Clinical use of pulsed photothermal radiometry

T. E. Milner[1], J. S. Nelson[1], N. Tran[1], A. Katzir[2]
L. O. Svaasand[3], S. L. Jacques[4]

[1]Beckman Laser Institute and Medical Clinic, University of California Irvine, USA
[2]School of Physics and Astronomy, Tel Aviv University, Israel
[3]University of Trondheim, The Norwegian Institute of Technology, Trondheim, Norway
[4]University of Texas M. D. Anderson Cancer Center, Houston, Texas USA

Abstract

The application of pulsed photothermal radiometry (PPTR) diagnostics to characterize port wine stain (PWS) lesions is discussed. A PPTR signal of a PWS in response to pulsed laser exposure is shown to consist of an initial "T-jump" due to epidermal melanin absorption and a "delayed thermal wave" resulting from laser generated heat in subsurface blood vessels diffusing to the skin surface. A prototype PPTR instrument incorporating an infrared fiber is constructed which facilitates convenient skin-site accessibility. Laser heating of the infrared collection optics results in an artifactual signal which overestimates the initial "T-jump". Magnitude of the error is measured and a method to eliminate it is suggested.

1. Introduction

Port wine stain (PWS) is a congenital, progressive vascular malformation of the dermis, involving capillaries, and possibly perivenular nerves, that occurs in an estimated 5 children per 1,000 live births. Although PWS may occur anywhere on the body, most lesions appear on the face and are noted to occur over the dermatome distribution of the first and second trigeminal nerves[1]. Historical treatments for PWS included scalpel surgery, ionizing radiation, skin grafting, dermabrasion, cryosurgery, tattooing, and electrotherapy. A more recent treatment option using a flashlamp pulsed dye laser has offered a superior approach in therapy due to its ability to destroy selectively cutaneous blood vessels. Light will pass through the epidermis and then be preferentially absorbed by hemoglobin (the major chromophore in blood) in the ectatic capillaries in the upper dermis. There, the radiant energy will be converted to heat causing thermal damage and thrombosis in the targeted vessels. However, the epidermis is not totally spared due to partial absorption of laser energy therein by melanin and to dissipation of heat from the injured vessels. Unfortunately, for many lesions, the threshold for epidermal changes following laser therapy is very close to the threshold for permanent blanching of the PWS.

Pulsed photothermal radiometry (PPTR) uses infrared radiometry to detect surface temperature rises induced by pulsed laser radiation. In practice, a pulsed laser is used to produce transient heating of the object under study. The subsequent temperature rise at the tissue surface creates an increase in infrared (blackbody) emission which is measured by a fast infrared detector. Absorption of the laser pulse produces an initial temperature "jump" ("T-jump") which denotes the temperature rise at the sample surface. The time dependence of the decay of the blackbody emission signal can be used to map the spatial distribution of heat deposition in the tissue under study.

0-8194-1103-5/93/$6.00

For the purposes of PPTR, PWS in human skin can be modeled as an absorbing surface layer (the epidermis which contains melanin) and a subsurface absorber (the blood vessels which contain hemoglobin). In the PWS model, if a pulsed laser light source irradiates the skin, an initial "T-jump" is observed due to heating of the epidermis by the laser pulse as a result of melanin absorption (Figure 1).

Subsequently, heat generated in the subsurface blood vessels due to hemoglobin absorption is detected by PPTR as a delayed thermal wave, which is characterized by a peak after the laser pulse, since heat diffuses from the buried blood vessels to the surface (Figure 1).

