

A new optimal wavelength for treatment of port wine stains?

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Abstract. In this study we investigate light penetration into skin in order to define the optimal wavelength for the treatment of port wine stains. A two-layer model (epidermis and dermis) is applied with a cylindric tube representing the ectatic blood vessel. Light propagation is calculated by the Monte Carlo method. Values for the optical properties of the skin were not only taken from the literature, but also derived from measurements of the spatially resolved reflectance on the human forearm applying a multilayer model. Using the new values, the maximal depth of selective vascular injury better fits experimental and clinical observations, compared to the values in literature. In addition, the optimal wavelength for treatment of port wine stains is shifted to longer wavelengths.

1. Introduction

Port wine stains are congenital vascular malformations consisting of ectatic blood vessels in the dermis which produce red skin. This vascular disorder often appears in the face causing severe psychical stresses to the patient. Based on the principle of selective photothermolysis (Anderson and Parrish 1981), the pulsed dye laser at 577 nm or 585 nm is successfully used to treat these lesions. The selection of laser parameters should allow selective damage to the vessels while sparing the neighbouring tissue.

The optimal laser parameters have been investigated in several papers. There is no doubt that the laser diameter should be large, for example 4 mm, to maximize penetration depth (Tan *et al* 1988, Kienle 1994). The exposure time of the laser pulse should be smaller than the thermal relaxation time of the blood vessel (Anderson and Parrish 1981), but long enough to prevent reversible damage (Hulsbergen Henning *et al* 1984) and microvascular haemorrhage (Garden *et al* 1986). The third parameter, the wavelength of the laser light, was suggested to be optimal at wavelengths where the absorption of oxyhaemoglobin has its maxima, for example at $\lambda = 415$ nm and $\lambda = 577$ nm (van Gemert *et al* 1982). The maximum at $\lambda = 577$ nm was chosen because of larger penetration depth compared to 415 nm, caused, for example, from less absorption of the laser light by melanin (van Gemert *et al* 1986).

However, experiments (Tan *et al* 1989) and theoretical modelling (Verkruysse *et al* 1993) showed that light of 585 nm penetrates deeper into tissue and hence is able to bleach port wine stains more effectively than that of 577 nm. The reason for this is that absorption of the dermis is smaller at 585 nm compared to 577 nm due to the smaller value of the absorption coefficient of blood in the dermis at 585 nm. Verkruysse *et al* (1993) showed this effect by calculating the maximum depth of vascular injury versus wavelength for different

amounts of blood in the dermis and for different vessel diameters. For selective damage of the enlarged vessel they used the criterion that the volumetric heat production due to absorbed photons has to be smaller at the epidermal-dermal junction than in the vessel. Since the precise processes which lead to the damage of the ectatic blood vessels are not known, they calculated the volumetric heat production in the vessel both at its top and in the middle. These calculations were based on Monte Carlo simulations of the light distribution within the tissue without a vessel and subsequent consideration of blood absorption in the vessel according to Beer's law. Recently, van Gemert *et al* (1995) published an investigation of the optimal wavelengths for treatment of port wine stains and telangiectasia using an analytical model. In their calculations the volumetric heat production at the top of the vessel was computed.

In this paper we exactly calculate the influence of the vessel represented by a blood cylinder in the dermis with the Monte Carlo method. The average amount of absorption in the vessel versus the absorption in the epidermis is taken as a new criterion to calculate selective damage of the vessel. Because of the questionable values of the optical properties of skin in literature, the absorption and reduced scattering coefficient of skin on a human forearm are derived from spatially resolved reflectance measurements. These parameters can be obtained using a multilayered model of the involved tissues and applying a nonlinear regression using Monte Carlo simulations for solving the forward problem (Kienle *et al* 1994). Both for the optical coefficients of skin given in the literature and for the ones measured in this study, calculations are executed to determine the optimal wavelength for the treatment of port wine stains lesions.

2. Model for light penetration in port wine stains

In order to determine the light penetration and hence the light absorption in tissue, the transport equation (Ishimaru 1978) has to be solved. Exact solutions (within statistical errors) can be found by Monte Carlo simulation (Wilson and Adam 1983, Wang and Jacques 1992), which is also applied in this study, because of its flexibility to handle complex geometries.

Figure 1 shows the geometry of the simulations. A laser beam with a diameter of 4 mm and an uniform radial distribution is incident perpendicularly onto the tissue which is considered to consist of one $d = 0.06$ mm and one infinitely thick layer, representing the epidermis and the dermis, respectively. The vessel is simulated as a cylinder with a diameter D placed in the dermis at depth a below the tissue surface. The layers are assumed to have infinite extensions in x and y directions, whereas the vessel is considered as infinite in the y direction. The probability of absorption per cubic millimetre in the centre of the beam for an incident photon, Δq , is calculated as a function of the tissue depth. This quantity is proportional to the temperature increase ΔT , if heat conduction can be neglected,

$$\Delta T = E\Delta q/\rho_d C \quad (1)$$

where E is the total incident pulse energy, ρ_d the density and C the specific heat of the tissue. This approximation is valid for laser pulses shorter than the thermal relaxation time (van Gemert *et al* 1982), which are used for treatment of port wine stains with the pulsed dye laser. To accelerate the simulations, the absorbed photons with y values between -1.6 mm and 1.6 mm are summed up (see figure 1). (The origin of the coordinate system is located in the centre of the incident beam at the tissue boundary.) It is carefully checked that $\Delta q(z)$ does not change noticeably within this range of y values.

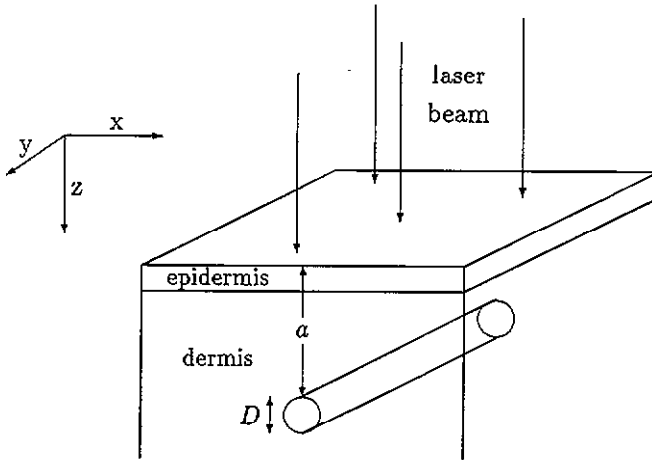


Figure 1. The geometry of the Monte Carlo simulations. The epidermis is $d = 0.06$ mm thick, whereas the dermis containing a homogeneous blood distribution is assumed to be infinitely thick.

The refraction index is set to 1.37 for all simulations and tissue types (Verkruyse *et al* 1993). Because only one vessel is explicitly regarded, the other vessels in the dermis were taken into account through a homogeneous distribution of blood in the whole dermis, thus considering also the vessels which are deeper in the dermis than the target vessel. This approximation using a homogeneous blood distribution can be applied, because the beam diameter and the average total photon path length is much greater than the distances between the vessels in the dermis. Therefore, the absorption coefficient of the dermis containing blood is changed according to the volumetric percentage of blood p in the dermis (Verkruyse *et al* 1993):

$$\mu_a = (p\mu_a^{blood} + (1 - p)\mu_a^{dermis})/100. \quad (2)$$

Vessels in dermis with $p = 1\%$, 5% and 10% are investigated. Oxygenated blood is assumed in the dermis and in the vessel. The scattering coefficient and the anisotropy factor of dermis, on the other hand, are not changed, because the reduced scattering coefficients μ'_s (which are the important scattering parameters for light propagation) of blood and dermis are of the same order of magnitude. (This is especially true, if one takes into account that the anisotropy factor of blood was measured by Steinke and Shepard (1988) as $g = 0.985$; compare table 1.)

