THERMOGRAPHIC METHODS DURING LASER-TISSUE INTERACTION

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Abstract: This paper describes the use of thermographic methods during laser/tissue interaction. The efficacy of laser application on human tissue is partly dependent on the skills and experience of the laser operator. The laser effects of photothermolysis occurring in the dermis are difficult to predict. With the intervention of an infrared radiometer it is possible to see the thermal distribution at the skin surface as the manually held laser light strikes the skin. It also useful in improving understanding as to the light/tissue interaction occurring in the dermis as a result of a proportion of returning energy to the skin surface. The radiometer is also useful when using computer scanning in the setting-up process. The tissue optical and thermal processes are also investigated by a Monte Carlo Model which is validated by the thermographic measurements.

The results indicated the reaction of a patient's physiology to thermal stresses, reducing the possibility of secondary skin damage during therapy and optimising various laser parameters both empirically through the validation of a theoretical model that predicts the radiative and thermal transport through the tissue medium.

1 Introduction:

Infrared Thermographic Monitoring (ITM) has been successfully used in medicine for a number of years and much of this has been documented by Prof Francis Ring [1995], who has established a database and archive within the Department of Computing at the University of Glamorgan, UK, spanning over 30 years of ITM applications [www.medimaging.org]. Examples include monitoring abnormalities such as malignancies, inflammation, and infection that cause localized increases in skin temperature, which show as hot spots or as asymmetrical patterns in an infrared thermogram.

A recent medical example that has benefited by the intervention of ITM is the treatment by laser of certain dermatological disorders. Advancements in laser technology have resulted in new portable laser therapies such as the removal of vascular lesions (in particular Port Wine Stains) reduction of hair (depilation) and wrinkles.

In these laser applications it is a common requirement to deliver laser energy uniformly without overlapping of beam spot to a sub-dermal target region, such as a blood vessel, without damaging surrounding and surface tissue. The temperature rise at the skin surface and the threshold to burning/scarring is of critical importance for obvious reasons. This type of therapy has not yet benefited significantly from a thermographic evaluation. Some of the reasons for this include the high-cost of the technology, lack of understanding as to the thermographic potential, instability, particularly older IR cameras with reference to cryogenic cooling [Thomas, 1999]. However, in recent years most of these problems have been addressed resulting in a new breed of portable, totally safe, very accurate and stable thermal measuring instruments with high levels of sensitivity.

1.1 Medical Infrared Thermography:

The establishment, development and consequential success of Medical Infrared Thermographic (MIT) intervention is primarily based on the understanding of the following:

a) Problem/condition to be monitored
b) Set-up and correct operation of infrared system
c) Appropriate conditions during the monitoring process
d) Evaluation of activity and development of standards and protocol

With reference to a) the condition to be monitored, there needs to be a good knowledge as to the physiological aspects of the desired medical process; in laser therapy an understanding as to the mechanisms involved in laser-tissue interaction. A good reference source of current practice can be found in the Handbook of Optical Biomedical Diagnostics, published by The International Society for Optical Engineering (SPIE). This will help in the identification of the optimum infrared technology. In this application fast data-capture (at that time 50Hz), good image quality (256×256 pixels), temperature sensitivity and repeatability were considered important and an Inframetrics SC1000 Focal Plane Array Radiometer (3.4 - 5μm, CMOS PtSi Cooled Detector) with a real-time data acquisition system (Dynamite) was used. There are currently very fast systems available with data acquisition speeds in terms of 100's of Hertz with detectors that provide excellent image quality. In b) the critical aspect is training. These days infrared equipment manufacturers design systems with multiple applications in mind. This has resulted in many aspects of good practice and quality standards. This is one of the reasons why industrial infrared thermography is so successful. This has not necessarily been the case in medicine. However, it is worth noting that there are a number of good infrared training organizations throughout the world, particularly in the USA. The advantages of adopting training organisations such as these is that they have experience of training with reference to a very wide range and type of infrared thermographic systems, in a wide range of applications. In c) consideration as to the conditions surrounding the patient and the room environment are important for optimum results. In the UK for example, Prof Francis Ring, University of Gamorgan has led the way in the development and standardisations of clinical infrared practice. Finally d) the evaluation of such practice is crucial if lessons are to be learnt and protocol and standards are to emerge.

Lasers have been used in dermatology for some 40 years [Wheeland, 1995]. In recent years there have been a number of significant developments particularly regarding the improved treatment of various skin disorders most notably the removal of vascular lesions using dye lasers [Glassberg *et al*, 1989, Barlow *et al*, 1996, Garden et al, 1988, Motley *et* al, 1996, Lanigan, 1996, Kiernan, 1997] and depilation using ruby lasers [Grossman *et al,* 1997, Gault, 1998, Clement *et al*, 1999].