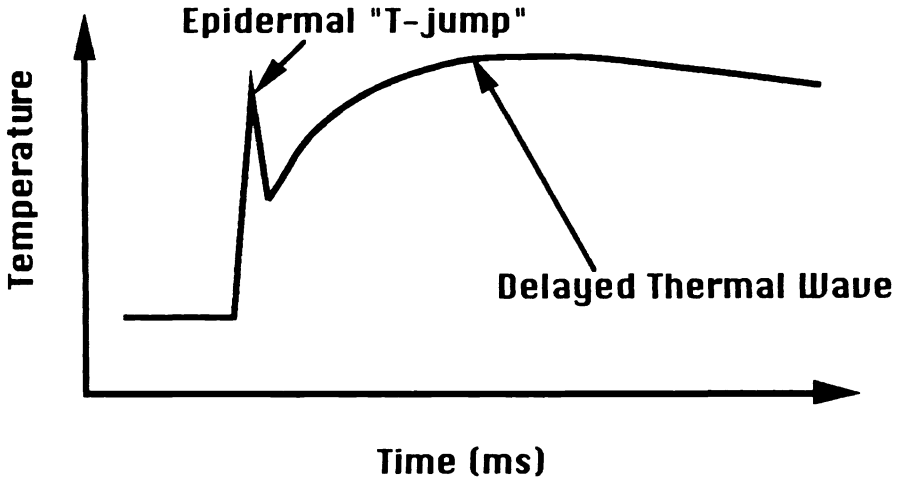

Figure 1 PPTR signal of model PWS.

2. Materials and Methods

Clinical PPTR measurements of PWS, using a conventional liquid-N_2-cooled infrared detector, are problematic; due to variable skin-detector geometry site-to-site calibration is difficult. A possible solution to this problem would use a thermo-electric cooled detector but this substantially adds to the total system cost. Because of its flexibility and maneuverability an infrared collection fiber has greater skin site-to-site accessibilty. Despite this benefit, an alternate signal processing scheme is needed to compensate for the substantial signal-to-noise ratio loss found in an infrared fiber collection system.

2.1 Prototype PPTR instrument

Modulation of the infrared signal combined with homodyne detection offers a solution to the signal detection problem. We have constructed a hand piece which houses infrared and visible collection fibers (Figure 2). The diffusely reflected laser light scattered from the skin, is collected by a GRIN® lens (NSG America, Somerset NJ) coupled to an optical fiber and detected with a silicon photo diode. The PPTR signal is collected by an infrared fiber (Ag-Halide, Tel-Aviv University, Israel), modulated with a mechanical chopper (3kHz), and focused onto a liquid-N_2-cooled HgCdTe detector (New England Research Center, Sudbury MA). The signal is demodulated with a lock-in amplifier (τ= 10 ms), digitized by an A/D converter and stored in a computer for further analysis.