To determine the maximal depth of vessel injury, the average Δq in the vessel, Δq_v , is compared with the average Δq in the epidermis, Δq_d . In the literature Δq_v is calculated either at the top or at the middle of the vessel (Verkruyse *et al* 1993). This criterion has a great influence on the determined optimal wavelength and maximal depth of vessel injury. It is not clear whether it is sufficient to coagulate one part of the vessel and damage a part of the vessel wall due to heat conduction or whether it is necessary to coagulate the whole vessel to damage the vessel irreversibly. However, there exist some indications for the latter assumption (Tan *et al* 1990, Kimel *et al* 1993). In this paper we use for Δq_v the average value of Δq of the whole vessel. This criterion is in a way a compromise of the ones above. However, the other criteria are also discussed.

The quantity $\Delta q(z)$ is calculated for different depths and diameters of the vessel at five wavelengths (577 nm, 585 nm, 590 nm, 595 nm and 600 nm). For this purpose the optical

Table 1. Optical coefficients for the Monte Carlo simulations (Verkruysse *et al* 1993, van Gemert *et al* 1992). In the case of blood oxygenated haemoglobin is assumed. The values of epidermis and dermis are used in subsection 4.1, whereas the values of blood are applied in subsections 4.1 and 4.2.

λ (nm)	Optical parameter	Epidermis	Dermis	Blood
577	μ_a (mm ⁻¹)	1.9	0.22	35.4
	μ_s (mm ⁻¹)	48	21.0	46.8
	g	0.787	0.787	0.995
585	μ_a (mm ⁻¹)	1.9	0.22	19.1
	μ_s (mm ⁻¹)	47	20.5	46.7
	g	0.790	0.790	0.995
590	μ_a (mm ⁻¹)	1.9	0.22	6.9
	μ_s (mm ⁻¹)	46	20	46.6
	g	0.800	0.800	0.995
595	μ_a (mm ⁻¹)	1.9	0.22	4.3
	μ_s (mm ⁻¹)	46	20	46.5
	g	0.800	0.800	0.995
600	μ_a (mm ⁻¹)	1.9	0.22	2.5
	μ_s (mm ⁻¹)	46	20	46.4
	g	0.800	0.800	0.995

properties of the involved tissues are needed (see section 3). Comparing the ratio $\Delta q_v/\Delta q_d$ as a function of the wavelength for different depths and diameters of the vessel, the optimal wavelengths can be determined. This is done for the optical parameters of skin from the literature (subsection 4.1) and for the coefficients derived in this study (subsection 4.2).

3. Optical properties of skin

3.1. Optical parameters from the literature

Several authors have investigated the optical properties of skin. Mostly, integrating spheres were used to measure the total transmittance and reflectance. From this the reduced scattering coefficient μ'_s and the absorption coefficient μ_a can be deduced applying a mathematical model. For example, van Gemert *et al* (1989) used data from Anderson and Parrish (1982) to calculate μ_a and μ'_s for dermis from Kubelka–Munk coefficients. For wavelengths about 600 nm they obtained for the reduced scattering coefficient $\mu'_s \approx 4 \text{ mm}^{-1}$ and for the absorption coefficient $\mu_a \approx 0.4 \text{ mm}^{-1}$. Jacques *et al* (1987) also used integrating sphere measurements and the diffusion approximation. They obtained $\mu'_s = 3.6 \text{ mm}^{-1}$ and $\mu_a = 0.27 \text{ mm}^{-1}$ at $\lambda = 633 \text{ nm}$. If light penetration in the dermis is calculated for these parameters, one finds a much lower penetration depth compared to measurements (Graaff *et al* 1993). The reasons for this discrepancy might be (i) the errors which result from using approximative theories and not an exact solution of the transport theory, (ii) the disregard of the photons which leave the sample laterally when measuring the total reflectance and transmittance and (iii) the faults induced from measuring on *ex vivo* tissue (Graaff *et al* 1993).

Although the optical parameters of skin in the literature are questionable, we use them in subsection 4.1 to calculate $\Delta q(z)$, because thereby the potential differences between our model, which regards the vessel explicitly, and the model of Verkruysse *et al* can be analysed.

The optical coefficients for epidermis, bloodless dermis and blood used in subsection 4.1 are from Verkruyse *et al* (1993) and van Germert *et al* (1992). They are summarized in table 1.

3.2. Optical parameters of the skin derived with spatially resolved reflectance

In this section we investigate the spatially resolved reflectance on a human forearm *in vivo* and use exact solutions of the transport equation to obtain the optical properties of skin. The experimental set-up for these measurements is described in detail elsewhere (Kienle *et al* 1994). In brief, a small laser beam from an HeNe laser with a wavelength of 633 nm is incident onto the tissue which is assumed to be infinitely thick. In the turbid medium the photons are scattered and re-emitted or absorbed. The re-emitted photons are imaged, spatially resolved, via an objective onto a CCD camera. From these intensity values the spatially resolved reflectance $R(\rho)$ is calculated, which describes the dependence of the re-emitted intensity on the radial distance ρ of the re-emitted photon to the incident beam. For a homogeneous medium it is possible to deduce from $R(\rho)$ the absorption and reduced scattering coefficient (Farrell *et al* 1992, Kienle 1994).

For these measurements the human forearm cannot be regarded as homogeneous (Kienle *et al* 1994). Therefore, a multilayer model is applied. The thickness of the epidermis and dermis is assumed to be 0.06 mm and 1 mm, respectively. Using a high-frequency ultrasound apparatus the distance from the tissue surface to the muscle was determined as 2 mm. Thus, the subcutaneous fat layer thickness is 1 mm. The muscle layer is assumed to be infinitely thick. To deduce the optical properties of skin with this multilayer model, μ_a and μ'_s of the muscle and fat layer are used from *ex vivo* measurements (Kienle *et al* 1994; see table 2).

Table 2. Parameters of the multilayer model. Data marked with an asterisk are results of the nonlinear regression to measure spatially resolved reflectance.

Layer	d (mm)	μ_a (mm ⁻¹)	μ'_s (mm ⁻¹)
Epidermis	0.06	0.8	1.75*
Dermis	1.0	0.015*	1.75*
Fat	1.0	0.0026	1.20
Muscle	∞	0.096	0.53

The absorption coefficient of the epidermis was measured by Marchesini *et al* (1992) as $\mu_a = 0.4$ mm⁻¹, whereas Verkruyse *et al* (1993) used $\mu_a = 1.9$ mm⁻¹. Considering the logarithmical average of these values, we assume for the absorption coefficient $\mu_a = 0.8$ mm⁻¹, and the anisotropy factor g is chosen to be 0.9 for all layers (Cheong *et al* 1990). The remaining parameters, the reduced scattering coefficients of the epidermis and the dermis, which are assumed to be equal, and the absorption coefficient of the dermis, are deduced from the measurement of the spatially resolved reflectance using a nonlinear regression and Monte Carlo simulations to solve the forward problem.

The measurement and the theoretical result from the nonlinear regression can be seen in figure 2. For the nonlinear regression the distance range 1 mm < ρ < 10 mm is used, because the data for $\rho < 1$ mm are influenced by an unknown amount of specular reflectance (Kienle 1994). Within this distance range the experimental data can be well described by the nonlinear regression. The absorption coefficient of the dermis is deduced as $\mu_a = 0.015$ mm⁻¹ and the reduced scattering coefficient of the dermis and the epidermis as $\mu'_s = 1.75$ mm⁻¹.

