

Imaging the Pigments of Human Skin with a Technique which is Invariant to Changes in Surface Geometry and Intensity of Illuminating Light

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Abstract:

A technique is described which enables quantitative histological data to be recovered from conventional digital images. Methodology is developed around the concept of image ratios, which are shown to be invariant to scene geometry and illumination intensity. Key to the success of this technique, is a function which maps uniquely from a vector of image ratios to the corresponding vector of histological parameters. The existence of this function is established using mathematical techniques drawn from differential geometry. The methodology is formulated generally then applied to a two-parameter model of human skin. A function relating image ratios to concentrations of melanin and blood is established and used to process a standard RGB image. The technique successfully maps out the distribution of blood and melanin across the entire image.

1 Method

As light optical radiation propagates through skin it is both scattered and absorbed. Scattering primarily occurs from the underlying tissue structure whilst absorption tends to result from the tissue pigments. Healthy skin can be considered the two-layered structure, depicted in figure 1. Incoming light first passes through the epidermis.

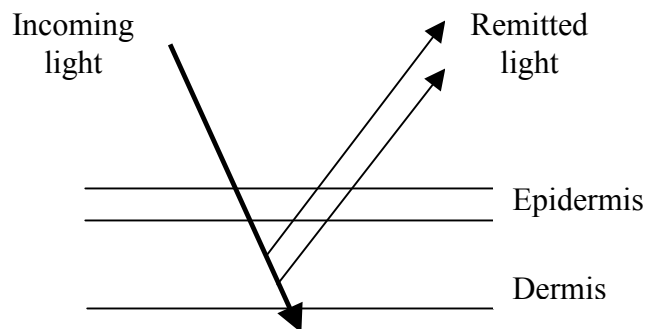


Figure 1: Tissue structure of normal skin

No scattering occurs in this layer but the presence of the pigment melanin causes a fraction of the incoming light to be absorbed. The light then passes into the dermis where it is scattered by the underlying collagen as well as being absorbed by the pigment haemoglobin. It has been argued [1,2] that the Kubelka-Munk theory [3] is sufficient to model radiation transport within skin. If scattering coefficients for collagen and specific absorption coefficients for haemoglobin and melanin are known, then it is possible to apply the Kubelka-Munk theory at a specific wavelength. This allows the corresponding fraction of remitted light to be predicted. By applying this theory at discrete wavelengths, across an appropriate spectral range, a remittance spectrum can be constructed.

In healthy skin three parameters are required to describe all histological variation: concentration of epidermal melanin, concentration of dermal blood and thickness of the dermal layer. It is convenient to think of the variation in terms of a 3-D parameter space, with axes: melanin, blood and dermal thickness. As the three parameters have differing effects on the remitted spectrum, every point within the parameter space corresponds to a unique spectrum, which can be obtained by using the Kubelka-Munk model of light transport. By convolving the spectrum with the spectral response curves of the image acquisition system, it is possible to obtain RGB values that correspond to a given point within parameter space. By constructing a mapping, relating RGB vectors to corresponding points in parameter space, it is possible to recover parameter values across a given image. This information can then be displayed in the form of grey-scale image, or parametric map. This fundamental principle has been used by Cotton and Claridge [4,5] to develop a system capable to analysing pigmented lesions. This application uses a four-parameter model of human skin, the three parameters already

described, with the addition of melanin in the dermal layer. This system has been developed into a commercially available system by Aston Clinica and is proving to be of immense value to clinicians in their diagnosis of melanoma. Although proving effective, the system requires exact calibration of the illuminating light source and does not take into account any variation in surface geometry. This latter assumption can result in inaccuracies when skin is imaged in the vicinity of a joint. In the following section a technique is described for recovering histological parameters from image data in a way that is insensitive to scene geometry and illumination intensity. This method is then applied to a two-parameter model of skin.

1.1 Achieving invariance to surface geometry and illumination intensity

The dichromatic reflection model, first proposed by Shafer [6], states that light remitted from an object is the sum of two components, the ‘body’ component and the ‘surface’ component. The body component refers to physical processes occurring after penetration of light into the material and the surface term to reflections that take place at the surface of the object. By using a system of cross-polarised filters on the illuminating source and the image acquisition system, it is possible to eliminate the surface component of reflection. This leaves only the body term, which is the product of a geometric factor and a colour term. The technique described here is applicable to problems, in which the spectral characteristics of the illuminating light source are known a priori. For such a system the illuminating light may be written as

$$E(\lambda) = \varepsilon_0 E_0(\lambda)$$

where ε_0 is a wavelength independent scaling factor determined by the intensity of the light source but which does not change with wavelength. This allows the dichromatic reflection model to be written as

$$i^n = \varepsilon \int E_0(\lambda) S(\lambda) R^n(\lambda) d\lambda$$

where $\varepsilon = \varepsilon_0 K$ and K is the geometric factor in the body term of the dichromatic reflection model. The function $R^n(\lambda)$ defines the spectral response of the n th filter and $S^n(\lambda)$ the remitted spectrum of the illuminated tissue. If an image acquisition system measures an $N+1$ dimensional vector of image values, then a vector of image quotients can be defined as

$$\mathbf{r} = \left\langle \frac{i_2}{i_1}, \frac{i_3}{i_1}, \dots, \frac{i_{N+1}}{i_1} \right\rangle \quad \mathbf{r} \in \mathbf{R}$$

where \mathbf{R} denotes the N -dimensional space of image ratios. All components of this vector will be independent of the constant ε and thus independent of illumination intensity and any geometrical factors in the imaged scene. The situation in which K histological parameters are required to describe all histological variation is considered and an appropriate parameter vector defined as

$$\mathbf{p} = \langle p_1, p_2, \dots, p_K \rangle \quad \mathbf{p} \in \mathbf{P}$$

where \mathbf{P} denotes the K -dimensional space of parameter variation. If a function exists which maps uniquely from any vector of image ratios to the corresponding vector of scene parameters, then it is possible to recover histological parameters from image data in a way that is insensitive to scene geometry and illuminating light. This idea, of dividing two image values, has been used successfully by Healey [7] who was able to identify metal and dielectric materials in a segmented image independently of scene geometry.