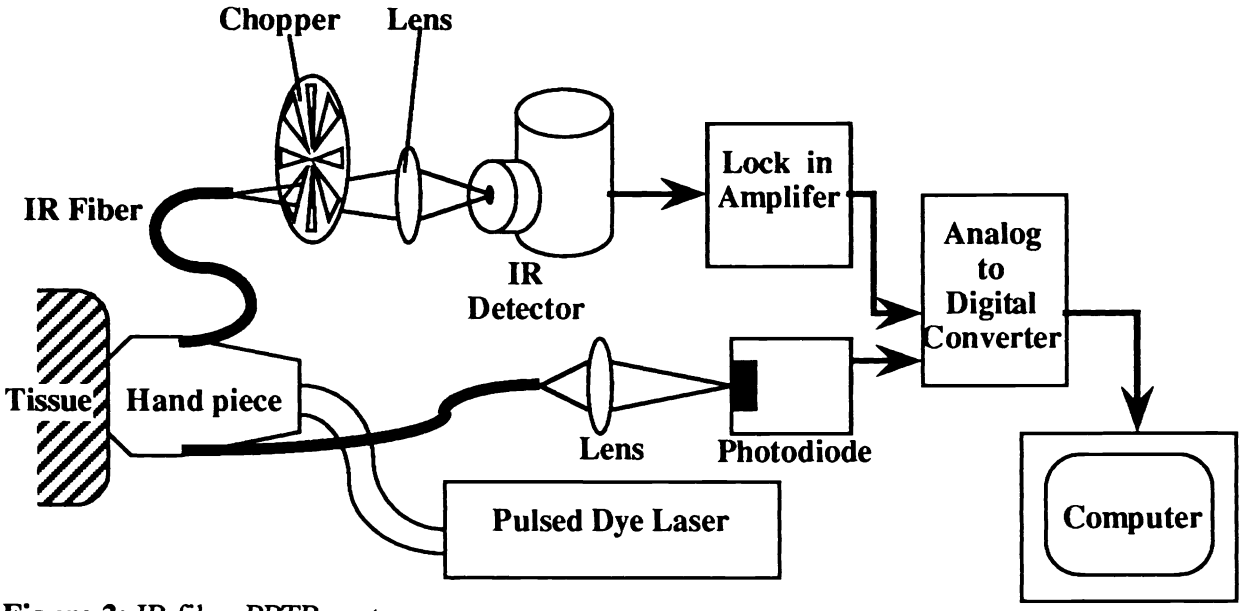

Figure 2: IR fiber PPTR system.

2.2 Laser heating of infrared optics

We have identified a physical effect which produces a signal artifact in measurements of the initial epidermal "T-jump." When laser light is incident on the PWS skin surface, approximately 4% of the incident laser light is specularly reflected and, depending on the PWS absorbance, 30-40% of the incident light is diffusely reflected. Some of the reflected light reaches the entrance pupil of the infrared collection system and is absorbed by the infrared fiber This causes rapid heating of the fiber leading to increased infrared emission which is measured by the HgCdTe detector. The artifact adds to the signal obtained from epidermal melanin resulting in a pseudo thermal measurement that overestimates the true epidermal heating.

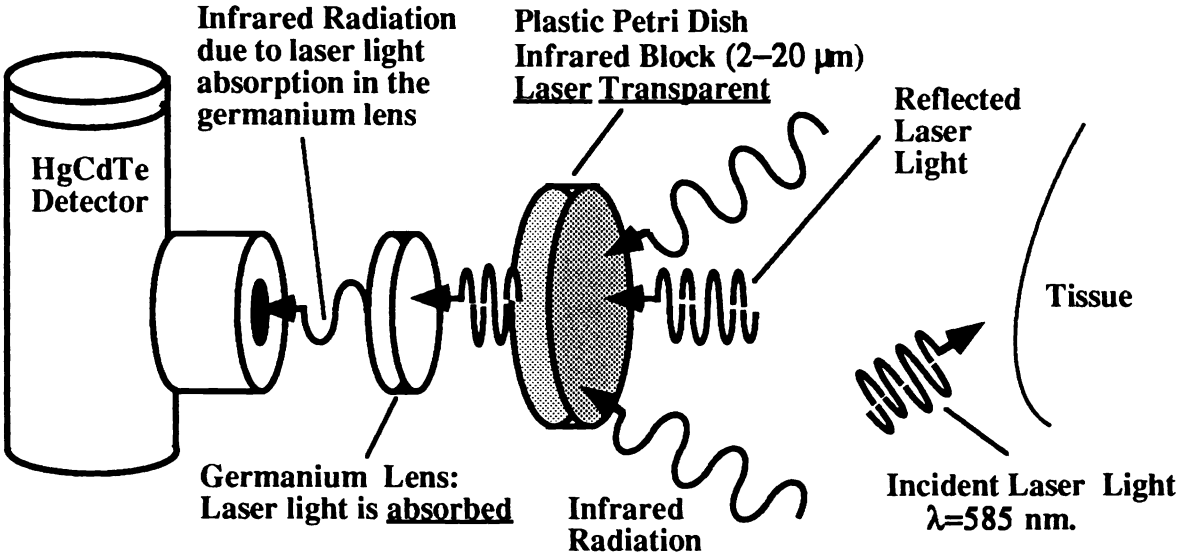

Figure 3 Experiment to demonstrate laser heating of the infrared optics.

We have confirmed the existence and measured the relative magnitude of the heating effect with various experiments and calculations. The simplest experimental confirmation of the effect was obtained by covering the entrance pupil of an infrared optical system with a plastic petri dish and observing the PPTR signal (Figure 3). Two PPTR measurements of a PWS were made; (1) without a plastic petri dish (2) with a plastic petri dish positioned at the entrance pupil of the infrared optical system.