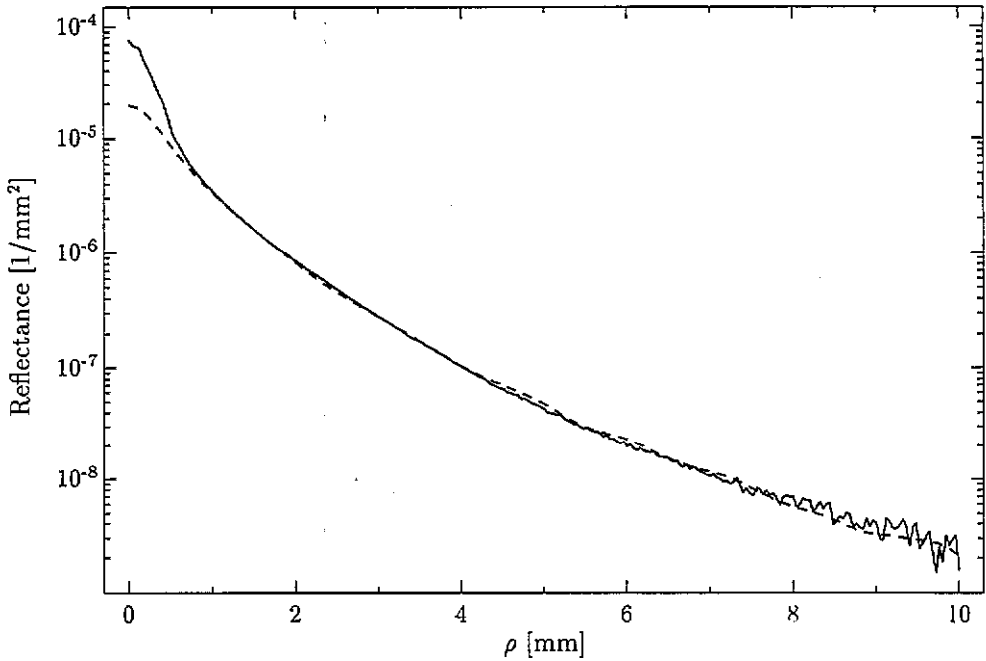


Figure 2. The measurement (solid) of the spatially resolved reflectance on a human forearm and the theoretical result (dashed) from a nonlinear regression.

Although the measurement of the optical properties of skin in this study is performed at 633 nm, the derived data can be used for all wavelengths between 577 nm and 600 nm. Applying Mie theory and distributions of scatterers with sizes similar to those expected in tissue, one finds that the reduced scattering coefficient only slightly decreases with increasing λ in the wavelength region used above (Kienle 1994). Regarding the alteration of the values for dermis and epidermis from the literature in table 1, this decrease can be neglected. Therefore, the approximation using the same value of μ'_s for all considered wavelengths should be justifiable. The absorption coefficient of dermis at wavelengths between 577 nm and 600 nm depends mainly on the absorption of blood. The absorption coefficient of blood is much smaller at 633 nm than at the considered wavelengths (about 70 times compared to 577 nm). Therefore, the derived values for μ_a at 633 nm are used as μ_a for bloodless dermis, and the absorption coefficient of dermis containing blood is calculated with (2) (see table 1). For the anisotropy factor a value of 0.9 is assumed. A change in the anisotropy factor providing that μ'_s is constant does not influence the light propagation significantly, if $0.8 < g < 0.95$ (Kienle 1994).

In the multilayer model it was assumed that μ'_s is the same for epidermis and dermis. This assumption is justified, because firstly μ'_s does not vary drastically between most tissue types at the same wavelength and secondly the epidermis is much thinner than the dermis. The latter implies that the scattering properties of the epidermis do not influence $R(\rho)$ or hence the derived scattering coefficient of the dermis to a great extent. For example, if μ'_s of epidermis is twice μ'_s of dermis, the assumption of equal μ'_s results in an error less than 10% for μ'_s of dermis with the method applied here.

The absorption coefficient of epidermis, on the other hand, shows great differences compared to μ_a of dermis and, in addition, μ_a is more important than μ'_s , if thermal processes are investigated. We assumed an absorption coefficient of the epidermis which

is in the range of those given in the literature. If this value is changed and a nonlinear regression with the experimental data from figure 2 is executed, the absorption coefficient of dermis also changes, but the reduced scattering coefficients of dermis and epidermis remain almost equal. Therefore, the use of other absorption coefficients of the epidermis does not change the general results of this study concerning the relationship of the different wavelengths. Particularly, this is true for a high volumetric blood content in the dermis, because there the contribution of the other chromophores to the absorption coefficient of dermis is low. However, the quantitative data of $\Delta q_v/\Delta q_d$ will change significantly, if μ_a of the epidermis is altered.

The optical coefficients of muscle used in the multilayer model (table 2) for the measurement on the human forearm are from *ex vivo* experiments. If the optical parameters of *in vivo* measurements on rabbit muscle are used (Kienle *et al* 1994), the experimental data, knowing that the muscle is 2 mm below the tissue surface, can only be fitted when there is no fat layer above the muscle. However, the optical parameters for skin derived from this model are close to those determined here (Kienle *et al* 1994).

Using the parameters of table 2, the total remission R_t can be calculated with the Monte Carlo method and the multilayer model. We obtain $R_t = 46\%$ for the values of skin derived in this study, whereas for literature data one obtains $R_t = 20\%$. Compared with measurements (Kienle 1994), the former result is closer to the experimental data.

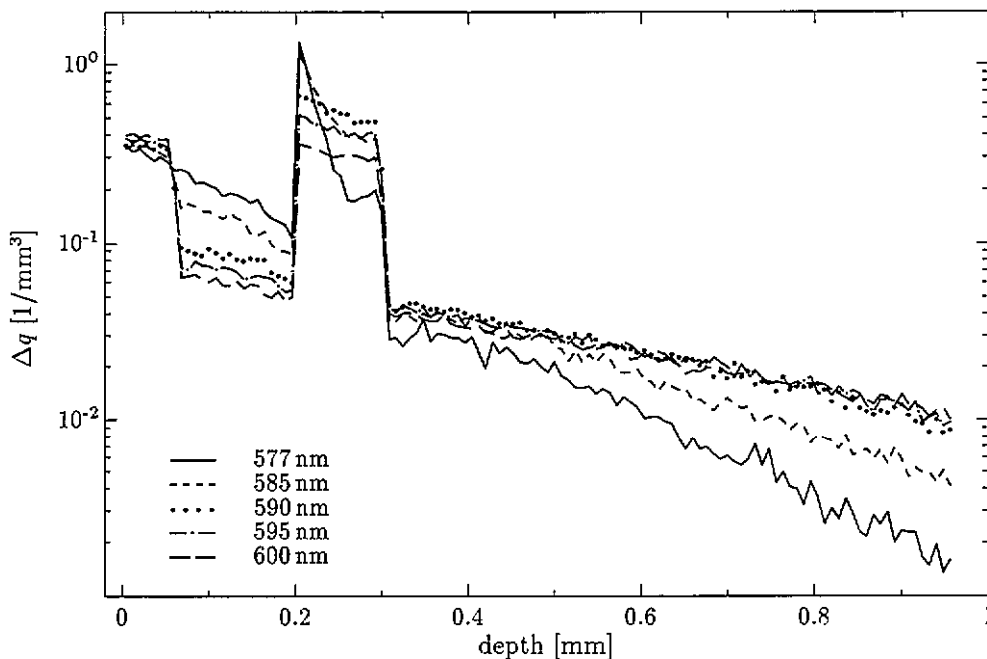


Figure 3. Δq as a function of the depth in the tissue for the wavelengths 577 nm, 585 nm, 590 nm, 595 nm and 600 nm and for a volumetric blood content of 5% in the dermis. The vessel has a diameter of 0.1 mm and is located between 0.2 mm and 0.3 mm below the tissue surface. The optical coefficients from table 1 are used.

