J. Soc. Cosmet. Chem., **39**, 121–132 (1988)

Improved optical discrimination of skin with polarized light

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Received July 20, 1987.

Synopsis

A polarized illumination system is described that optically segregates the skin into two distinct regions, making the assessment of the clinical condition easier. It is possible to highlight, preferentially, the light reflected from the dermis with suppression of light reflected from the surface of the stratum corneum and vice versa. Polarized reflection spectroscopy and polarized color photography are used to demonstrate the differences between reflected polarized light and light that is depolarized by multiple scattering in the skin. The former shows clear delineation of surface structure whilst the latter highlights the redness or erythemal condition of the skin.

INTRODUCTION

Visual assessment has always played an important part in evaluating the clinical condition of skin. Trained personnel can estimate the erythemal levels, the degree of roughness and dryness, and any other changes that can be visually identified. The major problem with *in vivo* evaluations is the variety of scales used (1,2) and the discreteness presumed (3). There is also the added difficulty of interpreting the descriptions used semiquantitatively to define the clinical conditions (4–7). Seitz *et al.* (8) have attempted to overcome these uncertainties by proposing the use of macrophotography of hands showing semiquantitatively determined clinical conditions and to use these photographs as reference standards. In this way, common reference standards would ensure long-term continuity of assessment.

As early as 1941 Dent (9) recognised the enhanced detail that could be photographed when using monochromatic light of decreasing wavelength. More recently his approach has been extended by using ultraviolet light, with even more dramatic effect (10,11). Any quantitative evaluation using this improved discrimination still requires subjective assessment.

In the past three decades alternative approaches based on electronic instrumentation have been developed. These have been successful in quantitatively evaluating particular clinical conditions.

Objective assessment of skin conditions using preferential interaction with light has been mainly based on reflection spectroscopic techniques. Using diffuse reflection spec-

troscopy, Edwards *et al.* (12) confirmed the earlier observation of Dent (9) that light penetration into whole skin was proportional to wavelength. Jacquez and Kuppenheim (13) confirmed that the erythemal condition was associated with the preferential absorption from a white light source of green light by hemoglobin. However, the demonstration by Jacquez *et al.* (14) that electronic measurements of intensity could be made to an absolute accuracy of 1.0%, clearly showed the potential of these devices to measure very small changes in reflected light intensity.

In the last two decades a number of small portable devices have been used to measure erythemal values in skin (15,16) with an accuracy and discrimination superior to that of the human eye (17,18).

The remaining problem when considering the evaluation of the clinical condition of skin is the impartiality of the electronic measuring device. Unlike the trained observer, a photosensor has no discretionary powers. Light from the surface of the skin containing topographical information cannot be measured to the total exclusion of light reflected from the bulk tissue containing information on physiological response. It was the aim of this investigation to construct an illumination system for the assessment of the clinical condition of skin which would highlight the surface condition to the exclusion of the erythemal condition or vice versa. To this end a polarized illumination system was developed.

EXPERIMENTAL

POLARIZED PHOTOGRAPHY

Although preliminary photography was successfully developed using a Polaroid SX70 camera to produce instant color prints, it was found that better results were obtained with higher quality photographic equipment as below.

The skin area was illuminated from opposite sides at approximately 20 degrees by two Xenon flashlamps (Broncolor Impact) and photographed from 1 m with a Hasselblad camera and 120-mm telephoto lens at f32 using Kodak Vericolor-2 type-S film. At least three photographs were taken of each specimen. A standard photograph for reference purposes was taken. Photographs of reflected polarized and depolarized light from the same specimen illuminated by polarized light were also taken. To achieve this, the system was modified by placing, over each flash lamp, a polaroid filter (18 inches by 18 inches) orientated to make parallel both the polarization vectors of each light beam and also the horizontal specimen plane. A third, rotatable, polaroid filter was fitted to the camera lens. When set parallel to the other polaroid filters, a polarized reflected light image was recorded whilst a perpendicular orientation gave a depolarized reflected light image. In each photographic field a bright aluminium plate partly covered with Kodak white reflectance coating was included as a reference for polarized and depolarized light intensity.

