

Digital Imaging Colposcopy: Corrected Area Measurements Using Shape-from-Shading

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Abstract—The quantitative measurement of areas on the cervix is of interest to researchers studying the natural history of human papilloma viral lesions. Measurement of areas from images obtained through a colposcope are, however, inherently in error due to the image being a two-dimensional projection of a three-dimensional object. The ability to correct for these errors through use of digital imaging colposcopy and a practical application of a shape from shading algorithm was developed in this study. The shape from shading technique requires empirical measurement of the relationship between observed light intensity and the viewing angle (referred to as a reflection map). It was found that a population mean reflection map provided a correction that was about as accurate as using an individual's own reflection map (making it unnecessary to measure a map for each exam). Digital red filtering of the images increased accuracy and precision of measurement.

Index Terms—Area measurements, colposcopy, imaging.

I. INTRODUCTION

DIGITAL imaging colposcopy refers to the conversion of the images of the cervix obtained through a colposcope into a digital format that can be displayed and manipulated by a computer. This is most conveniently accomplished by attaching a charged-couple device (CCD) video camera to a colposcope and passing the video images on to a frame grab board residing in a computer platform [1]–[4]. This arrangement allows for a real-time view of the cervix on a computer or video monitor and the capability to “capture” any of the images that are deemed useful.

There are a number of advantages to digital imaging colposcopy. For example, the images saved can be immediately evaluated for image quality and if not acceptable the image may simply be “captured” again. This eliminates the need to wait for the development of film to determine if a good picture was obtained. The digital images may also be subjected to a number of image-processing steps to accentuate various features of the image [4]. However, one of the most exciting opportunities afforded by digital imaging is the development of objective and quantitative techniques of characterizing colposcopic lesions.

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One characteristic of aceto-white lesions that has been found to be useful is the size of the lesion. The size of a lesion has been found to have a positive correlation with lesion severity [5]–[7]. Large lesions are more likely to be underdiagnosed than small lesions and it has been suggested that size be used as a criteria for treatment [8], [9]. Despite this recommendation, the consistent and accurate determination of lesion size is not presently a practical reality.

It is also known that many aceto-white lesions may regress and disappear on their own [10]–[12]. An objective means of measuring the area involved by a lesion would provide a consistent means of evaluating a patient serially over time. Early studies have indicated that such an approach might be feasible and beneficial [13], [14], particularly in the pregnant patient [15].

However, it has been recognized that area measurements obtained using a computer graphical approach introduces an inherent error due to the computer image being a two-dimensional representation of the cervix [2], [16], [17].

Digital colposcopy provides a convenient method to measure areas or distances in an image of the cervix. These measurements can be measured directly on the computer monitor by simply outlining or marking the area of interest using a computer input device such as a mouse or joystick. The area or distance can be automatically calculated by summing the number of pixels involved. Although this is a simple and fast procedure it can produce very misleading results since the image of the cervix on the monitor is a two-dimensional projection of a three-dimensional cervix. The measured distance or area can be either larger or smaller than the actual feature being measured depending upon the angle of view. These errors in measurement can be quite large (up to nearly 500%) and are particularly worrisome if used for lesion surveillance [17]. During surveillance, a change in viewing angle between exams could result in an apparent shrinkage of a lesion when in reality the lesion has enlarged.

We have investigated a solution to this problem by applying shape from shading technology to digital colposcopy [16]. This paper presents an empirical determination of mean reflection maps for the human cervix required for the practical application of shape from shading technology and demonstrates the use of such maps in correcting area measurements from the cervix.

II. METHODS

Volunteer female subjects ($n = 11$) were recruited from local gynecology practices and the Planned Parenthood organ-

ization. The subjects ranged in age from 18 to 52; four of the subjects were Afro-American, and seven were Caucasian. All subjects signed IRB approved informed consent forms.

Colposcopy exams were performed in a typical fashion using a ZeissTM Photocolposcope with a video adapter. The cervix was exposed using a Grave's speculum with black interior blades to reduce reflected light. The cervix was washed with normal saline followed by 5% acetic acid for at least 1 min. A standardization target [consisting of a 4×7 mm stiff paper target with two circular spots of different known gray level intensity, see Fig. 5(b)] was placed on the cervix within the colposcopic field of view using 6-in alligator forceps. The area designation target [a piece of thin filter paper with an array of 1.5-mm diameter black dots, see Fig. 5(a)] was then absorbed to the mucosa surface. This target was gently pressed with a wet cotton swab to insure that it conformed to the shape of the mucosal surface. Images were grabbed with the computer imaging system and the area designation target removed using the alligator forceps. A second image was then obtained using the normal illumination (white light from a Xenon high intensity bulb). A third image was then obtained with a Wratten 23A optical filter [18] swung into the colposcope light path. These images were obtained quickly, with minimal subject or colposcope movement, so that the images were reasonably aligned. This is simple to accomplish in practice. The analysis software, described below, is capable of further alignment of image pairs to compensate for small degrees of movement.