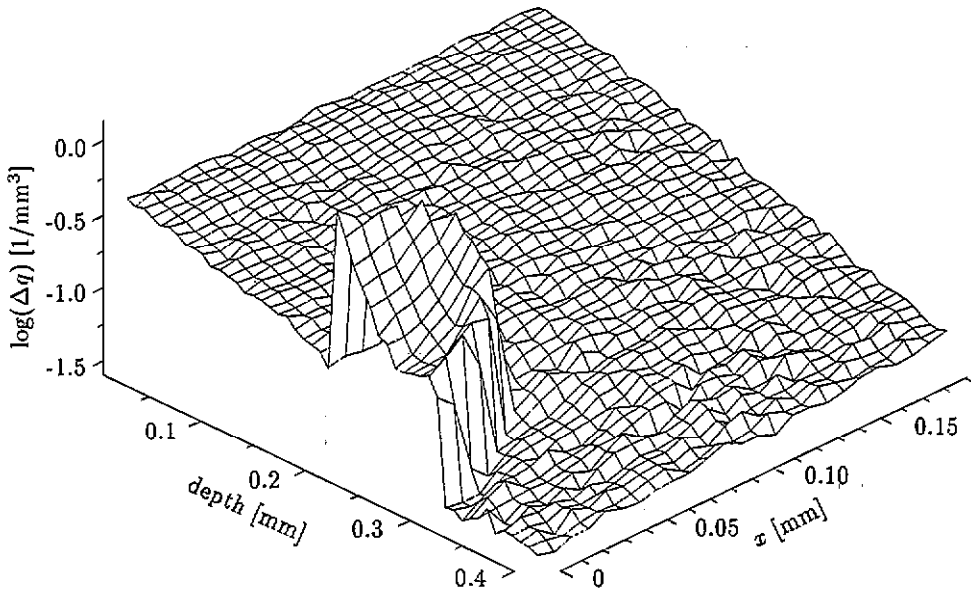


Figure 4. $\log(\Delta q)$ as a function of tissue depth and the distance in the x direction at 577 nm (see figure 1). The other parameters are the same as in figure 3.

4. Optimal wavelengths for the treatment of port wine stains

4.1. Calculations with optical parameters of skin from the literature

In this section the optical parameters given in table 1 are used. Figure 3 shows $\Delta q(z)$ for five different wavelengths and a volumetric blood content in the dermis of $p = 5\%$. The depth of the vessel is chosen as $a = 0.2$ mm and its diameter as $D = 0.1$ mm.

In figure 4 $\Delta q(x, z)$ can be seen in a three-dimensional plot for $\lambda = 577$ nm of the parameters in figure 4. Thus, data with $x = 0$ in figure 4 correspond to the curve for $\lambda = 577$ nm in figure 3.

In general it can be stated that for shorter wavelengths $\Delta q(z)$ in the dermis decreases faster with increasing depths than for longer wavelengths, because the absorption coefficient is greater and $\Delta q(z)$ is proportional to the fluence rate. The same is in principle true in the blood vessel. However, especially at $\lambda = 577$ nm, an increase of Δq at the bottom and at the sides of the vessel can be seen, which results from photons entering the vessel from the bottom or laterally. Thus, the smaller the absorption coefficient of the blood, the more uniform is the distribution of Δq in the vessel. For example, at 590 nm the high value of Δq and hence of the temperature existing at smaller wavelengths, which perhaps cause reversible damage, can be avoided.

As stated above, the criterion for calculating the vessel damage is important. If Δq at the top of the vessel is compared with Δq_d in figure 3, $\lambda = 577$ nm is the optimal wavelength for selective photothermolysis, whereas at the middle of the vessel $\lambda = 590$ nm has the highest value. Using the criterion of the average amount of Δq in the vessel compared to the highest value in the epidermis, $\lambda = 585$ nm is the best wavelength (see figure 7). Thus, the last criterion is a compromise of the other two and is used in the following to compare the simulations.

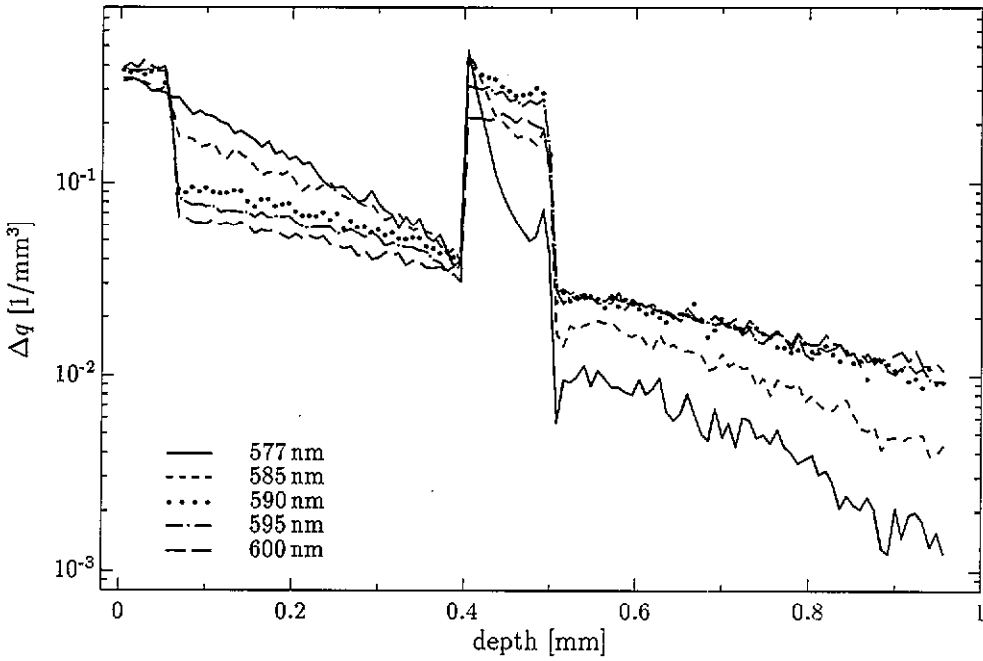


Figure 5. $\Delta q(z)$ for a blood vessel at $a = 0.4$ mm. The other parameters are the same as in figure 3.

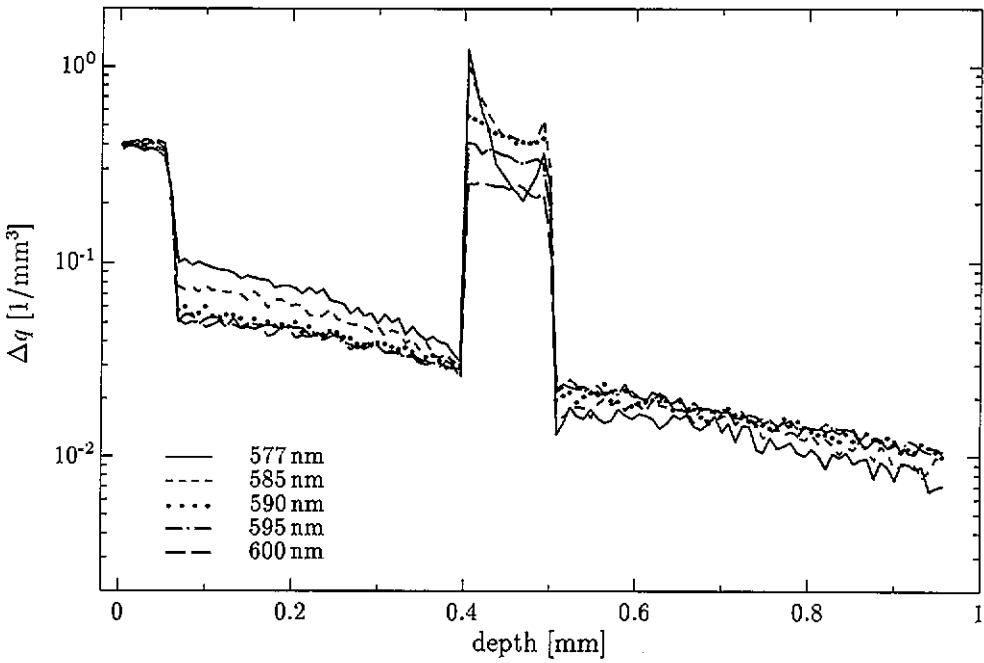


Figure 6. $\Delta q(z)$ for a blood vessel at $a = 0.4$ mm and a volumetric blood content in the dermis of 1%. The diameter of the blood vessel is $D = 0.1$ mm. The optical coefficients from table 1 are used.