REFLECTION SPECTROSCOPY

Diffuse reflectance spectroscopy was undertaken with a Perkin-Elmer Lambda 7 UV/VIS spectrometer, at a fixed 4-nm bandpass interfaced to a Perkin-Elmer 3600 UV data station. This standard transmission spectrometer was modified with an external inte-

grating sphere attachment with quartz fibre optics. The attachment was fitted to a custom-made support frame (Neo-Tech Engineering) to ensure a light-tight contact of the sample port of the sphere onto the skin of human volunteers. Polarized reflection spectroscopy was undertaken with a polaroid filter fitted over the sample port, thus allowing only polarized reflected light, from the skin, to re-enter the integrating sphere. With this arrangement it was not possible to measure the spectrum of the perpendicularly polarized reflected light and an alternative procedure was used.

In vivo polarized reflection spectroscopy of human skin was undertaken with a modified Jobin-Yvon JY3CS computer-controlled spectrofluorimeter. An incident illumination assembly for use with thin-layer chromatography plates replaced the conventional cell compartment. An off-axis concave mirror was placed as close as possible to the normal axis of the incoming incident light beam to collect reflected light efficiently. In this way very small changes in the sample-to-mirror distance (<5mm) did not critically affect the measured intensity (<[2.5%]), as was the case with the design originally supplied (>[10%]). Visible-grade polaroid filters were mounted immediately after the exit slit of the excitation monochromator and before the entrance slit of the emission monochromator. Reflection spectra from sites on the palmar surface of the hand were recorded by running both monochromators synchronously over the same spectral range. Four nanometer slits were used throughout, and in this way spectral bandpass compatibility was maintained with the Perkin-Elmer Lambda 7 spectrophotometer.

Parallel and perpendicular polarized reflection spectra were recorded and are referred to by the orientation vectors of the polaroid filters in the incident and reflected light beams. Since this was a single-beam system, all reflection spectra were corrected for the wavelength-dependent transmission of the polaroid filters by recording the spectrum of a diffuse white standard. This standard was prepared by painting a thick film of Kodak white reflectance coating onto a bright aluminium plate 7 cm by 7 cm. The recorded, paired spectra of the skin were then normalised by setting the parallel polarized reflection value at 700 nm to unity. In this way only relative spectral intensity changes could be compared.

TRANSMISSION SPECTROSCOPY

The methods used to measure the spectral transmission characteristics of polaroid filters were determined by their size.

Filters small enough to cover a camera lens were measured with the Perkin-Elmer Lambda 7 operating in transmission mode. In order to eliminate contributions from wavelength-dependent polarization changes from the grating monochromator, the instrument background was normalized with a sheet of polaroid in the sample beam. A second (test) polaroid filter, set to the same transmission vector orientation as the first, was introduced into the cell compartment but closer to the exit window. In this way, spectral intensity changes would be due to light absorption only, as expected for an isotropic light source.

Large sheets of polaroid filter were measured using a monochromatic illumination assembly composed of a 2.5 kW Xenon arc lamp and collecting optics (Oriel Scientific) and a Monospek 600 monochromator (Rank-Precision). A 1.8-metre-long quartz fibre optic (Schott) conducted the light to the polaroid filter placed in front of a UV/visible

photodiode detection system (Model 7052, Oriel Scientific). In this way there was no restriction on the size of filter or on the location over the surface where spectral transmission was to be measured. Multi-reflection in the fibre optic reduced the inherent polarization from the grating monochromator to an acceptably low level (dichroic ratio = 1.04).

IN VIVO TEST SPECIMENS

Human volunteers showing a wide range in the clinical condition of their hands were recruited. Particular attention was paid to the degree of redness exhibited. Photography was undertaken with either palmar or dorsal surface uppermost. Reflection spectroscopy was always undertaken on the palmar surface close to the base of the thumb or near the edge, i.e., thenar and hypothenar eminence, respectively. These fleshy parts of the hand made it easy to obtain a light-tight seal to the spectrometer.

RESULTS

The spectral transmission characteristics of the various polaroid filters used throughout this study were found to be very similar. Representative values for absolute transmission are given in Table I. Photography with such filters would tend to skew the color rendering towards the red since the constancy of transmission from green to red (550–700 nm) is not maintained in the blue (400–500 nm), where the transmission changes by a factor of approximately 2, i.e., 32% to 14%. This is not an unexpected result with a filter showing a predominantly brown color.