Two different computer imaging systems were used connected to a Sony charged-coupled device (CCD) color video camera through a video splitter. The first system was a Macintosh Quadra 840TM audio/visual computer (based on the Motorola 68040 microprocessor, 40 MHz, with 2 MB VRAM) which is capable of capturing 24-bit full-frame color images. The second system was an IBM-compatible PC with a 486 IntelTM processor (66 MHz) running under WindowsTM operating system with a Coreco OculusTM frame grab board capable of capturing full-frame 8-bit grayscale images. All images were stored as tagged information file format (TIFF) format images on a magneto-optical disk.

Correction of test image intensities to match those of a reference image involves the determination of a gain and offset value for the test image. This is accomplished by using a standardization target with two circular dots of known intensities. The reference image is an image of two strips of the same reflectivity as the circular dots, wrapped around a known cylinder obtained with a colposcope under average lighting conditions. The relationship between these images can be determined considering that image irradiance $E(x, y)$ can be described as an energy balance equation relating the irradiance on an image of an object to the radiance from the object. It can be expressed by

$$E(x, y) = L_o r(\hat{n}(x, y)) \quad (1)$$

where L_o is the radiance of the light source, $r(\hat{n})$ is the reflectivity of the object as a function of the surface normal direction, and \hat{n} is the unit normal vector to the surface corresponding to the point (x, y) in the image. Under the

assumption of linearity for all processes affecting the transformation of the image irradiance into pixel values we can write

$$I(x, y) = \alpha r(\hat{n}(x, y)) + \beta \quad (2)$$

where $I(x, y)$ is the image pixel intensity measured by the CCD camera, α (which includes L_o) and β are linear transformation variables that may vary from image to image. On the other hand, $r(\hat{n})$ depends only on the reflecting properties of the object in the scene and the locations of the light source and viewer. Since the light source and viewer are always in the same direction during colposcopy, the reflectivity r becomes a function of the cosine of the viewing angle ω only

$$I(x, y) = \alpha r(\cos(\omega)) + \beta. \quad (3)$$

This allows for the description of the observed illuminance of the standard strips as a function of viewing angle

$$I_{s1} = \alpha r_1(\cos(\omega)) + \beta \quad (4)$$

and

$$I_{s2} = \alpha r_2(\cos(\omega)) + \beta \quad (5)$$

where I is the observed illuminance of strip 1 or 2, α is a constant incorporating the illuminating light intensity and system gain, β is a linear transformation constant, ω is the viewing angle, and r is the albedo of strip 1 or 2.

The viewing angle with respect to the standard circular dots in the test image is determined by calculating the inverse cosine of the ratio of the apparent minor-to-major axes of the standard dots. The observed illuminance is determined from an average pixel intensity of the standard dots. This provides us with analogous equations from the test image except that the illumination intensity, gain, and offset may be different

$$I'_{s1} = \alpha' r_1(\cos(\omega)) + \beta' \quad (6)$$

and

$$I'_{s2} = \alpha' r_2(\cos(\omega)) + \beta'. \quad (7)$$

From these equations, the following relations can be derived:

$$\alpha' = \left(\frac{(I'_{s2} - I'_{s1})}{(I_{s2} - I_{s1})} \right) \alpha = A\alpha \quad (8)$$

$$\beta' = \left(\frac{(I_{s2} I'_{s1} - I_{s1} I'_{s2})}{(I_{s2} - I_{s1})} \right) + A\beta = B + A\beta. \quad (9)$$

Now if we consider the intensity of an independent pixel in the new image that is not a standard target, but instead a pixel representing the cervix with albedo r_3 , we have a new intensity I' equal to

$$\begin{aligned} I' &= \alpha' r_3 + \beta' \\ &= A\alpha r_3 + A\beta + B \\ &= A(\alpha r_3 \cos(\omega) + \beta) + B \\ &= AI + B \end{aligned} \quad (10)$$

where I is now the intensity that would have been observed in the reference image, i.e., the intensity has been normalized to that of the original reference image. The value I' is easily

transformed to that of the reference image by

$$I = (I' - B)/A. \quad (11)$$

Image analysis was performed using the Colpo 3D™ (Western Research Company, Tucson, AZ) software package. This software can open TIFF images from either the Macintosh or PC platforms and perform quantitative analysis on the images, including scaling the images, performing area measurements and pixel intensity measurements, automatic analysis of the standardization targets to standardize image intensities and implementation of the shape-from-shading algorithm for the correction of area measurements. Colpo 3D™ software can compensate for slight movement between image pairs, obtained as described above, by realigning the images. Since the shape-from-shading algorithm for area correction must be applied to gray scale images, the Colpo 3D™ software automatically converts 24-bit color TIFF images into a gray scale image and a digital red filtered version. Areas of interest to be measured are magnified by zooming in on the image and are traced using a mouse pointing device.