In addition, the optimal wavelength for selective photothermolysis depends on the depth of the vessel, on the blood content in the dermis and on the size of the vessel. Figures 5 and 6 show the results for a vessel at a depth $a = 0.4$ mm, calculated for a volumetric blood content of $p = 5\%$ and $p = 1\%$ in the dermis, respectively.

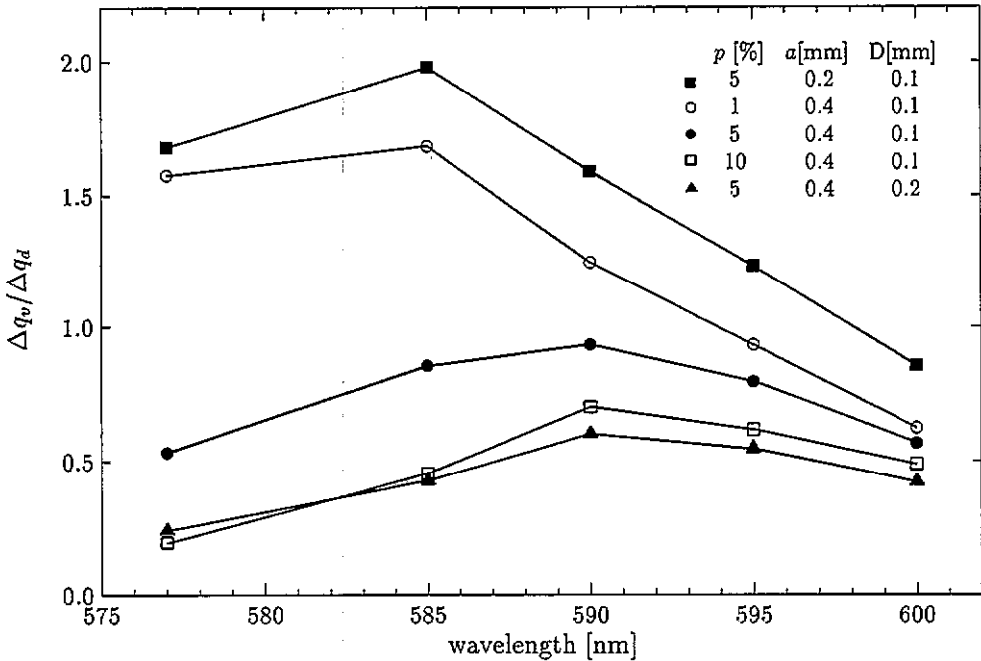


Figure 7. $\Delta q_v/\Delta q_d$ at the wavelengths 577 nm, 585 nm, 590 nm, 595 nm and 600 nm for different vessel diameters, depths and volumetric blood contents in the dermis. The optical coefficients from table 1 are used.

Comparing figure 3 with figure 5 it can be seen that with larger depths of the vessel not only does the absorbed energy in the vessel decrease, but also there is a shift to longer wavelengths regarding the optimal wavelength for photothermolysis. The main reason for this shift is the different absorption of the dermis. Regarding $\Delta q_v/\Delta q_d$ (see figure 7) the optimal wavelength is about $\lambda = 590$ nm.

If figure 6 is compared with figure 5, it is obvious that the best wavelength for deep photothermolysis also depends on the amount of blood in the dermis. Changing p from 5% to 1% results in a shift of the maximum in figure 7 from $\lambda = 590$ nm to $\lambda = 585$ nm. For $p = 10\%$ and $a = 0.4$ mm, figure 7 shows that the optimal wavelength is about $\lambda = 590$ nm.

Figure 7 also indicates that the diameter of the vessel is important, unless only heating of the top of the vessel is required for damage. In this case, there is only a small influence of the vessel size due to the effect of the vessel on the light propagation in the dermis. Comparing a vessel with $D = 0.1$ mm to one with $D = 0.2$ mm for $a = 0.4$ mm and $p = 5\%$, a shift to longer wavelengths can be observed, indicated by the relative values of $\Delta q_v/\Delta q_d$ at 585 nm and 595 nm in figure 7. This is once again caused by the different absorption of blood at the considered wavelengths. The greater μ_a , the greater the decrease of Δq in the vessel, which results for vessels with larger diameter in a smaller value of Δq_v .

As mentioned above, $\Delta q_v/\Delta q_d$ should at least be greater than unity for selective photothermolysis of the vessel to be possible. Figure 7 shows that this is, for example,

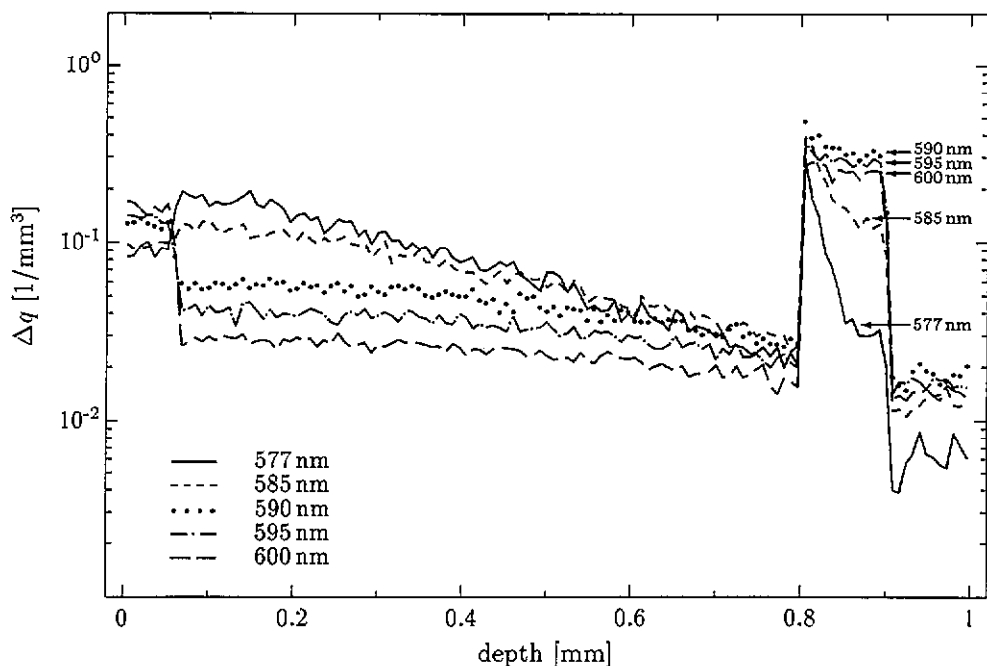


Figure 8. $\Delta q(z)$ for a blood vessel at $a = 0.8$ mm and a volumetric blood content in the dermis of 5%. The diameter of the blood vessel is $D = 0.1$ mm. The optical coefficients of dermis and epidermis are from table 2 and those of blood are from table 1.

not the case for a vessel with $p = 5\%$, $D = 0.1$ mm and $a = 0.4$ mm. However, these vessel parameters are typical for port wine stains (Barsky *et al* 1980). Therefore, these results are in contradiction to the successful treatment of port wine stains.