COLOR PHOTOGRAPHY OF HUMAN SKIN USING POLARIZED LIGHT

The changes in the optical discrimination of skin condition when viewed under unpolarized and polarized illumination are clearly demonstrated in Figure 1 for a hand showing a minimal erythemal condition. Figure 1A is a conventional color photograph taken with unpolarized light. The picture is composed of sheen from light reflected near to the critical angle and with modulated light intensity from the fine surface structure superimposed on the pink to red of the underlying tissue. Figure 1B is the same feature photographed in polarized light with the transmission vector of all polaroid filters parallel. As can be seen, there is attenuation of the underlying red and enhancement of the sheen and the surface detail. Finally, the complementary image, Figure 1C, where only

Table I
Wavelength-Dependent Light Transmission Characteristics of Visible-Grade Polaroid Filters

Wavelength	% Transmission
700	39
650	36
600	34
550	34
500	32
450	26
400	14

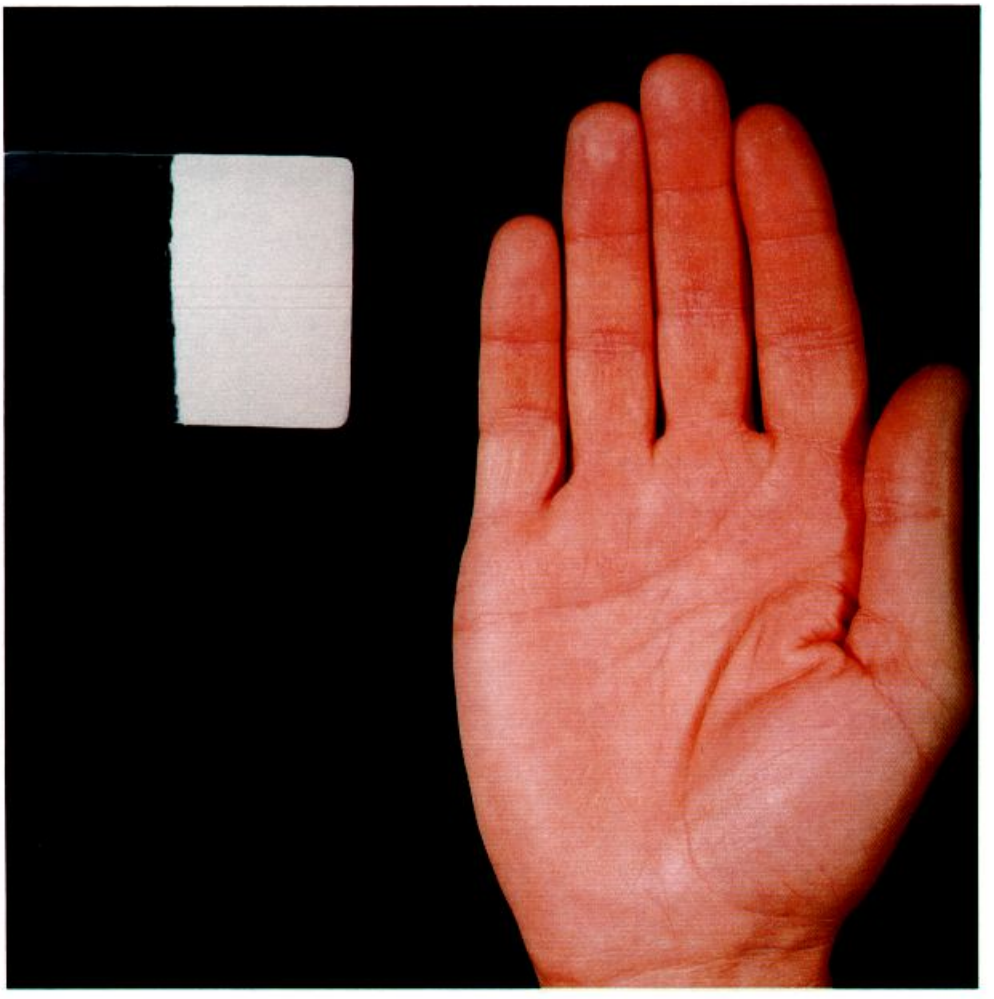

Figure 1. Color photography of the hand: A) Using conventional flash illumination. B) Using polarized flash illumination and a polaroid filter on the camera lens oriented parallel to the incident light. C) As (B) but with the polaroid filter turned through 90° and crossed with respect to the incident light.