Statistical analysis, such as the analysis of variance, was performed using the Statistica™ software package.

III. RESULTS

Shape from shading is based upon the observation that, when illuminating with a point light source, the amount of light reflected from an object is a function of the cosine of the viewing angle and surface properties of the object. Since the light source for most modern colposcopes is a fiberoptic “headlight” which illuminates the cervix from some distance, the criteria for a point light source is very well approximated during colposcopy. It would, therefore, be expected that images of the cervix obtained through a colposcope would vary in intensity as a function of the viewing angle between the cervix and the colposcope [16]. This phenomena is difficult to appreciate visually by direct inspection, however, becomes readily apparent when a line plot of light intensity across the image of a cervix is examined (for example, see Fig. 1).

The line plot in Fig. 1(c) shows a generally decreasing curve in intensity as the plot moves away from the OS (particularly evident to the left of the OS). This is due to the increasing curvature of the cervix away from the colposcope which results in a continually increasing angle of view. The spikes and irregularities along the plot from the white-light image correspond to local high-frequency features such as blood capillaries and aceto-white epithelium. The solid line represents a polynomial fit to those points corresponding to regions of the cervix that are normal. This line delineates the interpolated intensity that would be expected to arise from the curvature of the cervix, while the difference between the fitted line and the measured intensities represent the effect of the white epithelium and blood vessels.

The image can be subjected to red filtering (basically the opposite of the traditional green filter used to accentuate capillary features) to significantly reduce image contrast due to local features, but maintain the shading effect due to the changing slant angles. An example of this technique

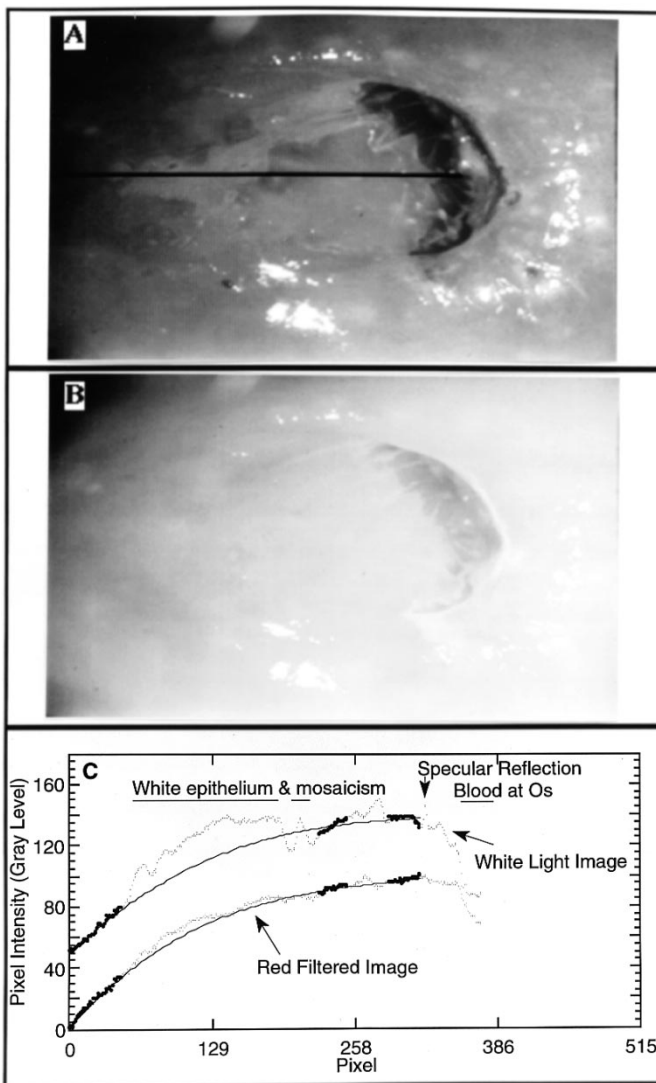


Fig. 1. Pixel intensities across gray scale and red filtered image of the human cervix. (a) Gray scale image of the cervix including the OS, and a white epithelium lesion with mosaicism that was determined to be CIN II on biopsy. The dark horizontal line marks the region from which a line plot of pixel intensities was obtained. (b) Computer-extracted red filtered image from the same 24-bit color image from which the image in (a) was derived. (c) Line plot of pixel intensity (dots) as a function of pixel location in either image (a) (white light image) or image (b) (red filtered image). The regions corresponding to the white epithelium lesion, blood, the OS and specular reflection are marked. The pixel intensities that occur at regions of the mucosa that are normal are shown as emboldened dots. The solid line is the third-degree polynomial fit to those emboldened dots for the respective curves.