4.2. Calculations with new optical parameters of skin

In this section we use the optical coefficients for dermis and epidermis derived in subsection 3.2 (see table 2) and the values for blood from table 1. Figures 8 and 9 give $\Delta q(z)$ for a vessel with $D = 0.1$ mm, $a = 0.8$ mm and a volumetric blood content in the dermis of 5% and 1%, respectively.

Regarding these figures, it is obvious that with the new optical coefficients of skin, even for a relatively deep vessel, $\Delta q_v/\Delta q_d$ exceeds unity (see also figure 10). In general the features concerning the wavelength dependences of the curves in figures 8 and 9 are similar to those in figures 5 and 6, respectively. However, there is a slight shift towards longer wavelengths for the optimal wavelengths.

Figure 10 gives also data for a vessel with $D = 0.1$ mm, $a = 0.8$ mm and $p = 10\%$, and a vessel with $D = 0.2$ mm, $a = 0.8$ mm and $p = 5\%$. For both, the maximum value of $\Delta q_v/\Delta q_d$ is between $\lambda = 590$ nm and $\lambda = 595$ nm.

In figures 11 and 12 $\Delta q_v/\Delta q_d$ can be seen for vessels with a constant diameter of $D = 0.1$ mm at different depths, $a = 0.4$ mm, $a = 0.8$ mm, $a = 1.2$ mm and $a = 1.6$ mm. The volumetric blood content in the dermis is $p = 5\%$ and $p = 1\%$, respectively.

Figure 11 shows how the maximum of $\Delta q_v/\Delta q_d$ increases from about $\lambda = 585$ nm for $a = 0.4$ mm to about $\lambda = 595$ nm for $a = 1.6$ mm. Figure 11 also predicts that vessels up to a depth of $a = 1.2$ mm can be successfully treated. For $p = 1\%$ the optimal wavelength

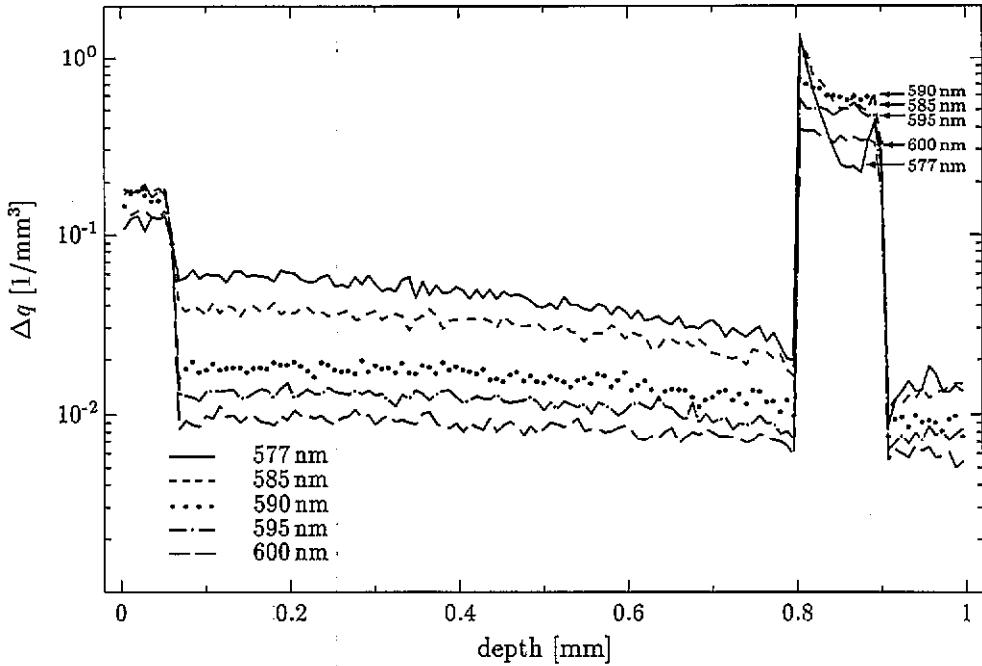


Figure 9. $\Delta q(z)$ for a blood vessel at $a = 0.8$ mm and a volumetric blood content in the dermis of 1%. The diameter of the blood vessel is $D = 0.1$ mm. The optical coefficients of dermis and epidermis are from table 2 and those of blood are from table 1.

is shifted from values between $\lambda = 577$ nm and $\lambda = 585$ nm for $a = 0.4$ mm to values between $\lambda = 585$ nm and $\lambda = 590$ nm for $a = 1.6$ mm.

5. Discussion

In subsection 4.1 $\Delta q(z)$ is calculated for a port wine stain model with optical skin data from the literature. Comparing the results with those of Verkruysse *et al* (1993), two differences can be found, although the same optical coefficients are used. First the maximal depth for vascular injury is slightly smaller in the present study and second there is also a slight shift towards longer wavelengths for the optimal treatment parameter. This can be explained with the help of figure 13, which compares the curve in figure 5 for $\lambda = 590$ nm (solid) with data calculated according to the model of Verkruysse *et al* (1993) (dashed), where a Beer law for the fluence rate in the vessel is assumed.

Calculating the vessel explicitly in the Monte Carlo simulations results in a decrease of the fluence rate near the vessel, because of the photon sink caused from the blood. Consequently, Δq_v and the maximal depth of injury decrease compared to the case where the vessel is not integrated in the Monte Carlo simulations (dashed).

The second difference can be explained with figure 14, which gives the same calculations as in figure 13, but at $\lambda = 577$ nm. At $\lambda = 577$ nm the relative difference between the two curves in the middle of the vessel is greater than at $\lambda = 590$ nm. Thus, the optimal wavelength is shifted to longer wavelengths when the vessel is explicitly considered. The reason is probably that more light penetrates from the side or from the bottom into the middle of the vessel at $\lambda = 590$ nm compared to $\lambda = 577$ nm and, therefore, the difference of Δq in the vessel of the solid curve and the dashed curve is smaller at $\lambda = 590$ nm

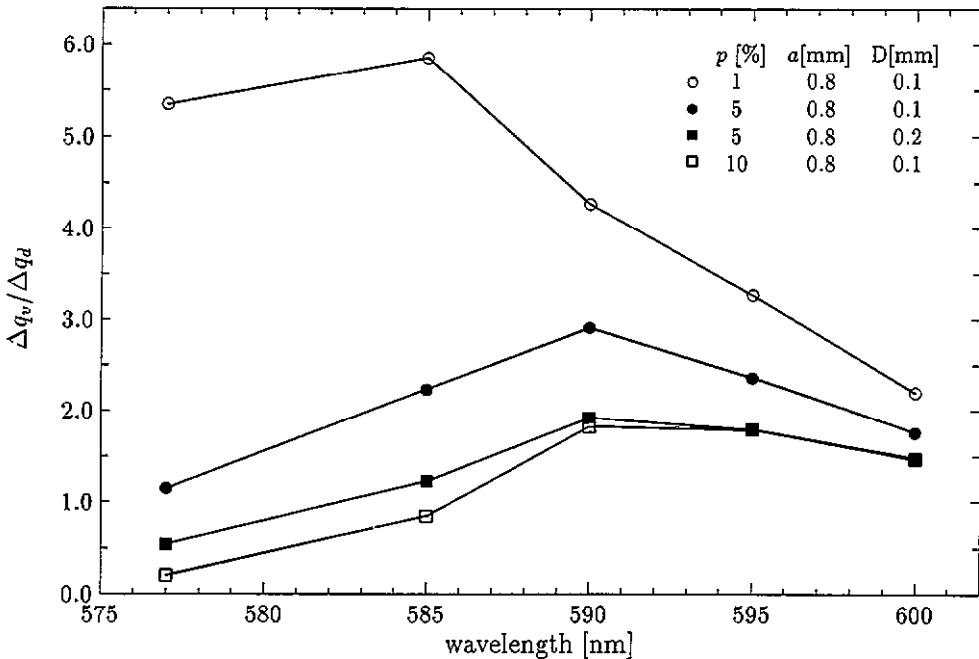


Figure 10. $\Delta q_v/\Delta q_d$ as a function of wavelength for different parameters with $a = 0.8$ mm. The optical coefficients of dermis and epidermis are from table 2 and those of blood are from table 1.

compared to $\lambda = 577$ nm.