3. Results

3.1 Patient measurements

We have carried out initial clinical tests of the fiber-based PPTR instrument diagrammed in Figure 2. With a sub-therapeutic light dose of 4 Jcm^{-2} from a flash lamp pumped pulsed dye laser (λ_p=585 nm, t_p=0.45 ms) the PPTR signals from a patient's normal and PWS skin were measured (Figure 4).

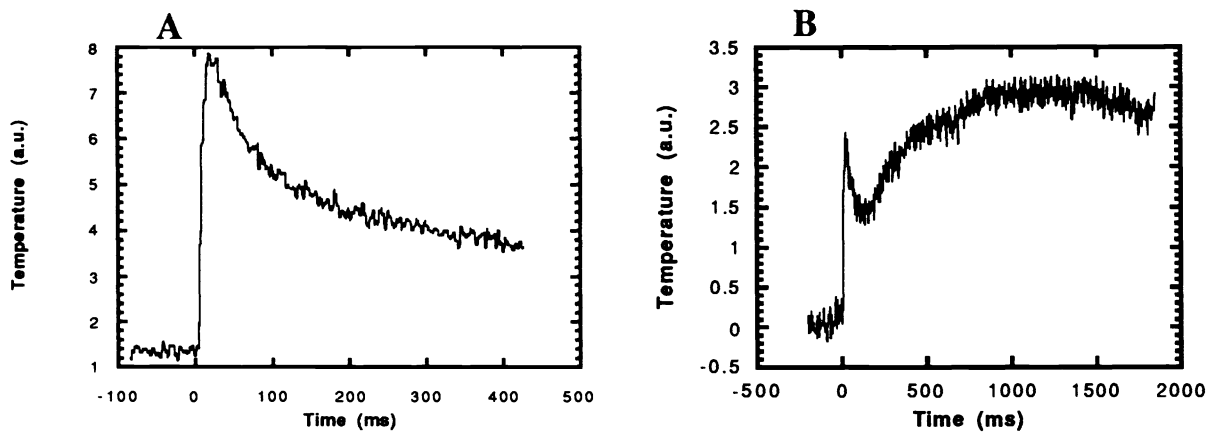

Figure 4 Patient PPTR measurements of (a) normal and (b) PWS skin.

The PPTR signal from normal skin shows an initial epidermal "T-jump" followed by characteristic monotonic cooling. The PPTR response of PWS tissue shows a relatively smaller epidermal "T-jump" and a large amplitude delayed thermal wave which peaks at ~1 s corresponding to a PWS depth of approximately 750 μm. Due to light absorbed in the PWS, the smaller observed "T-jump" is partially due to a decrease in back scattered laser in the skin.

3.2 Effect of lens heating

In the system diagrammed in Figure 3 plastic strongly absorbs incident infrared radiation ($\lambda > 2$ μm) and one would not expect a measurable thermal signal in response to a laser pulse. However, a thermal signal was repeatedly observed and found to follow closely the shape of the laser pulse. After eliminating electrical noise as a possible source, we concluded that the thermal signal results from visible laser light absorption in the germanium lens. A simple calculation reveals the thermal relaxation time corresponding to visible light absorption in germanium is on the order of nanoseconds. Hence, it is expected that the thermal signal will closely follow the 0.45 ms laser pulse; this was consistently observed in our experiments.

We determined the relative importance of the lens heating effect through experiments conducted on our chicken comb animal model. The histoanatomy of PWS and the red chicken comb are analogous. Two PPTR signals were observed corresponding to clear

and plastic covered entrance pupils. To correct for the lens heating effect, the signal collected with the pupil covered was scaled to compensate for reflective losses from the plastic and subtracted from the clear aperture signal; the original measurement and the corrected signal are shown in Figures 5a and 5b, respectively.