2 Laser-tissue interaction:

The mechanisms involved in the interaction between light and tissue depend on the characteristics of the impinging light and the targeted human tissue. To appreciate these mechanisms the optical properties of tissue must be known. It is necessary to determine the tissue reflectance, absorption and scattering properties as a function of wavelength. A simplified model of laser light interaction with the skin is illustrated**,** in Figure 2.1.

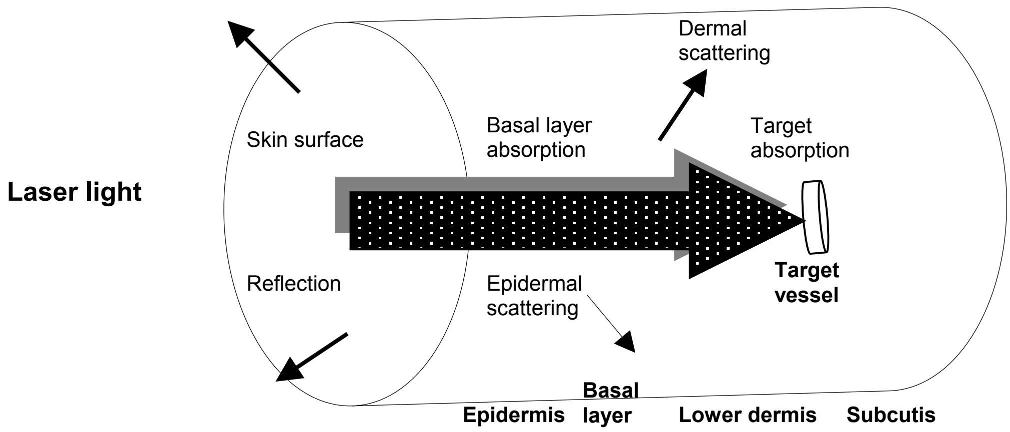

Figure 2.1 Passage of laser light within skin layers

This research study involves laser radiation penetrating the epidermis, basal and preferentially absorbed within the blood layers located in the lower dermis and subcutis.

There are three types of laser/tissue interaction namely photothermal, photochemical and ionisation, Table 2.1.

EFFECT	**INTERACTION**
PHOTOTHERMAL	
Photohyperthermia	Reversible damage of normal tissue (37-42˚C)
	Loosening of membranes (odema), tissue welding (45–60˚C)
Photothermolysis	Thermal-dynamic effects, micro-scale overheating
Photocoagulation	Coagulation, necrosis (60-100˚C)
Photocarbonisation	Drying out, vaporization of water, carbonization (100-300˚C)
Photovaporisation	Pyrolysis, vaporization of solid tissue matrix (>300˚C)
PHOTOCHEMICAL	
Photochemotherapy	Photodynamic therapy, black light therapy
Photoinduction	Biostimulation
PHOTOIONISATION	

Photoablation	Fast thermal explosion, optical breakdown, mechanical shockwave

Table 2.1 Interaction effects of laser light and tissue

Selective photothermolysis is the term that describes the process of a sub-dermal feature preferentially absorbing the laser energy, in order to achieve some localised therapeutic benefit. Selective photothermolysis is the specific absorption of laser light by a target tissue in order to eliminate that target without damaging surrounding tissue. In the treatment of Port Wine Stains (PWS), a dye laser of wavelength 585 nm has been widely used [Clement *et al*, 2000] where the profusion of small blood vessels that comprise the PWS are preferentially targeted at this wavelength.
The spectral absorption characteristics of light through human skin is well established [Anderson and Parrish, 1981].
The application of appropriate laser technology to medical problems depends on a number of laser operating parameters including matching the optimum laser wavelength for the desired treatment.
The first evaluation was designed to access whether or not measurements of the surface temperature of the skin were reproducible when illuminated by nominally identical laser pulses. In this case a 585nm dye laser and a 694nm ruby laser were used to place a number of pulses manually on tissue. The energy emitted by the laser was highly repeatable. Care was taken to ensure that laser and radiometer position was kept constant and that the anatomical location used for the test had uniform tissue pigmentation.