the orientation of the polaroid filter on the camera is changed to perpendicular with respect to those on the lamps, give a record of those areas where depolarization has occurred. It is clear that the distribution of red in the skin does not have any obvious superimposed contribution from surface reflected light. Not only has the sheen gone but also the bulk of the fine detail from the stratum corneum.

These differences are more obvious if the photographs are scrutinized with a $\times 3$ to $\times 5$ magnifying lens. The surface sulci are more clearly delineated in Figure 1B than in Figure 1A. Figure 1C, however, shows clearer delineation of underlying redness than Figure 1A. A mottled pattern is obvious in the depolarized image which is not at all clear in the unpolarized image. Major folds in the palmar surface and that of the fingers, where the stratum corneum is thinner, are clearly outlined in red with greater contrast than in Figure 1A. Under magnified examination a subtle network of fine white lines

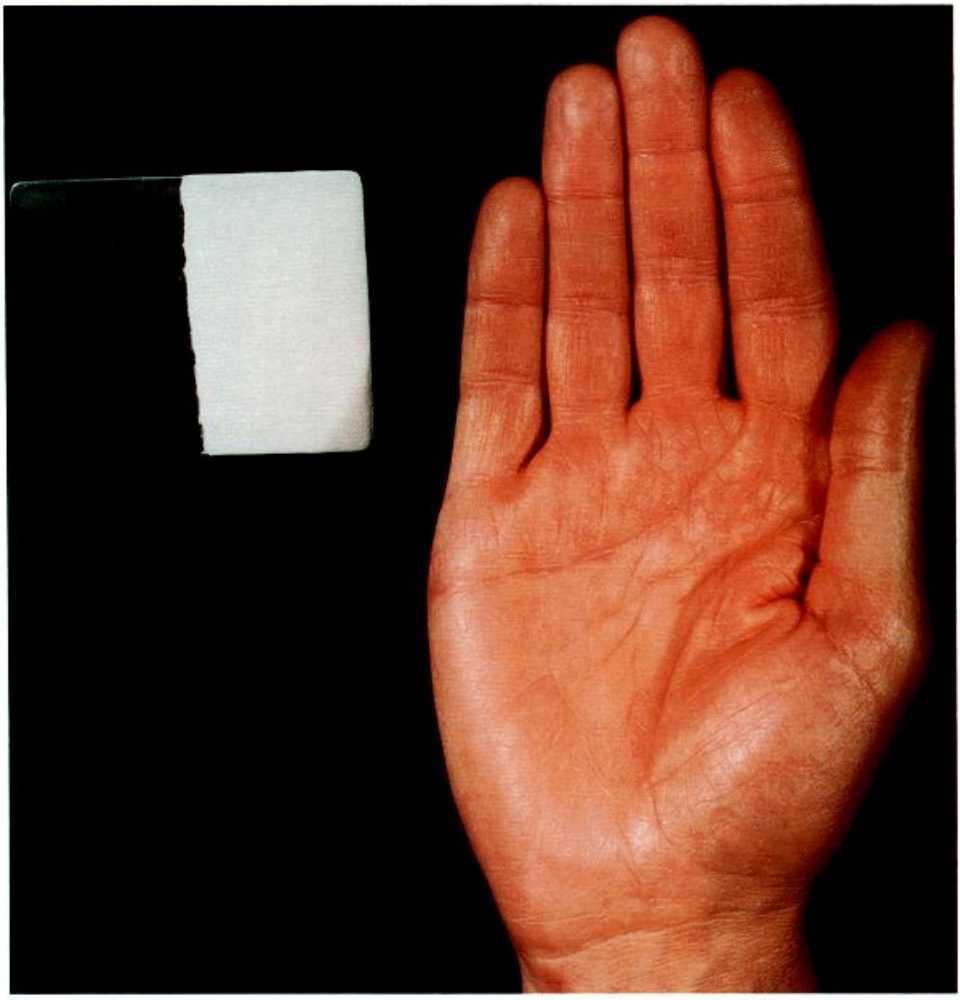

Figure 1B

can be seen within the skin folds over the joints of the fingers. These same areas have the appearance of dryness in Figure 1A and must originate from regions of surface scatter and depolarization of reflected light. A similar effect can be found on the palmar surface near to the base of the fingers and the hypothenar eminence.