1.2 Establishing Uniqueness

Any function, which is to map from the space of image ratios to parameter space to must be 1-1. If this is not the case, ambiguity will arise as it could be possible to recover more than one set of parameter values from a given vector of image ratios. To establish this condition, it is first necessary to consider with the function f , which maps from points in parameter space to points in the space of image ratios. This function is a vector valued function of a vector variable and is defined as

$$\mathbf{r} = f(\mathbf{p}).$$

To implement this function, it is first necessary to compute the spectral reflectance of the material of interest for the given set of parameter values, or point in parameter space. This is done using the Kubelka-munk model of light transport with the appropriate parameter values. Using the computed spectral reflectance, along with the spectral responses each of the filters $R^n(\lambda)$ in the image acquisition system, a vector of image values can be calculated. From this vector a corresponding vector of image ratios can then be computed. To establish whether the function f is 1-1, the determinant of the Jacobian matrix, defined as,

$$\mathbf{J} = \begin{pmatrix} \frac{\partial f_1}{\partial p_1} & \frac{\partial f_1}{\partial p_2} & \dots & \frac{\partial f_1}{\partial p_K} \\ \vdots & \vdots & \dots & \vdots \\ \frac{\partial f_N}{\partial p_1} & \frac{\partial f_N}{\partial p_2} & \dots & \frac{\partial f_N}{\partial p_K} \end{pmatrix} = \begin{pmatrix} \frac{\partial r_1}{\partial p_1} & \frac{\partial r_1}{\partial p_2} & \dots & \frac{\partial r_1}{\partial p_K} \\ \vdots & \vdots & \dots & \vdots \\ \frac{\partial r_N}{\partial p_1} & \frac{\partial r_N}{\partial p_2} & \dots & \frac{\partial r_N}{\partial p_K} \end{pmatrix}$$

must be analysed [8]. If the determinant is non-zero at a point in parameter space then there exists a neighbourhood around this point where the function f can be approximated linearly. This means that any point within this region will map under a 1-1 mapping to a unique point in the space of image ratios. By discretising parameter space into suitably small intervals and establishing that the Jacobian is non-zero across the whole space, it is possible to establish the 1-1 condition for all possible parameter values. This can be thought of as analogous to the one-dimensional case where the absence of a zero derivative ensures no turning points and thus a 1-1 condition over a defined functional range.

With this condition established a function, g , can be defined as

$$\mathbf{p} = g(\mathbf{r})$$

which relates the vector of image ratios to the corresponding vector of parameter values. This is best achieved using some form of interpolation technique. This allows a piecewise continuous function to be constructed which is valid across the whole of parameter space. Using this function, parameter values can then be obtained at every pixel and corresponding parametric maps produced.

2 Results



Figure 2: (a) RGB facial image (b) parametric map of melanin (c) parametric map of blood

The technique was applied to facial images acquired using a standard RGB digital camera. As it is necessary to measure the same number of image ratios as histological parameters, a two-parameter model of skin was used.

The dermal thickness was measured using the system developed by Astron Clinica [9] and assumed to be constant across the face. This is thought to be a reasonable assumption as, although thickness varies between individuals, it is fairly constant for a relatively small area of an individual.

Using the responses of the imaging acquisition system along with the spectral characteristics of the illuminating light source, a 2-D vector of image ratios was computed for every point in a discretised parameter space. From a consideration of the determinant of the Jacobian, uniqueness was established. Using this discrete data was constructed using a triangle-based cubic interpolation method which was implemented in matlab. This function was used to process the image shown in figure 2a to produce the parametric maps of melanin and blood. These have been shown in figures 2b and 2c respectively.

The images show that the method is able to differentiate between melanin and blood born pigments. The melanin image demonstrates how moles are detected, there being two under the left eye which do not show in the blood parametric map. The images also demonstrate the uniform distribution of melanin across the face. This is in contrast to the uneven distribution of blood, which tends to have locally increased concentrations, for example in the lips and where spots are present.

3 Discussion

Preliminary results suggest that the technique described in this paper could enable parametric maps to be produced independently of curvature in an imaged scene. With an invariance to illumination intensity, it will not be necessary to accurately position the camera and illuminating light source before image acquisition. This will allow much wider application of the system developed by Cotton and Claridge [4,5].

Work is now underway to increase the number of histological parameters in the model to allow analysis of more complex skin lesions. This should enable the development of a system that can assist clinicians in the diagnosis of non-melanoma skin cancer, such as basal cell carcinoma that tends to occur on the face. It will also allow for the assessment of wounds where it is not possible to make contact with the imaged tissue, such as with diabetic foot ulcers.

It is envisaged that this methodology will be applicable to imaging other tissues. Two potential applications have so far been identified. These are imaging the ocular fundus [10,11] and the gastrointestinal tract. Success in both these applications requires a system which is able to recover histological data in a way which is invariant to surface geometry and illuminating light. Thus, the methodology presented in this article could prove key to their success.

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