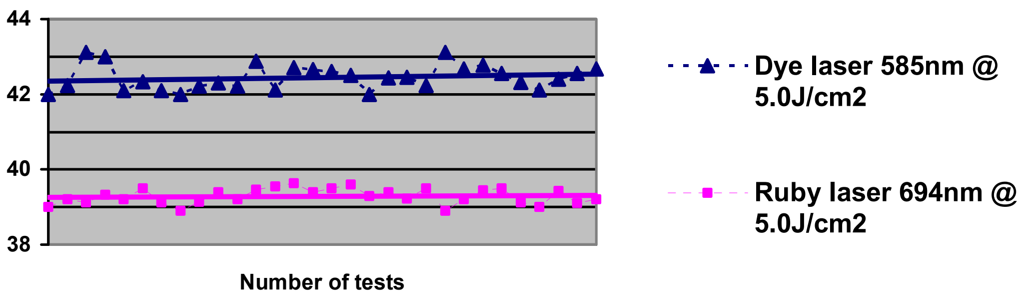

Figure 2.2 Repeatability of initial maximum temperature (Dye laser 585nm @ 5.2 J/cm^2, 5mm)

Figure 2.2 shows maximum temperature for each of twenty shots fired on the forearm of a Caucasian male with type 2 skin. This result is one of many recorded and shows typical results that are reproducible. Maximum temperature varies between 48.90°C and 48.10°C representing a variance of 1°C (± 0.45°C). This level of reproducibility is pleasing since it shows that, despite the complex scenario, the radiometer is c epeatedly and accurately measuring surface tissue temperatures. In practice the radiometer may b nform the operator when any accumulated temperature has subsided allowing further treatment w °C eeding some damage threshold.
E sity is an very important laser parameter and can be varied dependent on application. It is normal in the discipline to measure energy density (sometimes called fluence) in J/cm^2. In treating vascular lesions most utilize an energy density for therapy of 5-10 J/cm^2 [Garden & Barkus, 1996]. The laser operator needs to be sure that the energy density is uniform and does not contain hot-spots that may take the temperature above the damage threshold inadvertently. The infrared system can help fine tune the laser and reduce the possibility of excessive energy density.

The model used in this research to validate laser spot size and positioning, uses a trapezoidal profile therefore, power is not a constant. In the subject such a uniform profile has become known as ‘top-hat’. It can be observed that the temperature on the surface of the skin tracks the optical profile of the laser spot. It is also interesting to confirm that the temperature rise is confined to the illuminated region, for this the applicability of thermal imaging to the challenge of accurately setting and confirming the optical parameters of lasers when used in dermatology.

2.1 Thermographic Results of Laser Positioning:

During laser therapy the skin is treated with a number of spots, applied manually depending on the anatomical location and required treatment. This research has found that spot size, directly affects efficacy of treatment. The wider the spot size the higher the surface temperature [Thomas *et al*, 2002]. The type and severity of lesion also determines the treatment required. Its colour severity (dark to light) and its position on skin (raised to level). Therefore the necessary treatment may require a number of passes of the laser over the skin. It is therefore essential as part of the treatment that there is a physical separation between individual spots so that:

- The area is not over treated with overlapping spots that could otherwise result in local heating effects from adjacent spots resulting in skin damage.
- The area is not under treated leaving stippled skin.
- The skin has cooled sufficiently before second or subsequent passes of the laser.

Figure 2.3 shows two laser shots placed next to each other some 5mm apart. The time between the shots is 1 second. There are no excessive temperatures evident and no apparent temperature build-up in the gap. This result concurs with Lanigan [1996], advising a minimum physical separation of 5mm between all individual spot sizes.

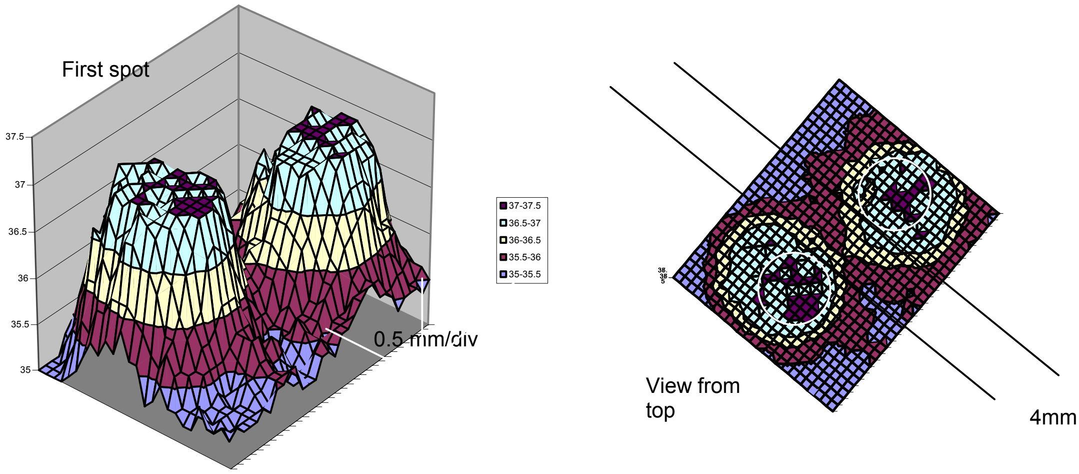

Figure 2.3 Two dye laser spots with a minimum of 4mm separation (585nm @ 4.5 J/cm^2, 5mm spot)