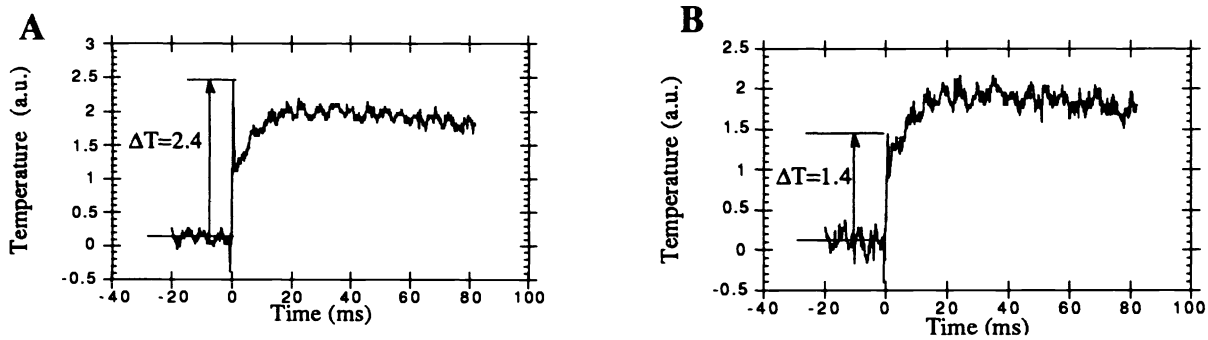

Figure 5: Effect of laser heating of the infrared optics. Figures 5a and 5b show the uncorrected and corrected initial epidermal "T-jump," respectively.

It is apparent that the size of the initial epidermal "T-jump" can be significantly related to the lens heating effect; the corrected temperature jump is noticeably less than in the original measurement. One can conclude accurate PPTR measurements require elimination of artifactual heating of the collection optics.

4. Discussion

4.1 Epidermal "T-jump"

Prior to the institution of laser PWS therapy, the physician must know on an individual patient basis the maximum incident light dose that may be delivered to the subsurface PWS blood vessels without damaging the normal overlying epidermis. In transport media where scattering can be neglected, the laser induced initial "T-jump" is

$$\Delta T = \frac{\mu_a \phi_o}{\rho C} \tag{1}$$

where μ_a (cm^{-1}) is the epidermal absorption coefficient, ϕ_0 (Jcm^{-2}) is the incident light dose, and ρC ($J°K^{-1}cm^{-3}$) is the product of the tissue density and specific heat. However, in turbid media, such as human skin, the backscattered light augments the delivered incident light dose. In this case, the "T-jump" is[2,3].

$$\Delta T = \frac{\mu_a \phi_o}{\rho C}\left[1 + 2\frac{1+r_i}{1-r_i}R_d\right] \tag{2}$$

where r_i is the averaged internal reflectance at the air/tissue interface (approximately 0.5). R_d is the diffuse reflectance resulting from light that enters the skin, is scattered, and subsequently reemerges from the tissue (for example, if R_d is 0.50, and the term $2(1+r_i)/(1-r_i)$ is approximately 6, the quantity within the brackets is equal to 4; thus, the backscattered light skin can cause ΔT to be four times greater than from a clear medium without scattering). By measuring ΔT, ϕ_0, R_d, and using approximate values for ρC, and

r_i, the epidermal absorption coefficient, μ_a, can be determined. Initial "T-jump" measurements can be made at sub-therapeutic incident light doses and once μ_a has been calculated, "T-jumps" at all incident light doses may be predicted using equation 2.

The laser heating effect can be eliminated by incorporating a dielectric interference filter into the infrared collecting optics (Figure 6). The special purpose filter is unique because few materials exist that have the required infrared and laser wavelength optical properties needed for proper functioning. A candidate filter consists of a 10-16 layer stack of quarter-wave high-low index materials on an infrared transmitting substrate. Further instrument development will test the performance of a dielectric filter to eliminate heating in the collection optics.