The optical coefficients of skin derived in this study (subsection 3.2) reflect a relative large penetration depth in the tissue. Tan *et al* (1990) measured the maximal depth of damaged vessels in port wine stain skin using a radiant exposure of $6\text{--}7 \text{ J cm}^{-2}$ as 0.72 mm at $\lambda = 577$ nm and 1.16 mm at $\lambda = 585$ nm. Assuming a volumetric blood content of 5%, the values of Δq_v at the two wavelengths can be compared. For $\lambda = 577$ nm and $a = 0.8$ mm we have $\Delta q_v = 0.1 \text{ mm}^{-3}$ (see figure 8). The same Δq_v value at $\lambda = 585$ nm is obtained at a depth of approximately $a = 1.1$ mm (not shown). Thus, this corresponds to the experimental data quoted above.

Tan *et al* (1989) also histologically studied the depth of laser-altered blood vessels in albino pig skin after applying laser pulses with different radiant exposures. Comparing their results for $\lambda = 585$ nm and $\lambda = 590$ nm, it can be summarized that the depth of vascular damage at both wavelengths is equal to 1.8 mm at a radiant exposure of 10 J cm^{-2} . At smaller radiant exposures there is a decrease in damage depth, which is more pronounced at $\lambda = 590$ nm than at $\lambda = 585$ nm. This behaviour is similar to our theoretical data (see figure 12) where the volumetric blood content is 1%, which should be close to the value of normal skin. There, for $a = 1.6$ mm $\Delta q_v/\Delta q_d$ is similar at $\lambda = 585$ nm and $\lambda = 590$ nm and for smaller a $\Delta q_v/\Delta q_d$ is greater at $\lambda = 585$ nm compared to $\lambda = 590$ nm. This means that the vascular damage depth is larger at $\lambda = 585$ nm than at $\lambda = 590$ nm for this depth range. Two remarks have to be made at this point. First, in these examples we should use Δq_v rather than $\Delta q_v/\Delta q_d$ for comparison, because Δq_v corresponds to the amount of damage. However, Δq_d does not vary drastically with wavelength. Thus, $\Delta q_v/\Delta q_d$ can be used for this qualitative comparison. Second, the vessel diameters in normal pig skin are smaller than $D = 0.1$ mm, which is used in figure 12. This should result in a deeper depth

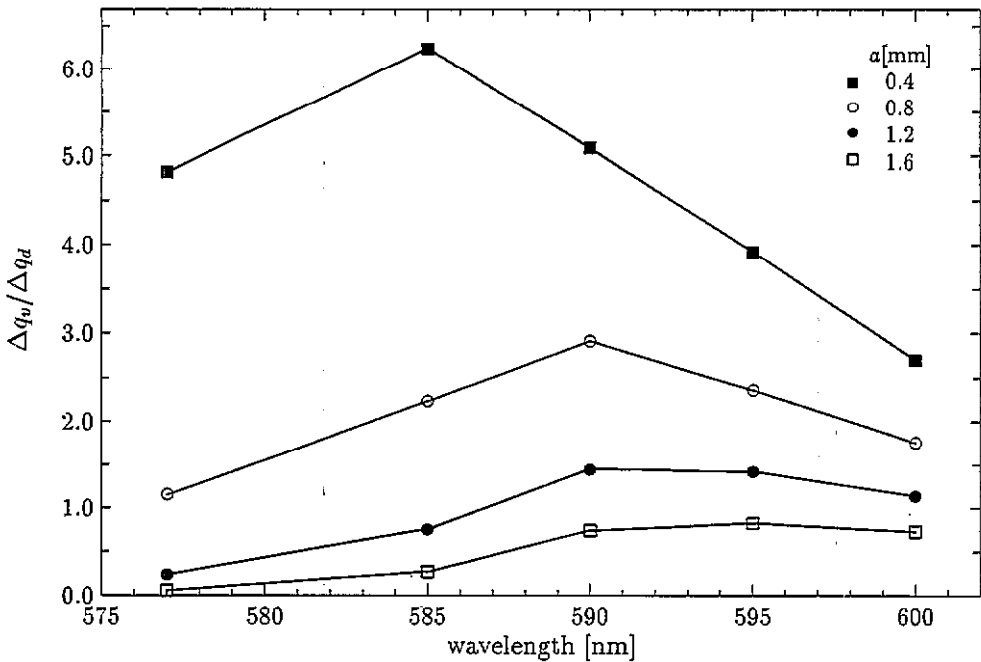


Figure 11. $\Delta q_v/\Delta q_d$ as a function of the wavelength for different depths with $p = 5\%$ and $D = 0.1$ mm. The optical coefficients of dermis and epidermis are from table 2 and those of blood are from table 1.

of vessel damage at $\lambda = 585$ nm compared to $\lambda = 590$ nm.

Tan *et al* (1989) also reported that in albino pig skin histologic changes after laser pulses at $\lambda = 590$ nm did not remain localized to the blood vessels and that preliminary studies in human skin exposed to 590 nm irradiation, in contrast to 585 nm, showed scar formations. Whereas the reason for the former observation is unclear, the latter could be due to the higher amount of Δq in the epidermis at $\lambda = 590$ nm compared to $\lambda = 585$ nm, if the same irradiance is applied at both wavelengths. In figure 8 this difference and consequently the difference of the initial temperature increase, if a short-pulsed laser is used, are about 20%. This higher Δq at $\lambda = 590$ nm is caused from the smaller absorption coefficient in the dermis at this wavelength compared to $\lambda = 585$ nm, because this results in a higher fluence rate at $\lambda = 590$ nm in the epidermis compared to $\lambda = 585$ nm.

With (1) and the results for $\Delta q(z)$, the temperature increase ΔT in the vessel can be calculated. With $\Delta q_v = 0.22$ mm⁻³ for $\lambda = 585$ nm in figure 8, we obtain $\Delta T \approx 50$ K, when an incident energy density of 6.5 J cm⁻² and $\rho_d C = 3.5$ J cm⁻³ K⁻¹ (van Gemert *et al* 1986) is assumed. This temperature should be sufficient for vascular damage due to coagulation (Anderson and Parrish 1981). The depth of the vessel, $a = 0.8$ mm, is somewhat lower than the measured damage depth, $a = 1.16$ mm (Tan *et al* 1990), mentioned above. However, figure 8 is calculated for a volumetric blood content of 5%, which is greater than the average value measured in port wine stain skin (about 4%) (Barsky *et al* 1980) and therefore the penetration depth is smaller.