The reference metallic surface in each photograph demonstrates clearly the near extinction of the polarized light with a crossed polaroid filter. Depolarized light from the reference white standard shows little change with the orientation of the polaroid filter on the camera, and therefore any observed changes in detail or color must originate from preferential interaction with the skin.

POLARIZED REFLECTION SPECTROSCOPY OF HUMAN SKIN

Figure 2 shows the reflection spectra from the hypothenar eminence of the hand obtained using an integrating sphere with the Perkin-Elmer Lambda 7 spectrophotom-

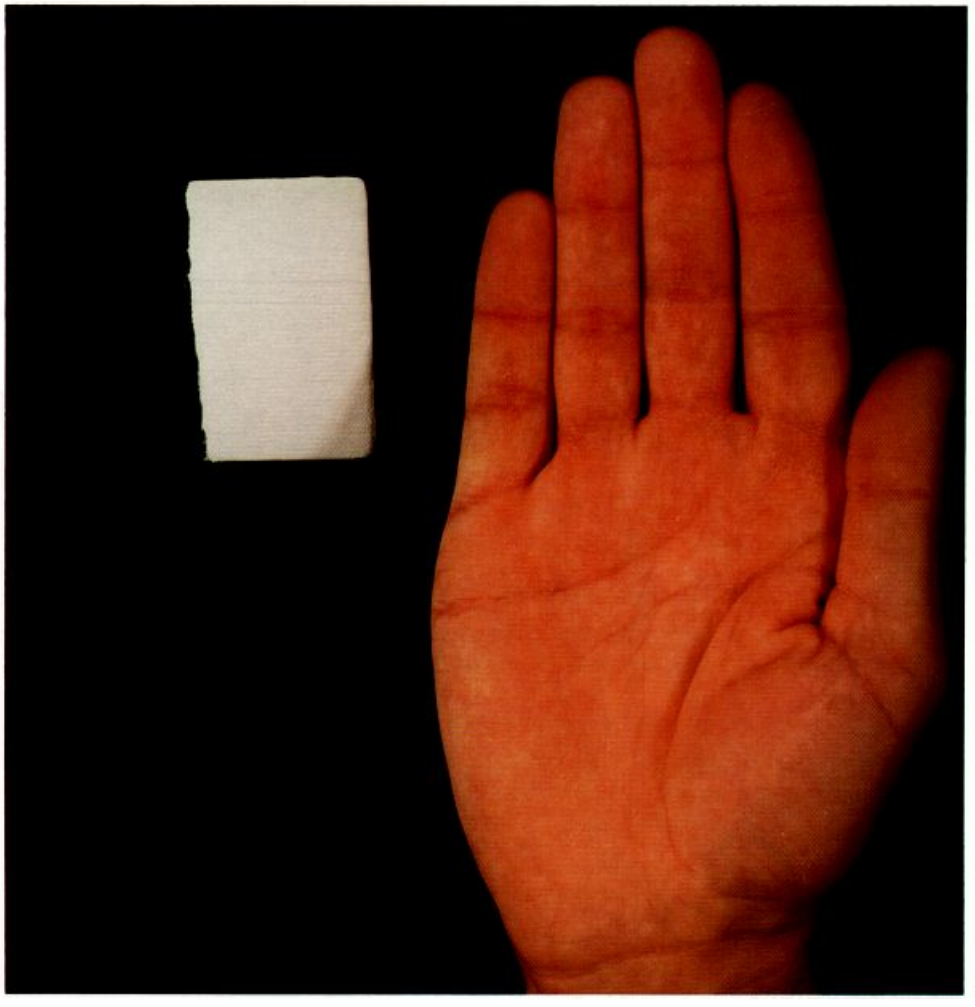

Figure 1C

eter. Contour A is a diffuse reflection spectrum typical of those published in the literature (14). Reflection is highest in the near infrared and red, i.e., 700 nm to 600 nm. The rapid fall between 600 nm and 400 nm is characteristic of absorption by hemoglobin with a doublet between 600 nm and 500 nm and the Soret band near 420 nm. Contour B, the polarized reflection spectrum, shows a completely different wavelength dependence, with very little contribution from hemoglobin and no appreciable change in the level of reflection from the far red (70%) to the violet (60%). The inflections near 570 nm and at 425 nm are probably hemoglobin, but the bands at 500 nm and 460 nm are new. The latter may, however, be present as a slight inflection in contour A.

Figure 3 is representative of the polarized reflection spectra from the hand using the Jobin-Yvon spectrofluorimeter in synchronous scan mode. Contour A was obtained with polarizer and analyser perpendicular and is the depolarized reflection spectrum. The major attenuations of the reflected light are again attributable to hemoglobin. In the polarized reflection spectrum (contour B), the overall level of reflected light inten-