is shown in Fig. 1(b), where the image from Fig. 1(a) was digitally filtered by a computer algorithm to extract an image comparable to the use of an optical red filter. A line plot across the same area as Fig. 1(a) shows clearly the loss of high-frequency features, while the angle-dependent shading effect is preserved (see Fig. 1(c)). A polynomial fit to same areas representing normal epithelium after red filtering is shown. In this case, the polynomial fit and the measured intensities are nearly coincident over the entire cervix, including the regions of white epithelium. Red filtering, therefore, is a very useful technique to eliminate the contribution of white epithelium or blood capillaries from the background shading information.

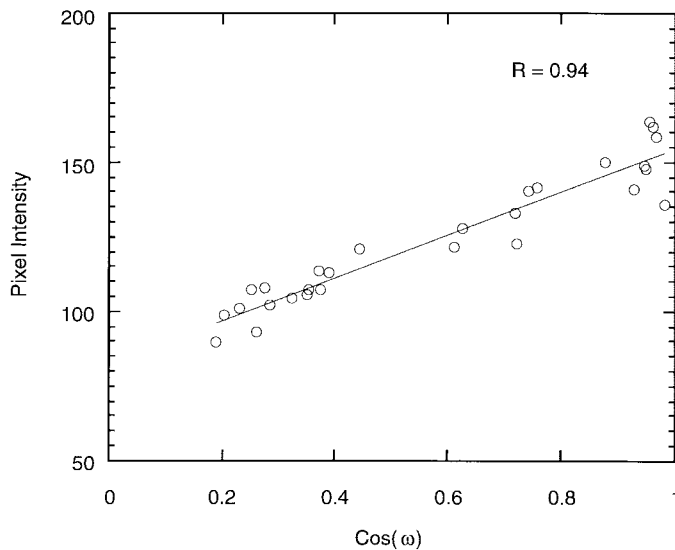


Fig. 2. Reflection map of the cervix. The pixel intensities in the image of a cervix are plotted as a function of the cosine of the viewing angle (ω). The linear correlation coefficient R is 0.94. The reflection map is defined as the linear relationship between the $(\cos \omega)$ and the light intensity.

The relationship between the intensity of light reflected from the cervix and the viewing angle is referred to as the reflectance map of the cervix. The reflectance map is determined by physical characteristics of the surface of the cervix (i.e., surface albedo), the wavelength of the illuminating light, and the intensity of the illuminating light. Theoretical considerations argue for a linear function of light intensity with the cosine of the viewing angle [16] and as described in (3). We assume that the viewer and light source are always simultaneously at the same angle with respect to the cervix, which is the case for a typical colposcopic exam. It should be noted that while the viewing angle can be determined from the light intensity, the sign of that angle is ambiguous and we do not recover a three-dimensional coordinate. A reconstruction of the three-dimensional structure of the cervix is not possible, however, the local slant angle, a function of the three-dimensional structure, is sufficient information to correct observed areas to actual areas.

To verify this relationship with routine colposcopy, the reflectance map was measured empirically by imaging the cervix and plotting the recorded light intensity (i.e., image pixel brightness) as a function of the cosine of the viewing angle (ω). The viewing angle was determined by the geometrical properties of circular dots that were placed on the cervix, as described in the methods. Fig. 2 shows a representative reflection map (i.e., observed light intensity as a function of viewing angle) measured from a human cervix. The graph confirms a linear relationship of pixel intensity with the cosine of the viewing angle with a correlation coefficient of 0.94. The pixel intensity is less at greater viewing angles (i.e., smaller $\cos(\omega)$) which is the expected result.