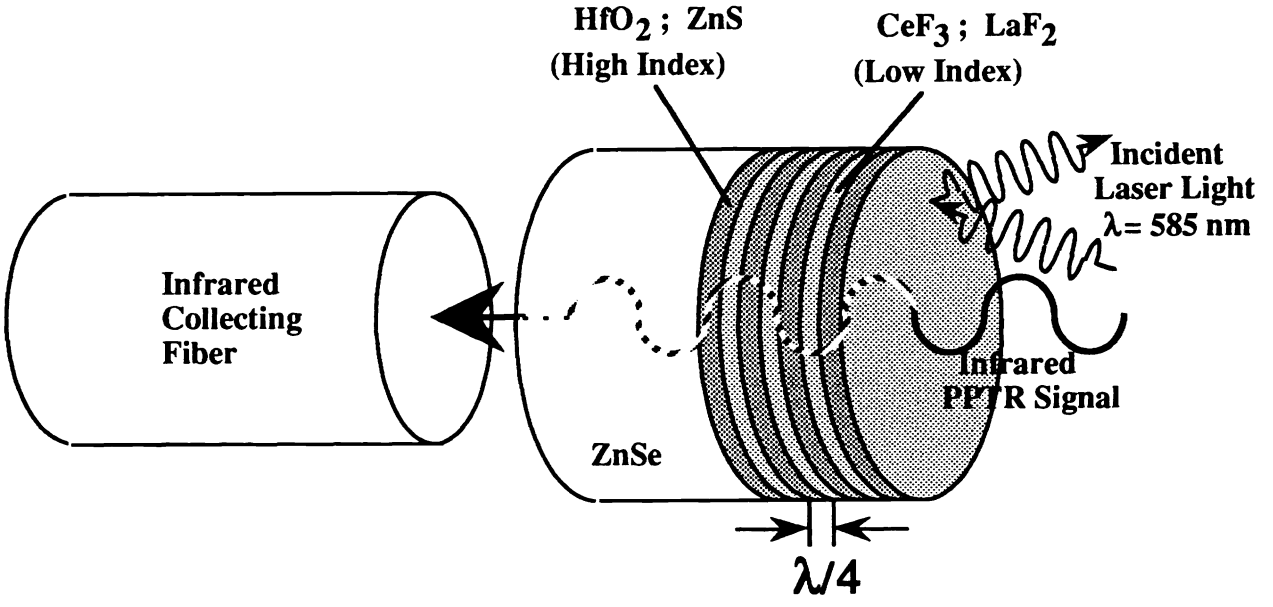

Figure 6 Interference filter positioned at the tip of infrared collection fiber.

5. Conclusions

PPTR is a useful technique to characterize PWS prior to the application of laser therapy. A PPTR measurement can determine the concentration of epidermal melanin, depth and initial temperature of the PWS. The fiber based prototype instrument we have demonstrated can conveniently make site-to-site PPTR measurements of a patient's PWS. Heating of infrared collection optics by diffusely reflected laser light results in a signal artifact that overestimates the initial "T-jump". For correct laser dosimetry, future work will seek to eliminate artifactual signals and determine the initial temperature in the skin immediately after the laser pulse.

6. Acknowledgments

This research was supported with grants from the Whitaker Foundation and NIH Biomedical Research Technology Program (R03 RR06988-01). Sun Microsystems (Mountain View, CA) provided computer resources necessary to complete the project.

7. References

1. J. B. Mulliken, A. R. Young, Vascular Birthmarks - Hemangiomas and Malformations. W. B. Saunders Co., Philadelphia, PA, 1988.
2. R.R. Anderson, H. Beck, U. Bruggemann, W. A. Farinelli, S.L. Jacques, and J. A. Parish, "Pulsed photothermal radiometry in turbid media: internal reflection of backscattered radiation strongly influences optical dosimetry," *Appl. Opt.*, Vol. 28, pp. 2256-2262, 1989.
3. S. A. Prahl, I. A. Vitkin, U. Bruggemann, B. C. Wilson, and R. R. Anderson, "Determination of optical properties of turbid media using pulsed photothermal radiometry," *Phys. Med. Biol.* Vol. 36, pp. 1203-1217, 1992.