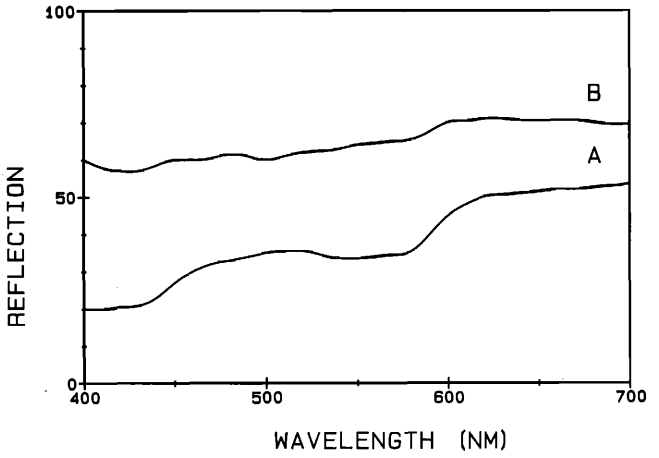

Figure 2. *In vivo* reflectance spectra of skin: A) Using an unmodified integrating sphere. B) Using an integrating sphere with a polaroid filter on the sample port.

sity is up (88% relative to 77% in A) and the attenuation by hemoglobin absorption is down (allowing only 32% reflection in A and 48% reflection in B). In this case, however, the contribution from the latter still dominates the spectral contour, unlike the equivalent spectrum in Figure 2. Contour C is the difference between A and B. The relative reflection difference is about 10% from 700 nm to 530 nm. The minima of about 5% at 620 nm and 600 nm give the impression that a small inverted hemoglobin spectrum contributes between 500 nm and 600 nm with the Soret band at 430 nm. This is superimposed on a small reflection continuum from 700 nm to 400 nm.

DISCUSSION

It is unfortunate that in subjective assessments of the clinical condition of skin, it is often the subtle spatially dependent changes that supply the necessary information. The

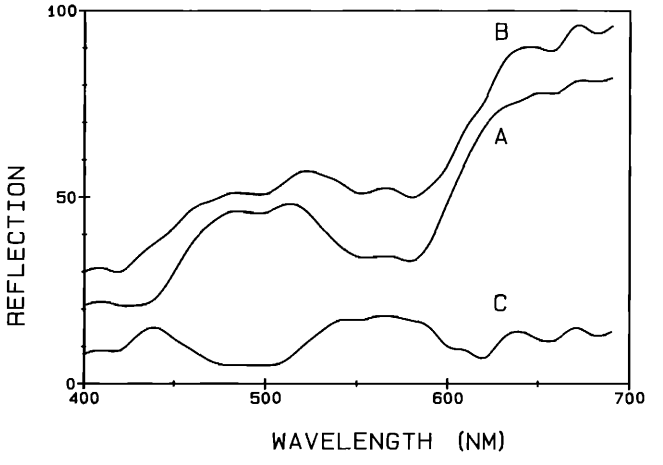

Figure 3. *In vivo* reflectance spectra of skin using a synchronously scanning spectrofluorimeter: A) Polaroid filters are perpendicular in the incident and reflected light beams. B) Polaroid filters are parallel in the incident and reflected light beams. C) (B)—(A) Difference spectrum.

origin of the reflected light, i.e., surface or bulk tissue of skin, is not a problem in subjective assessment since training allows the expert to look at, or into, the skin as required. Using polarized illumination and viewing the polarized and depolarized reflected light, there is available for the first time a means of preferential selection of pictorial information from different levels in the skin.