The intention is to optimise the situation leading to a uniform therapeutic/aesthetic result without either striping or thermal build up. This is achieved by firstly determining the skin colour (Chromotest) for optimum energy settings, followed by a patch test and subsequent treatment. Increasing the number of spots to 3 with the 4mm separation reveals a continuing trend, Figure 2.5. The gap between the first two shots is now beginning to merge in the 2s period that has lapsed. The gap between shots 2 and 3 remains clear and distinct and there are clearly visible thermal bands across the skin surface of between 38-39°C and 39-40°C. These experimental results supply valuable information to support the development of both free-hand treatment and computer-controlled techniques.

Second spot

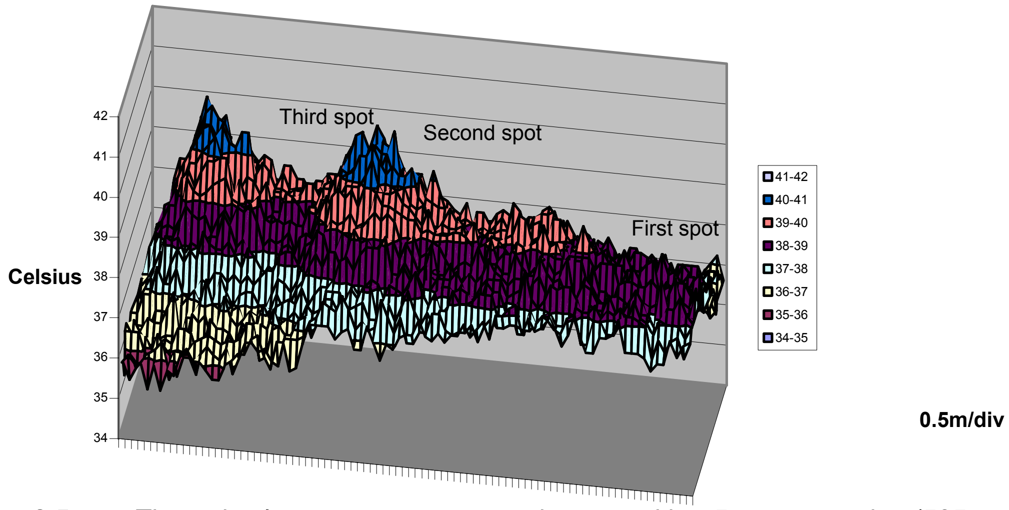

Figure 2.5 Three dye laser spots, two seconds apart with a 5mm separation (585nm at 5 J/cm^2, 5mm spot)

2.2 Laser Overlapping:

Overlapping should be prevented particularly when treating the face and neck. Overlapping of adjacent shots could result in regions of the lesion receiving up to twice the fluence. This could cause significant epidermal or dermal damage as the heat generated by the first shot may not have subsided before the onset of the second shot. Also the optical parameters of the overlap region would have been altered by the first shot prior to the application of the second shot, although the clinical response of these regions is significantly different from a normal untreated lesion. An example of overlapping occurring during such treatment is illustrated in the following experiment, Figure 2.6. When applying multiple spots with insufficient separation (in this case 2 mm) the heating effect of adjacent spots begin to merge, producing a higher maximum temperature across a larger surface area. This of course is undesired, particularly when there are a number of laser passes across the skin surface.

When comparing the above results with an identical laser on the same skin but this time with more than the minimum separation, 5mm, the importance of a controlled thermal build-up can be seen. Two shots with to low a gap results in overlapping give rise to a maximum temperature of 42.5°C. Whilst two identical shots placed on the same skin 5mm apart give a maximum temperature of 37°C Figure 2.7, with no apparent excessive temperatures. This difference is enough to affect treatment increasing the risk of adverse effects, particularly when treating sensitive skin on the face and neck.

