 5$

$Q = -3.75$ D

Thus, when the object is 5 cm in front of the + 5 D lens, the image seen by the eye stimulates 3.75 D accommodation. This and some other values for u are plotted on the graph below, where Q and u are seen to have a straight-line relationship. Thus, we can say that Q is directly proportional to u.



This relationship, the Badal principle, shows that the image divergence Q at the eye (that is, at F′) is directly proportional to object distance u, and, in the above example, F′ is located at N of the eye; but when we dealt with the *lensometer,* we positioned F′ of the field lens at the vertex of the "unknown" lens, and u was the distance from the field lens to the reticle target. It is clear that the two situations are exactly comparable, and now you can see what keeps the dioptric lensometer scale linear. Q.E.D.

CLINICAL POINT:

The exact method of using a particular lensometer should be ascertained from the instruction booklet accompanying the instrument. Usually, insufficient stress is given to the *proper* procedure for determining the power of a bifocal add; this is especially important when the add is incorporated into a high powered plus lens.

It is often stated that the correct method of reading the lens power of an add on the lensometer will depend on whether the add is on the front of the basic lens ("fused" type) or on its rear surface (Ultex type). This is only partly true. There is a more significant factor to be considered: any add, or for that matter, any *spectacle* lens — but *not* the theoretical *thin* lens with which we have been dealing — exerts a *different* dioptric power on object rays coming from a distant object compared to those arising up close! For example, a + 6 D spectacle lens has an effectivity of + 6 D on parallel object rays (with a vergence of zero) but only + 5.75 D on rays with a vergence of − 2.50 D from an object 40 cm away. This difference in power effectivity is related to the meniscus *shape* and the *thickness* of the spectacle lens. To demonstrate exactly why this is so, however, would require the use of "thick lens" formulas which I continue to ignore altogether.

As you already know, to obtain the basic *distance* power of any lens, you must use the lensometer to measure the *back* focal length; that is, an unknown lens must be turned so that its back surface faces *away* from the examiner. However, the lensometer can also be used to obtain a close approximation to the *near* effectivity (that is, near power) of this same lens if one measures its "*front* focal length"; this is because a spectacle lens' *front* focal length is a better indicator of the ability of a "thick lens" to deal with near objects. To determine this "front" focal length, you must turn the lens around in the lensometer lens clamp so that its back surface faces *towards* the examiner. The same procedure is necessary for measuring bifocal adds. (If the add on an aphakic lens is read with the lens in the "normal" instead of the "reversed" position, the indicated power will be too high in plus compared to its actual effectivity).

The correct way to find the power of any *bifocal* add on the lensometer is as follows: With the lens' back surface toward you, first read the total lens power through the add and then read the power *not* through the distant lens' axis but through a region comparably placed to the add, above the axis. (This allows you to compare powers as well as induced prismatic and distortional errors which *are* truly comparable between the two lens regions.) The two powers are then subtracted, the difference being the correct *add*.

What we have been speaking of is the add fused to the front lens

surface; but even when an add is ground on the *rear* surface, its power should still be determined the same way as just described. However, for the rear add, it makes somewhat *less* of a difference.

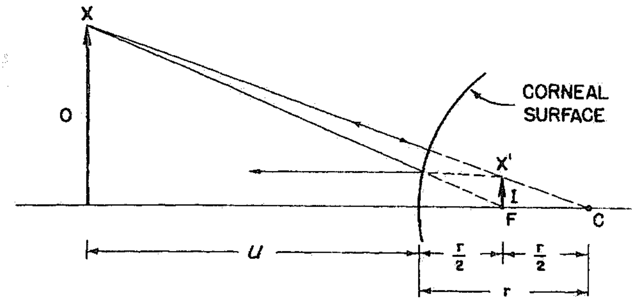
Learn to read the bifocal add correctly.

The Keratometer

This instrument for determining the corneal curvature is used mainly for the fitting of contact lenses. Its *modus operandi* is based on the ability of the anterior corneal surface to act like a highly-polished convex mirror and reflect light. The optical design of the keratometer allows a measure of the *size* of the reflected image and thereby, the radius of curvature of that anterior corneal surface can be ascertained. *That* is the key of keratometry.

As we learned when we studied reflection, the radius of curvature of a mirror will determine the *size* of the image it produces from a given sized object. So, if we illuminate an object of *known* size, place it a *known* distance from the cornea, and are able to *measure* the size of the reflected image, we can deduce the radius of curvature.