A clearer understanding of the mechanisms involved can be obtained by considering the schematic representation in Figure 4. Polarized light incident on the skin can reflect from the surface or penetrate into the skin. The former is simply regular reflection from an air-dielectric interface of refractive index 1.5 (19). At near normal incidence no change in the polarization will occur and 100% polarized light will be reflected. There will be some attenuation of the reflected light intensity where the angle of incidence with the non-planar skin surface approaches the polarization angle (20), i.e., in regions showing glare. Light that passes through the air-stratum corneum interface can only return to the surface by reflection from the bulk tissue of the skin. Anderson *et al.* (21), using diffuse reflection data, concluded that the principal contribution to the remittance of light originated in scattering from collagen fibres in the dermis.

In the current investigation not only is the scattering changing the direction of the incident light beam through approximately 180 degrees but it is also depolarizing the light. Teale (22) has shown that this is exactly what happens in turbid media, with each scattering event reducing the polarization, expressed as anisotropy, by 0.7 of its original value. Anisotropy is defined as

$$r = (I\,ll - I\,L)/(I\,ll = 2I\,l)$$

where I ll is the measured light intensity with polarizer and analyser parallel (parallel

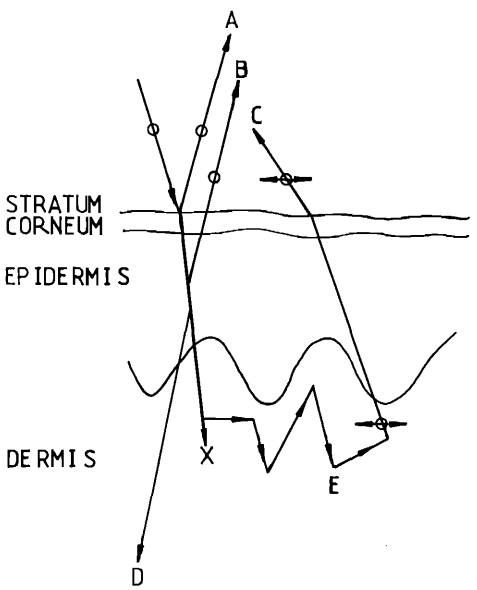

Figure 4. Schematic representation of optical paths through skin with polarized light: ○, Light polarized perpendicular to the plane of the page. ⊖, Depolarized light. A) Surface reflection. B) Epidermal remittance. C) Dermal remittance. D) Forward scattered. E) Dermal scattered. X) Absorption.

polaroids). I L is the measured light intensity with polarizer and analyser perpendicular (crossed polaroids).

Where scattering is most efficient, in the dermis, leading to depolarization, visible light has the highest probability of being absorbed by the hemoglobin in the peripheral blood supply to this region. Each change in direction with a scattering event amounts to an increase in the optical pathlength in an absorbing medium. It is to be expected, therefore, that progressive depolarization would be accompanied by increased spectral attenuation by hemoglobin. As a consequence, there would be a preponderance of reflected red light and an enhanced contrast of those regions that are normally pink or erythematous. This is exactly the effect demonstrated by the color photography of the depolarized light component of the remitted light.

Surface reflected light is all but eliminated by the crossed polaroid on the camera lens since there is a dramatic loss of surface detail when comparing Figure 1C with 1B. The enhanced redness in the latter cannot be a photographic artefact or the reference white in each image would show different hues. Such an effect can be seen when comparing the reference whites in Figure 1B and 1C with that in 1A where no polaroid filters were present in the optical train.