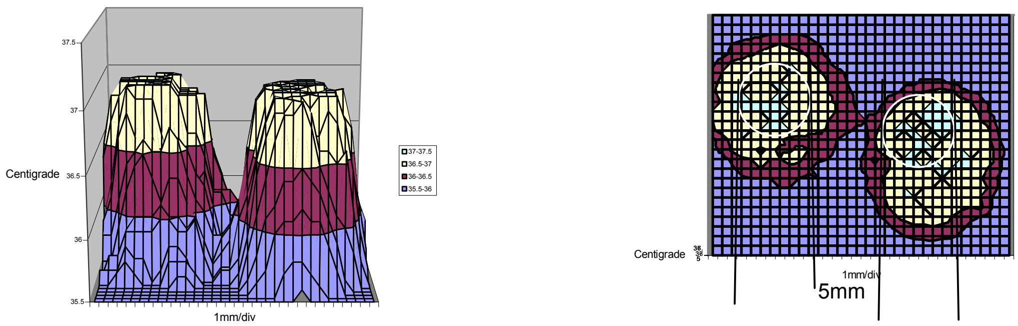

Figure 2.7 Effect of increasing separation between individual dye laser spots (585nm, 5J/cm^2)

The need to control more accurately the exact location of each laser spot is essential for optimum treatment efficacy. One method of achieving this is by computer scanning.

3.1 Computerised Laser Scanning:

Having established the parameters relating to laser spot positioning the possibility of achieving reproducible laser coverage of a lesion by automatic scanning becomes a reality. This has potential advantages, which include:

- Accurate positioning of the spot with the correct spacing from the adjacent spots,

- Accurate timing allowing the placement at a certain location at the appropriate lapsed time.

There are some disadvantages that include the need for additional equipment and regulatory approvals for certain market sectors. A computerised scanning system has been developed [Clement *et al*, 1999] that illuminates the tissue in a pre-defined pattern, Figure 3.1.

Figure 3.1 Treatment pattern produced held system [Clement *et al*, 1999]

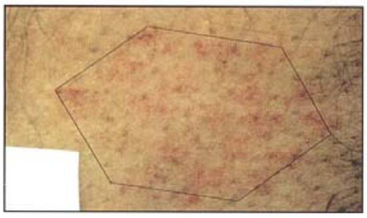

Figure 3.2 Treatment coverage using the hand- by the computer controlled scanning system

Sequential pulses are not placed adjacent to an immediately preceding pulse thereby ensuring the minimum of thermal build-up. Clement *et al* [1999] carried out a trial, illustrating treatment coverage using a hand-held system compared to a controlled computer scanning system. Two adjacent areas (lower arm) were selected and shaved. A marked hexagonal area was subjected to 19 shots using a hand-held system, Figure 3.2.An adjacent area of skin was treated with a scanner whose computer control is designed to uniformly fill the area with exactly 19 shots. Such tests were repeated and the results statistically analysed showed that, on average, only 60% of area is covered by laser spots [Clement *et al*, 1999]. The use of thermography allowed the validation and optimisation of this automated system in a way that was impossible without thermal imaging technology. The following sequence of thermal images, Figure 3.4, captures the various stages of laser scanning of the hand using a dye laser at 5.7 J/cm^2.