Since the intensity of light reflected from the surface of the cervix (i.e., the reflection map) is dependent upon the intensity of the light source, it is necessary to correct images for different lighting conditions to obtain a generally useful reflection map. This correction is also needed to correct

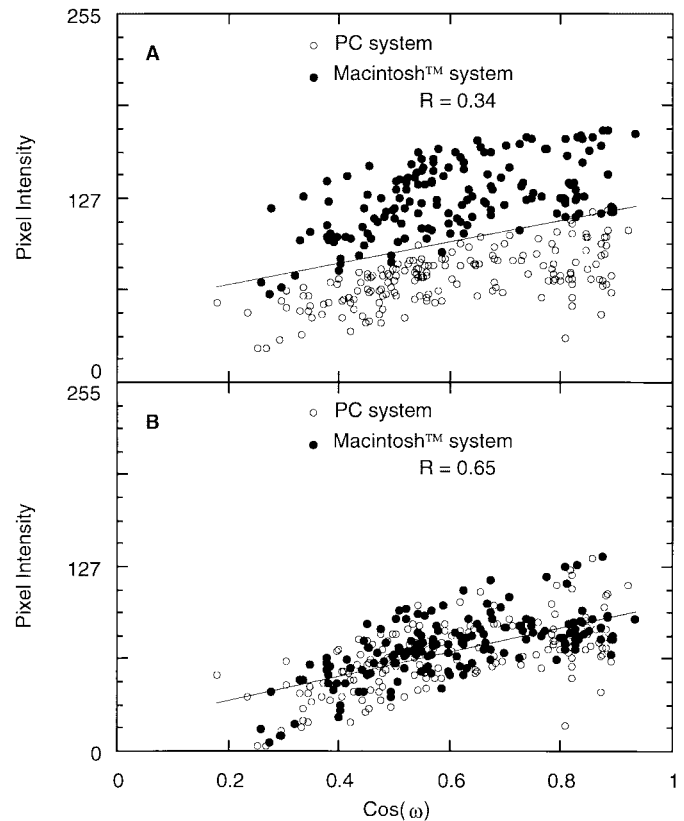


Fig. 3. Application of the standardization target to generalize intensity measurements. (a) Pixel intensities measured from images of 11 different imaging sessions with 11 different women obtained with either the PC-based imaging system (open circles) or the Macintosh™ imaging system (closed circles). (b) The same pixel intensities as in (a) after using the standardization target present in each image to correct for differences in the gain, offset, and illumination intensity between exams and systems. The linear correlation coefficients, R , for the data sets are shown.

for differences in gain and offset of image acquisition that are present when using different imaging systems. These corrections are accomplished by including a standard target within the field of view. A standard target contains two dots of different shades that have been well characterized with respect to their light reflecting properties. By measuring the intensity of these dots it is possible for the Colpo 3D™ software to compensate for the effects of the illuminating intensity, gain and offset. These measurements and corrections are performed automatically by the Colpo 3D™ software.

Fig. 3(a) shows the measurement of light intensity as a function of the cosine of the viewing angle obtained from 11 women using two different computer and frame grab systems. The differences in illumination, gain, and offset result in a wide range of values and the linear regression coefficient of only 0.34. There is an obvious systematic difference in the intensities obtained using the PC imaging system compared to Macintosh™ imaging system. Comparing the PC system to the Macintosh system using analysis of variance (ANOVA) indicates that the data sets do not estimate the same population regression ($F = 354, p < 0.00001$).

The standard targets were used to correct the image intensity measurements and a plot of the corrected intensities is shown in Fig. 3(b) for comparison. The data now appear

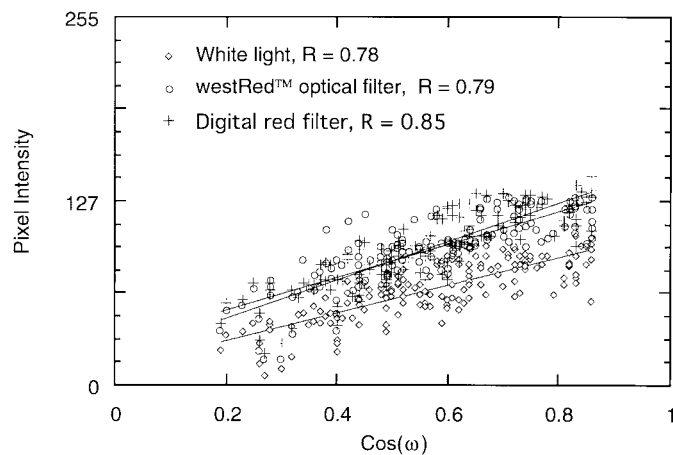


Fig. 4. Comparison of reflection maps obtained from white-light images and red filtered image. Pixel intensities from images obtained from 11 women were determined from normal white-light images, images obtained using a Wratten 23A optical filter, or after computerized extraction of a red filtered equivalent image (digital red filter) from the original white light image. These intensities were corrected using the standardization target included in the field of view and plotted as a function of the cosine of the viewing angle (ω). The linear correlation coefficient, R , was determined.

to be equivalent and this is confirmed with ANOV analysis which indicates that the corrected data do represent the same population regression ($F = 3.14$, $0.05 < p < 0.10$). The linear regression coefficient is increased to 0.65. Such a linear regression calculation provides a mean best estimate for the reflection map of the human cervix.