The calculated anisotropy of the remitted radiation at 650 nm (taken from Figure 3), at approximately 0.06, requires multiple scattering to reduce the residual polarization to such a low value. As many as seven scattering events would be required if the anisotropy is reduced to 0.7 for each scattering interaction in the skin. It is not possible to be precise since the calculated anisotropy does not refer uniquely to the dermal remittance but includes polarized reflection from the surface of the stratum corneum. The site-specific sources of the polarized and depolarized light cannot be included in any spectrophotometric corrections that can be undertaken. It must be presumed that the anisotropy value is an upper limit.

Although the depolarization component has been discussed with respect to only one orientation, the isotropic nature of the light requires an equal intensity for the opposite or perpendicular orientation. Both spectral and photographic recordings of the polarized reflected light therefore contain a depolarized light component. This is consistent with the residual erythema shown in Figure 1B and the hemoglobin spectrum in Figure 3, contour B. The difference between the polarized and depolarized spectra, contour C, is therefore closer to the corrected polarized reflection spectrum if depolarization is complete before remittance of the light from the skin. Although there may be some contribution from an inverted hemoglobin absorption spectrum to the difference spectrum, it is small. No significance can be attached to the absolute intensity, since spectral normalization was undertaken with the polarized and depolarized contours. The featureless nature of the reflection is consistent with the epidermal reflection spectrum published by Anderson *et al.* (21). The lack of spectral detail in the difference contour gives it a superficial similarity to the polarized reflection spectrum obtained with the integrating sphere, Figure 2, contour B. There is a major inconsistency, however, in that the latter shows much higher absolute reflection intensity, but it must be remembered that the difference contour was generated from normalized spectra. Part of the problem relates to the instrumental background correction using a standard white reference overlaid by a sheet of polaroid film. However, application of a sheet of polaroid to the skin with sufficient pressure to ensure a light-tight seal to the integrating sphere would result in compression of the skin. Pressure constriction would reduce blood flow

and hence the measured contribution from hemoglobin absorption to the reflection spectrum. Depolarization due to multiple scattering would therefore give higher available light intensities in the absence of hemoglobin, with the parallel oriented vector of the isotropically reflected light contributing to the polarized spectrum. Any additional spectral absorption must be attributed to other light-absorbing moieties that are normally lost under the hemoglobin spectrum.

Absence of pressure on the skin is therefore essential if quantitative data are to be compared with the pictorial contrasts shown in Figures 1B and 1C. Since no window was present when the spectrofluorimeter was used to record polarized and depolarized spectra, the differences shown in Figure 2 are probably better semiquantitative evaluations of the relative pictorial differences of Figures 1B and 1C. This result is confirmation of the skill of the trained assessor in that all surface topography information is based on evaluating small differences in the amounts of reflected light.

When trained assessors score a clinical condition, they can orient the specimen or change the viewing angle in order to optimize the contrast in the feature of interest. The major advantage of polarized and depolarized image recording is in the optimization of contrast without the need to orient specific regions.

Since the polarized image is a composite of polarized and depolarized reflected images, pictorial subtraction of the superimposable depolarized image should leave a difference image of polarized reflected light only. This would have little color and contain only detail of the surface topography of the stratum corneum. If depolarization of light due to surface scattering from rough or dry skin were present, as is suggested by scrutiny of the magnified fine surface detail from Figure 1C, then these features would be attenuated or lost from the difference image.

This may appear to reduce the optical segregating power of the technique but could be used to advantage by pictorial manipulation with an image processor capable of working in color.

A measure of the enhanced redness in the depolarized image would allow *in vivo* quantitative analysis to be undertaken with the minimum contribution from skin surface reflection. The latter, however, could still be present as superimposed white or off-white detail since it would arise from fine surface detail depolarization.

If discriminating white from pink or red was possible, then the fine surface scattering sites on the skin could be distinguished from the gross surface topography, an operation which the trained assessor does well. The advantage of objective image analysis, however, lies in the very small changes which can be quantitatively determined using an electronic sensor with a finer discrimination of light intensity than is possible by eye.

ACKNOWLEDGEMENTS

The authors wish to thank Justin North of Graphic Design Section, Unilever Research, Colworth House, for the color photography and preparation of the prints.

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