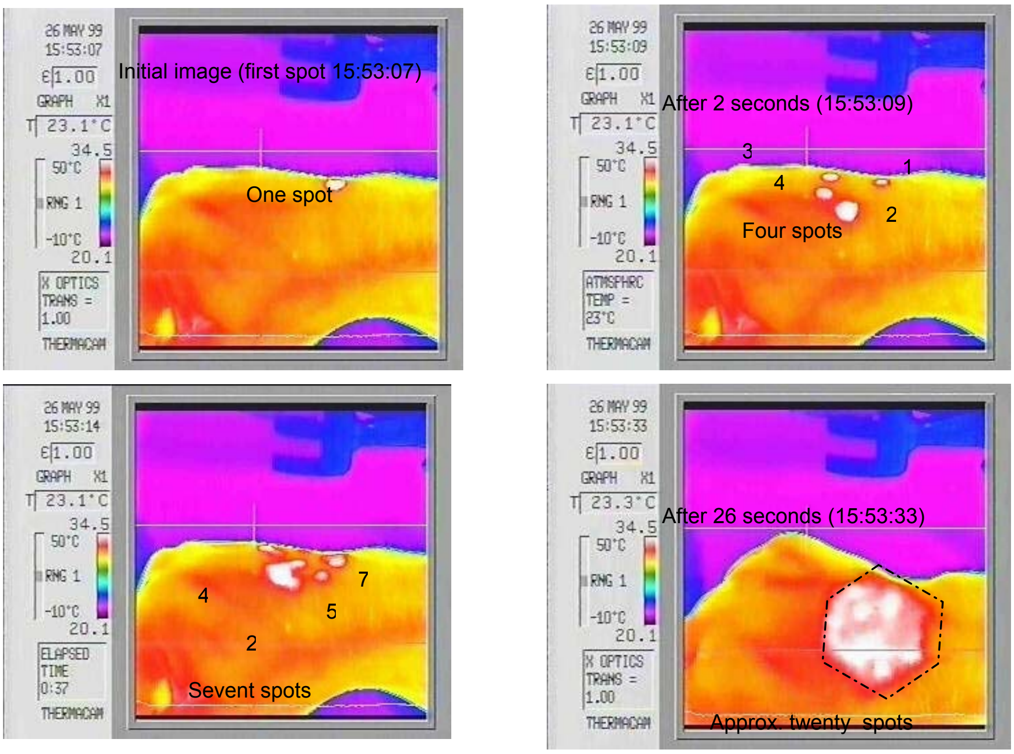

Figure 3.4 Sample of thermograms during computer laser scanning

The optimum temperate range confirms that the spot temperature from individual laser beams will merge and that both the positioning of spots and the time duration between these spots dictate the efficacy of treatment.

4 Computer Model:

In order to realistically model laser-tissue interactions, it is necessary to consider the following; Transport of the incident laser radiation through the tissue media, scattering and absorption of the photons as they propagate through the tissue, local energy deposition, due to absorbed photon energy being converted into

thermal energy and thermal transport mechanisms, with conduction being dominant within the tissue, together with appropriate surface boundary conditions that permit convection losses to be incorporated.
A two-stage approach is adopted, where the radiation transport is modelled using the well-established Monte-Carlo method [e.g. Wilson & Adam 1983]. Simple exponential attenuation with penetration depth (the well-known Beer-Lambert Law) is inappropriate where the scattering process is significant compared to photon absorption. Particular refinements to the Monte-Carlo method include photon weighting techniques that permit more rapid statistical convergence [Cashwell & Everett, 1959, Welsh & van Gemert, 1995].
The two-dimensional Cartesian thermal transport equation is:

$$\nabla T^2 + \frac{Q(x,y)}{k} = \frac{1}{\alpha}\frac{\partial T}{\partial t} \tag{4.1}$$

where $T(x,y,t)$ is the local instantaneous temperature, k is the thermal conductivity (varies with each layer)α is the thermal diffusivity (defined as $k/\rho C$, where ρ is the layer density and C is the specific heat of that layer)The volumetric source term, $Q(x,y)$, is obtained from the solution of the Monte-Carlo radiation transport problem.

Wilson & Adam [1983] first applied the Monte-Carlo method to model laser-tissue interaction. The Swansea research group has extended this technique to include:

- Optical physics such as Fresnel scattering and refraction at internal tissue interfaces
- Coupled radiation and thermal transport modelling, where the laser energy deposited within the tissue structures is represented by a heat source term in the well-known time-dependent thermal transport equation.
- Convective Boundary conditions. This is important in evaluating the possible benefits of forced-air skin cooling.
- Modelling of the temporal profile of the laser pulse, permitting investigation of pulse shape and duration.
- A thermal damage model, using a standard Arrhenius integral approach to predict the tissue necrosis induced through localised thermal effects. This is particularly pertinent to modelling the Port-Wine Stain problem, where the target blood vessels are thermally disrupted , leading to necrosis.
- Photo-chemistry kinetics, used to model photodynamic therapy problems [see for example omlc.ogi.edu/pdt/index.html].
- Multiple blood vessels in both a 2D model and a 3D model [Daniel, 2002].

Coding was accomplished in Visual C++, running in a Windows XP environment.
The thermographic data has been particularly useful in validating the computational model. We present two brief case studies to illustrate this.

4.1 Case Study 1 Port Wine Stain:

Figure 4.1 illustrates a typical problem with multiple blood vessels buried at different depths within the dermis. A laser wavelength of 585nm is preferentially absorbed by haemoglobin within the blood, but there is partial absorption in the melanin rich basal layer in the epidermis. The objective is to thermally damage the blood vessel, by elevating its temperature, while ensuring that the skin surface temperature is kept low.

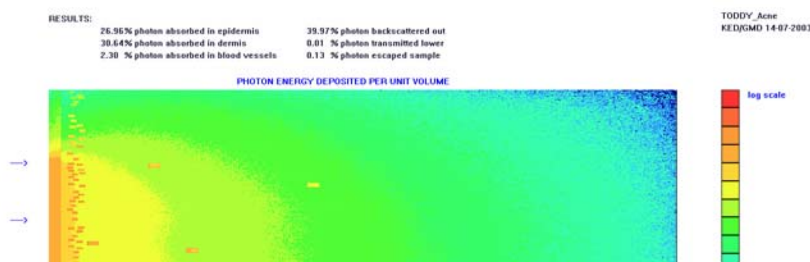

Figure 4.1 Photon distribution for a PWS typical problem

The model suggests that it is possible to selectively destroy the PWS blood vessels, by elevating them to a temperature in excess of 100 °C, causing disruption to the small blood vessels, whilst maintaining a safe skin surface temperature. The thermographic results presented in section 2 validate the predictions of the computer model [Daniel 2002].