The reflectance map is also dependent upon the wavelength of light observed. Therefore, the use of red filtering of the image produces a different reflection map than that of white light. The reflection maps obtained using a Wratten 23A optical filter, the digital red filter provided in the Colpo 3D™ software, and white light are compared in Fig. 4. The red filtered reflection maps are clearly different from those of the conventional white light imaging. The red digital filter was designed to closely match that of the optical Wratten 23A filter and this is reflected in the similarity between these two reflection maps.

These experiments provide the average reflection maps from a group of women that can be used as the basis for calculating the viewing angle from the pixel intensities of digital images of the cervix. Knowing the viewing angle at each relevant image pixel allows for the computer to correct the area measurement by a factor equal to the inverse of the cosine of the viewing angle.

The performance of this approach was tested by placing on the cervix a small piece of tissue paper with a 1.5-mm-diameter test dot printed on the paper. The tissue paper absorbs tightly to the underlying mucosa and assumes the shape of the cervix (see Fig. 5(a)). The image of the dot is captured by the computer and then the dot immediately removed. A second matched image is then immediately captured without movement (see Fig. 5(b)). This results in a matched set of image pairs that differ only by the presence of the dot, with a known area, in one image. The apparent area of that dot can then be outlined in image one and that outline transferred by the computer to the second image. This results in an outline

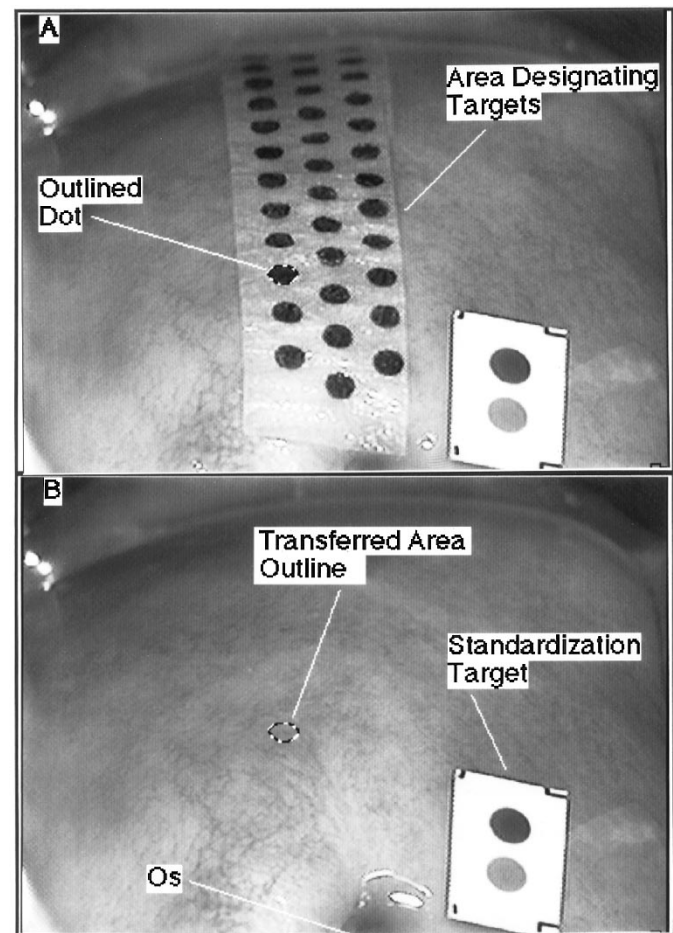


Fig. 5. Cervix image pair used to characterize performance of area correction algorithm. (a) The first image of a cervix pair that includes an area designation target that adheres to the surface of the cervix and assumes the shape of the cervix. One of the dark circular areas has been outline using a mouse pointing device. (b) A second image of the cervix obtained immediately after removal of the area designating target present in the first image. The area outlined in the first image has been transferred to the same location in the second image where area measurements can be obtained. The standardization target may also be noted in the field of view.

of a known area on the second image of the cervix. That area can then be calculated directly and after application of the shape from shading area correction algorithm. It is necessary to use two images since the shape from shading cannot be accomplished on the first image with the tissue paper target in the way.

As the viewing angle deviates from the 0° , orthogonal view, the apparent size of the dots decrease (this can also be clearly seen in Fig. 5(a) since all the dots are the same size but appear smaller as they get further from the cervix). Fig. 6 shows the area measurements obtained for the dots as a function of the distance of the dot from the cervix. The viewing angle increases with distance from the cervix. The magnitude of the error in area measurement can be seen to be quite large with some dots appearing to be only about 20% of their actual size. The correction of the area measurement using the shape from shading algorithm, however, results in a more consistent and accurate measure of the area of the dot regardless of the viewing angle.