We note that the dependence of $\Delta q_v/\Delta q_d$ on wavelength in figures 7 and 10–12 would show a symmetry about 577 nm, if wavelengths smaller than 577 nm had been calculated, because the absorption coefficient of oxyhaemoglobin has a maximum at 577 nm. However, the wavelengths greater than 577 nm are more favourable for selective photothermolysis,

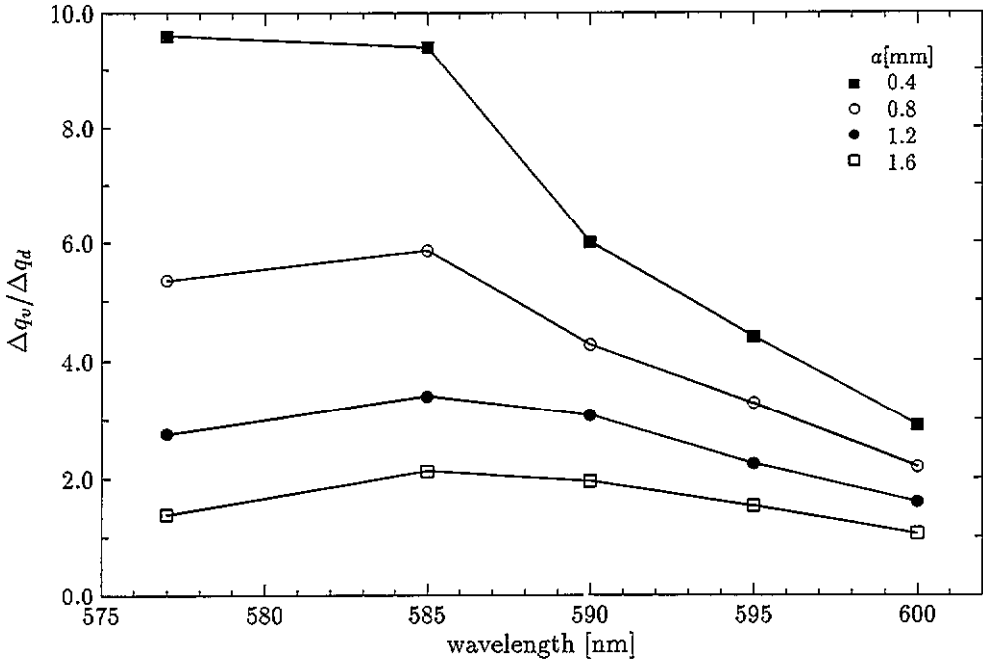


Figure 12. $\Delta q_v/\Delta q_d$ as a function of the wavelength for different depths with $p = 1\%$ and $D = 0.1$ mm. The optical coefficients of dermis and epidermis are from table 2 and those of blood are from table 1.

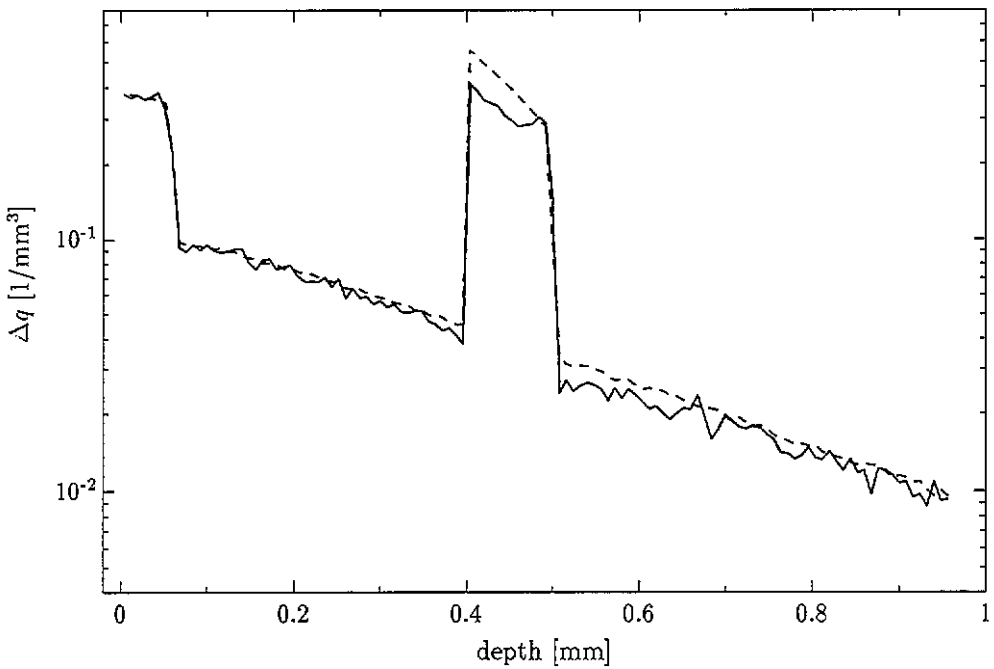


Figure 13. Δq at $\lambda = 590$ nm for the same parameters as in figure 5. The solid curve is calculated as in figure 5 and the dashed curve according to Verkruyse *et al* (1993).

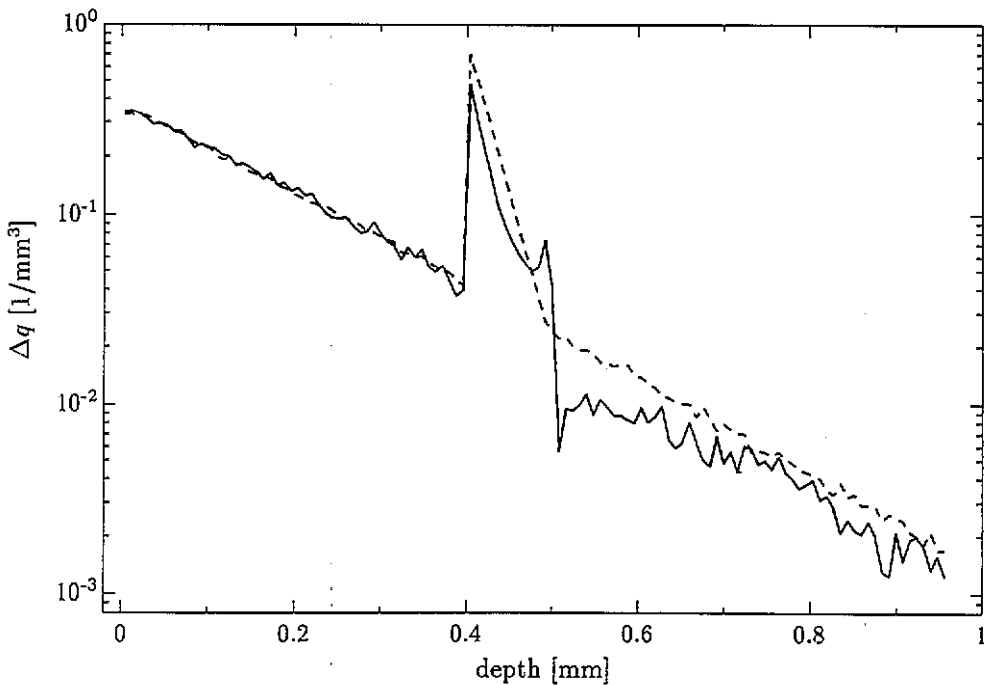


Figure 14. Δq at $\lambda = 577$ nm for the same parameters as in figure 5. The solid curve is calculated as in figure 5 and the dashed curve according to Verkruyse *et al* (1993).

because the scattering coefficient of skin and the absorption coefficient of melanin decrease with longer wavelengths.

6. Conclusion

In this study we have presented a model for calculating the light absorption in port wine stain skin, which exactly considers a cylindric tube representing a blood vessel. The surrounding vessels were taken into account through an average dermal blood concentration. It could be shown that this results in a slightly longer optimal wavelength for selective photothermolysis of the blood vessel compared to a simpler model (Verkruyse *et al* 1993).

We also derived the optical coefficients of the human skin *in vivo*, which were used to calculate the optimal wavelengths for different sizes and depths of the blood vessel and for different volumetric blood contents in the skin. It was discussed that the calculated optimal wavelength depends on the model for vascular damage. In this study, the average absorption in the vessel was compared to the highest absorption in the epidermis. This criterion gives optimal wavelengths which lie between those for the criteria used in literature. It could be shown that for many combinations of the parameters mentioned above wavelengths of about $\lambda = 590$ nm are predicted to have better treatment results than wavelengths of about $\lambda = 585$ nm, which is believed to be the optimal wavelength. This is especially true if the volumetric blood content in the dermis is high. This means, considering that normally several sessions are needed for the treatment of port wine stains, that $\lambda = 590$ nm should, at least, be more effective in the first treatment sessions.

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