4.2 Case Study 2 Laser Depilation

The 694 nm wavelength laser radiation is preferentially absorbed by melanin, which occurs in the basal layer and particularly in the hair follicle base, which is the intended target using an oblique angle of laser beam (see Figure 4.3).

A Monte Carlo analysis was performed in a similar manner to Case Study 1, where the target region in the dermis is the melanin rich base of the hair follicle. Figures 4.5 and 4.6 show the temperature-time profiles for 10 Jcm^2 and 20 Jcm^2 laser fluence [Trow, 2001]. These figures suggest that it might be possible to thermally damage the melanin-rich follicle base whilst restricting the skin surface temperature to values that cause no superficial damage. Preliminary clinical trials indicated that there was a beneficial effect, but the choice of laser parameters required optimising.

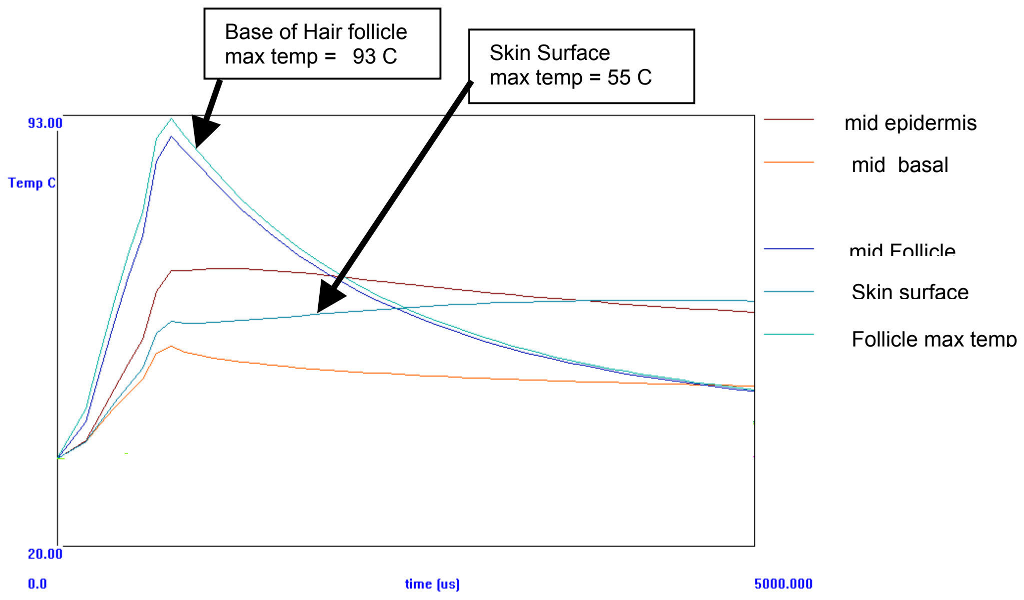

Figure 4.4 Temp -time profiles for $10Jcm^2$ ruby (694nm) 800 μs laser pulse on caucasian skin type **III.**