Since the cornea is such a high-powered mirror (− 250 D), an object does not have to be *too* far away to be effectively at the cornea's "optical infinity". (In the refraction section, I demonstrated this to you with an example). With the object at optical infinity, the reflected image will be situated practically in the corneal focal plane, positioned halfway to the center of curvature. This is shown in the diagram below where object O is at a relatively short object distance *u* from the cornea. Image I is located at just about the focal plane of the mirror.



In the keratometer, the distance between the target object and its reflected image is fixed. This set distance is assured by the use of a small, fixed-focus, high-powered eyepiece telescope which will allow the corneal reflection to be in critical focus only when it is situated a certain, fixed distance away. The "object" is situated in the same mount as the telescope so when the image is seen in sharp focus, the distance between the object and image will always be the same. Thus, the distance from the object to the image $(u + \frac{r}{2})$ is constant (equal to some number K).

$$\frac{\text{Image size}}{\text{Object size}} = \frac{\text{Image distance from cornea}}{\text{Object distance from cornea}}$$

$$\frac{I}{O} = \frac{\frac{r}{2}}{K - \frac{r}{2}}$$

Since K is quite large compared to $\frac{r}{2}$ ($\frac{r}{2}$ is only about 3.8 mm),

$K - \frac{r}{2}$ is approximately equal to K

So, $$\frac{I}{O} = \frac{\frac{r}{2}}{K}$$

$$r = 2K\frac{I}{O} = \frac{2K}{O} \cdot I$$

Since distance 2K and object size O are known and fixed, the radius of curvature r is exactly proportional to I, the size of the corneal

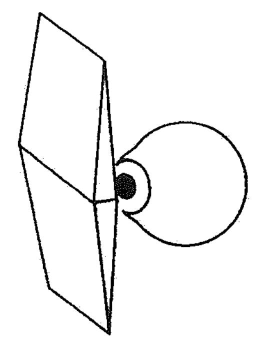
image. Find I, multiply it by a known constant, and you have what you are looking for — r.

The trick is how to *measure* image size I. It is usually quite small, but that is not the problem since we could magnify it optically. If we were trying to find the r of a comparably-sized steel ball, we'd have no difficulty measuring the size of the reflected image, since the ball will remain nice and still. Not so with the eye which is constantly moving about; trying to get it to hold still so that we might measure the size of the corneal reflection would be one mean chore. We need some help. Aha! enter the principle of "doubling".

"Doubling"

Let's demonstrate "doubling" using a biprism:

BIPRISM



Look in your trial lens set and you'll find a "lens" which has two small 5^{Δ} prisms mounted together in the same "mount" — their bases are in contact along a central line. Through such a biprism (while holding it so that the baseline is horizontal and "splits" the pupil) look at a distant, horizontal line. You should see *two* horizontal lines — one projected upward 5^{Δ}, the other downward 5^{Δ}.

CLINICAL POINT:

For the "squintologists" among you, if at the same time that the biprism is used in front of one eye the *other* eye looks at the same horizontal line, the latter will see that line located *between* the other two. Since binocular fusion has now been disrupted, if there is any tendency to a "cyclophoria", the central line (or the outer *pair* of horizontal lines) may appear tilted. Thus, any non-parallelity among the lines may indicate that a cyclophoria is present.

* * *

So a biprism "doubles" the view. Continue to hold the biprism baseline horizontal but shift your attention to a *vertical* line, say 10 cm long. You should see two vertical lines superimposed to a varying degree which depends on how far away you are from them; they could be quite separated if you are far enough away. As you approach the line, the two image lines will come closer together. When they just touch, you have "arrived"; you have created an optical condition that allows you to obtain an exact measure of the length of that line.

Let's reason this through:

Here each prism is 5^{Δ} and will cause a displacement of 5 cm at one meter. So, at one meter distance, one prism displaces the line image 5 cm up, the other, 5 cm down. If the images touch, the *total* displacement must be exactly 10^{Δ} or 10 cm at 1 meter, so the size of the object *must* be 10 cm. This is because "at touch" each prism must have displaced the line image a distance of one-half its actual length. So, if you know the distance from the prism to the object and you know the amount of prism causing the doubling, when the images touch, you can always learn the size of the original object.

The rule is as follows: the dimension (in cm) to be measured by the doubling device (there *are* others besides the biprism) is the product of the total deviation in prism diopters and the distance (in meters) between the device and the object to be measured. With the same biprism used above, if the "touch" of the inner extremity of the line took place when you were 40 cm away, the line would have to be (10^{Δ}) (0.4 m) or 4 cm long.