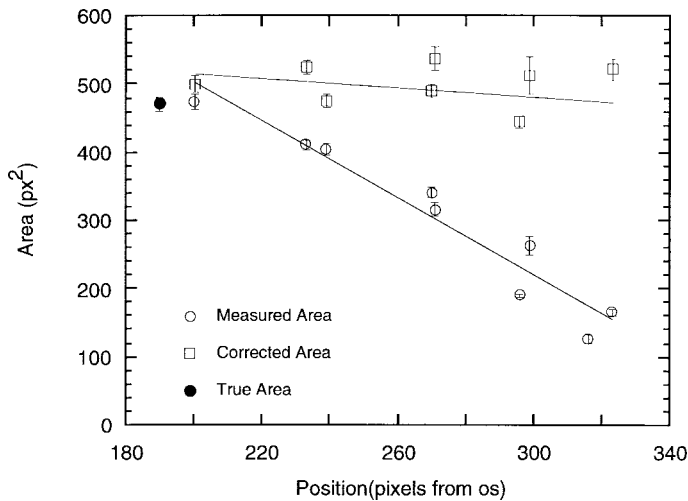


Fig. 6. Correction of area measures using shape-from-shading technique. The apparent area of circular dots placed at various distances from the OS of a subject were measured directly from the image (open circles). The measured areas were then corrected using the mean reflection map determined as in Fig. 4. The true area of the circles was determined prior to application to the cervix. The areas were each measured four times and the error bars are shown.

TABLE I
COMPARISON OF AREA CORRELATION USING MEAN REFLECTION MAPS AND INDIVIDUAL REFLECTION MAPS FROM GRAY SCALE IMAGES AND DIGITALLY RED FILTERED IMAGES WITH THE TRUE AREA AND THE APPARENT AREA OBSERVED AT A $45^\circ \pm 4^\circ$ VIEWING ANGLE OF THE CERVIX

Type of measure	Uncorrected Measurements		Average Area (px ²)	SD [†] (px ²)	CV [‡] (%)
	n(areas)	n(subjects)			
True Area (at 0°)	4	na*	471	10	2
Measured (at 45°)	44	11	340	31	9.1
Corrected Measurements					
Correction method	n(areas)	n(subjects)	Average Area (px ²)	SD(px ²)	CV(%)
Gray Scale Image					
Mean Map	44	11	492	107	21.6
Individual map	44	11	498	87	17.5
Red Filtered Image					
Mean map	44	11	477	74	15.5
Individual map	44	11	468	56	12.0

*na, not applicable, true area was measured on a lab bench.

[†]SD, standard deviation

[‡]CV, coefficient of variation

The accuracy of this technique using an individual's reflection map was compared to that using an average reflection map determined from a group ($n = 11$) of women. This comparison was done for both a normal white light image and the digital red filtered image. Individual reflection maps were determined as described in Fig. 2. These data were pooled and the mean reflection map determined by a linear least squares analysis of the combined data (see Fig. 4). The area for four dots on the cervix that were at a viewing angle of $45 \pm 4^\circ$ for each of 11 subjects were measured. The area of each dot was corrected using either the subjects own reflectance map or the average reflection map determined for either a white light image or a digitally red filtered image (see Table I).

The true area of the dots was measured with the colposcope carefully aligned at a 0° angle to sample dots on a flat surface.

The true average size was 471 square pixels (px²) with a coefficient of variation of 2%. This variation represents the error of outlining the dots and any variation in dot size. The average apparent size of dots observed on the cervix at about a 45° angle was 340 ± 31 px². This underestimates the true dot size by an average of 28%. Correction of the area measurement using either the mean reflection map or the individual reflection map obtained from white-light images provide very similar results. They both slightly overestimate the true size by about 5%. The individual reflection map produces more precise measurements with a coefficient of variation (CV) of 17.5% versus 21.6% for the mean map.

The use of the reflection maps from red filtered images to correct the area measurements provide the best results with average area measurements that are within the error of the true area measurements. Again the individual reflection map provides a small increase in precision with a CV of 12% compared to a CV of 15.5% for the mean map.

IV. CONCLUSION AND DISCUSSION

We have empirically measured the reflectance map of the human cervix for both white-light images and red filtered images. The reflection map combined with the use of a simple standard target to normalize images obtained at different times or with different imaging systems allows for the correction of area measurements using a shape from shading algorithm. This approach allows for a computer software solution to the inherent errors in area measurement that can be accomplished in a nearly automatic and user-transparent manner by the computer. For example, under viewing conditions where the apparent area is only 20% of the true area, the computer can correct that area providing a consistent area measure regardless of changes in the viewing angle (see Fig. 6 as an example).