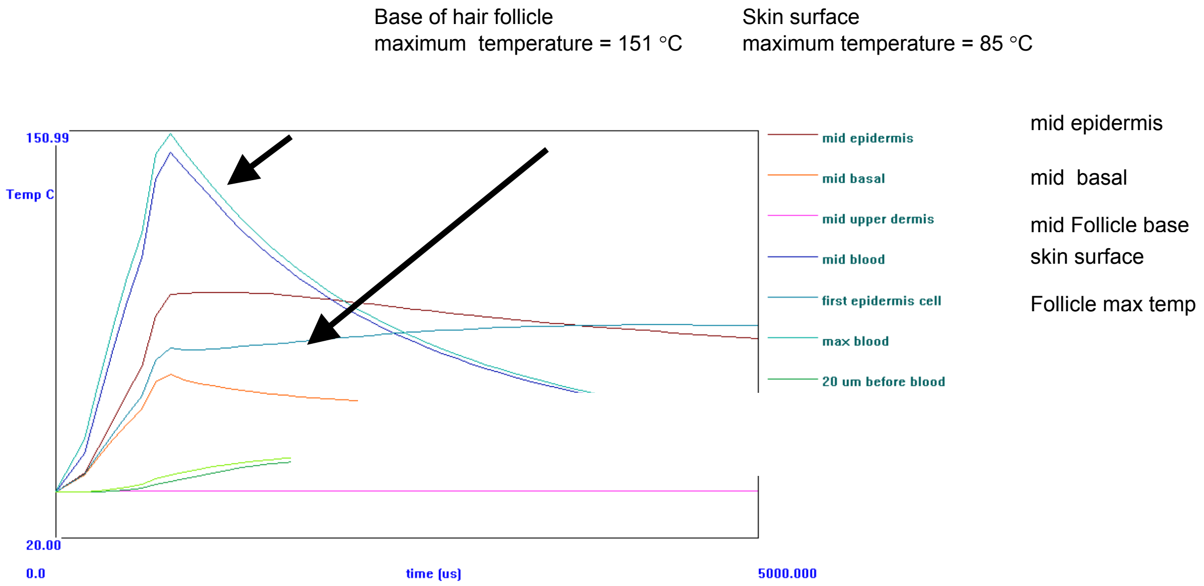

Figure 4.5 Temp –time profiles for 20Jcm^2 ruby (694nm) 800 µs laser pulse on Caucasian skin type III.

Thermographic analysis has proved indispensable in this work. Detailed thermometric analysis is shown in Figure 4.6. Analysis of this data shows that in this case, the surface temperature is raised to about 50 °C. The thermogram also clearly shows the selective absorption in the melanin-dense hair. The temperature of the hair is raised to over 207 °C. This thermogram shows the theory of selective wavelength absorption leading to cell necrosis. Further clinical trials have indicated a maximum fluence of 15 Jcm^2 for type III caucasian skin. Figure 4.7 illustrates a typical thermographic image obtained during the real-time monitoring.

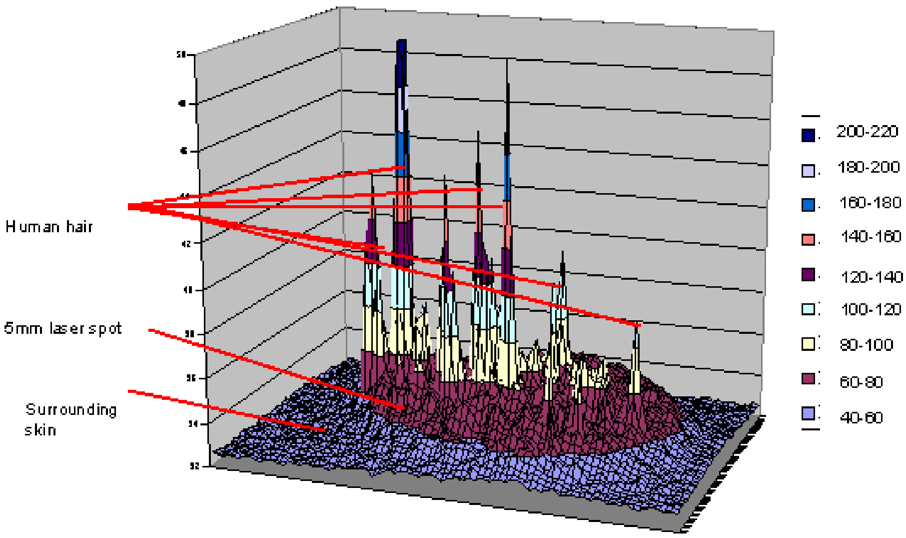

Figure 4.6 Post-processed results of Figure 4.6 for 5mm diameter 694 nm 20 Jcm^2 800µs ruby pulse

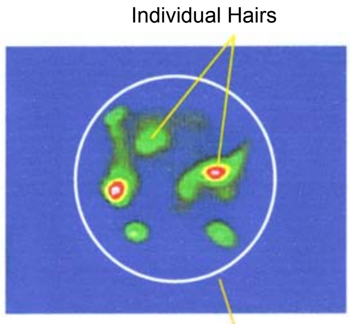

Figure 4.7 Simplified thermogram of ruby laser pulse, with 5mm spot and 20Jcm^2

6 Conclusions:

Infrared thermal imaging provides an important diagnostic tool for optimising laser parameters and facilitates calibration of a theoretical model, where surface convection conditions are a significant part of the treatment process. To date many thousands of patients have been treated with no burns or scarring reported and pain and treatment times have been minimised.

In summary therefore the tissue temperature can vary with position at constant energy density this is very important particularly in a litigious culture where a dermatologist cannot afford to scar a patient. It is difficult to imagine how this could have been proven without the application of thermography in the discipline.

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