The great advantage of "doubling", and this *is* the mechanism employed in the keratometer to measure the size of the image reflected from the cornea, is that any oscillation of the cornea during measurement will affect both the doubled images equally; that is, motion of the eye will not cause the *separation* between the "doubled" images to change. This allows you to twiddle the knobs to arrive at the "contact" position accurately and without difficulty.*

So now you know how the image size **I** is measured and how the corneal radius r is attained.

The keratometer knob of most instruments reads in diopters of *refractive* power. This requires an assumption (not a *measurement)* of some refractive index n' for the cornea. Then, knowing n' together with the radius of curvature, you can find the refractive power P through the relationship $(\frac{n'-n}{r})$. This is the power indicated on the calibrated dial.

This is how the keratometer works. Be gracious and toss a little credit to the fantastic genius Helmholtz who discovered all this for himself!

The Keratoscope (Placido) Disc

This disc is usually a flat card or lighted source which presents a series of luminous, concentric circles. This disc is placed in front of the cornea at a 5 to 10 cm distance and the examiner views the reflection of these circular rings by the cornea through a small peephole cut in the center of the disc. There is usually (or should be) a

* By the way, the "doubling" principle is also used in the Goldmann applanation tonometer biprism, which optically creates the two "half-circles" you see from the "whole" fluorescein circle formed at the contact edge between the flat prism face and the corneal surface. If you increase the pressure of this prism against the cornea, you will increase the size of contact zone and thus the "half-circles" will enlarge in size. The biprism is built so that when the proper "contact" end-point is reached with this instrument, you will know you have flattened the cornea to a fixed diameter of 3.05 mm. *That* is the fixed size of the circle you are measuring with the biprism. The tonometer itself is calibrated to measure the force required to flatten the cornea to this fixed amount. It is this *force* (times area) that is the equivalent of the intraocular pressure you are measuring.

high plus viewing lens built into the peephole so that the viewer can get a very close look at the reflected rings without straining his accommodation too violently!

Corneal astigmatism, if present to more than a minimal degree, will show up as an elliptical, though regular flattening of the reflected circular object rings. Via this test, moderate to marked "with-the-rule" astigmatism would look to the observer as a series of horizontally elongated ovals. Irregular astigmatism, as that given by a conical cornea, will cause an asymmetry and distort the ring shapes out of perfect smoothness.

Keep in mind that to provide an *obviously* distorted image, the cornea must be quite distorted itself. If the distortion error is only a small one, while it may be terribly effective at reducing a patient's visual acuity, it is *not* likely to be visible by this crude test. *Much* more sensitive as an indicator of irregularity is the *retinoscopic* light reflex or even the "red reflex" given by an ophthalmoscope. (These were mentioned previously). Close observation of *these* reflexes (in contrast to those given by the Placido disc) will indicate even tiny surface or refractive disturbances of the cornea or lens. Remember, a distortion of the reflexes as seen through these instruments can be even a more sensitive indicator of trouble than a careful slit lamp examination of the refractive media.

You don't need a Placido disc to perform "keratoscopy". Every time you watch the reflection from the corneal surface of a moving hand-held flashlight — as when you are searching for a minor surface dent from a foreign body or a shallow ulcer — you are using keratoscopy. This flashlight method is just about as capable of indicating a corneal irregularity as is the Placido disc itself.

The Purkinje-Sanson Images

The catoptric (formed by reflection) images from the major eye surfaces are known by the eponym Purkinje-Sanson (P-S). The four major ones are created by the anterior and posterior surfaces of the cornea and lens. Only the posterior lens surface is a concave mirror; the other three act as convex reflectors.

In an experimental setup, if one reflects a standardized object source from an eye which is changing its accommodative status, there will be a relative shift in the size and position of the P-S images. This experiment will show that it is mainly the change in the anterior surface of the lens that is responsible for the functional physiology of accommodation.

The P-S images also allow one to *measure* the curvature of all four of the eye's curved surfaces. You have already seen how the first, very bright P-S image — that formed by the anterior surface of the cornea — permits the measure of the corneal radius of curvature via the keratometer. But one can also measure the others. However, when doing so, you must take something "extra" into account; the sizes of the other three P-S images are dependent not only on the reflection occurring at the corresponding surface, but also on the *refraction* of that image by each of the surfaces anterior to it *and* on the re-reflection possibilities of each of the images!

The clinical importance of the P-S images is that the anterior one forms the basis for keratoscopy and keratometry, plus the flashlight reflection tests (Hirschberg and Krimsky) which can grossly indicate the alignment status of the visual axes of the two eyes.