The most useful method was to analyze red filtered images. These images greatly reduce the effect of surface features such as mosaic vessels or aceto-white epithelium. The accuracy of the correction is greatest with red filtered images and the precision of the measurements is also maximized. The use of mean reflection maps have been found to provide nearly identical area measures as the use of individual maps with only a small decrease in precision. This has the important practical consequence that area measurements can be corrected using reflection maps stored in the computer without having to determine such a map for each subject being examined. Since mean reflection maps are now incorporated in the Colpo 3DTM software, it is not necessary to repeat these determinations with multiple images of the subject; rather a single image including a standardization target is all that is required to make a corrected area measurement.

The simplest means of producing the red filtered images is to obtain the original image using a color CCD camera and frame-grab system. The resulting color digital image can then be automatically processed by Colpo 3DTM software to produce a digital red filtered version of the image. Colpo 3DTM software can convert 24-bit color images obtained from either PC-based imaging system or the MacintoshTM system. However, a nearly identical result can be obtained using a

gray-scale CCD camera and frame-grab system in conjunction with an optical red filter. This approach generally requires grabbing two images (i.e., one normal image for optimal feature recognition and the red filtered image), but has the advantage of allowing the use of a more economical imaging system or the continued use of an older existing imaging technology.

We anticipate that the clinical utility of this method would primarily involve the serial measurement of the size of aceto-white lesions for purposes of investigating the natural history of these lesions or for documenting the regression of lesions during a wait and observe treatment protocol. The simplest approach would be to use a color CCD camera attached to a colposcope and connected to a personal computer with the ability to capture 24-bit color images. At the time of the colposcopic exam the physician would place a standardization target on the cervix in a manner which permits visualization of both target and lesion. The live image would be examined to ensure that the illumination levels were satisfactory and that the lesion of interest is unobscured and its boundary clearly visible.

The illumination settings would be done exactly as they are presently for colpophotography, i.e., care would be taken to ensure that the area of interest is not saturated, and light levels would be adjusted using the rheostat on the light source and/or the aperture settings on the camera lens. We reiterate that the system we describe here involves use of normalized light intensities, so variations in light levels from one image to another (resulting from variations in the lamp intensity, for example) are completely compensated in the algorithm due to linearity of the relationships between relative intensity and slant angle.

The satisfactorily illuminated image would then be digitized and immediately assessed visually to assure a proper field of view. If the image is not satisfactorily framed it takes only a moment to reimage the area of interest. The digital image would then be calibrated by clicking on four points on the standard target. This is the only calibration interaction required of the physician in the clinic; all of the actual calibration calculations are made automatically after the target area is identified by the four mouse clicks.

The lesion of interest would then be outlined using a pointing device, typically a computer mouse, and the corrected area measurement would be automatically produced using the mean reflection map provided with the software. Digital imaging colposcopy does enable the user to do some real-time manipulation of image properties, such as image contrast adjustment in the displayed image. This can sometimes be helpful in visualizing the boundaries of the lesion. It is important to appreciate that our methodology is addressed at providing correct area measurements of lesions that the physician sees. If the physician is unable to see a boundary to the lesion which can be completely outlined then the system breaks down to the extent that either no measurement can be made or only a lower limit can be placed on the size of the lesion. This should not be construed as a failure of the area-correction algorithm, but rather a limitation which applies to all examinations of any lesion which is not well visualized.

We note also that this method depends upon the linear imaging characteristics of CCD cameras and the point source nature of the illumination system. Scanned images of film-based colpophotography do not at present provide valid area corrections. Calibration of properly illuminated film images is a possibility but, for the present, clinical application of this technology will involve use of a digital camera with coaxial point source lighting, typically a CCD camera mounted to a photocolposcope. In addition, the accuracy of the area measurements of aceto-white lesions have not been verified by direct comparison of computed results with those obtained by *in vitro* measurement of surgical specimens. Such experiments are of clear interest and will be pursued in future work to further verify the system accuracy.

Clinically then, the process is simple: 1) during an otherwise normal colposcopic examination a small calibration target is placed on the cervix; 2) the illumination on the cervix is adjusted as usual; 3) an image is made, requiring a single keystroke (or trigger release); 4) the image is examined and, if satisfactory, the calibration target area is delineated with four mouse clicks; 5) the lesion is outlined in the image using the mouse; and 6) the area measurement function is selected (another keystroke or mouse click). At this point the program returns the correct area of the lesion for the physician's consideration. The entire process is quick and simple and provides the physician with a more accurate measurement for use in patient care management.

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