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Selection of optimal infrared detector for pulsed photothermal profiling of vascular lesions

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ABSTRACT

Selection of infrared (IR) detector is a key consideration in designing an experimental setup for temperature depth profiling using pulsed photothermal radiometry (PPTR). In addition to common detector characteristics, such as the spectral response, detector noise, and response speed, application-specific details must be taken into account to ensure optimal system performance. When comparing detectors with different spectral responses, blackbody emission characteristics must be considered in terms of influence on radiometric signal amplitude, as well as on background shot noise. In PPTR, optical penetration depth of the sample in the acquisition spectral band is also an important factor, affecting the stability of the temperature profile reconstruction. Moreover, due to spectral variation of IR absorption coefficient in watery tissues, the acquisition band must be appropriately narrowed to ensure the validity of the customary approximation of a constant absorption coefficient value in signal analysis. This reduces the signal-to-noise ratio, adversely affecting the stability and quality of the temperature profile reconstruction. We present a performance analysis of PPTR depth profiling in human skin using commercially available IR detectors (InSb, HgCdTe), operating in different spectral bands. A measurement simulation example, involving a simple, hyper-Gaussian temperature profile, and realistic noise levels, illustrates their expected performance.

Keywords: Pulsed photothermal radiometry, temperature depth profiling, image reconstruction, infrared absorption coefficient, port wine stain.

1. INTRODUCTION

Pulsed photothermal radiometry (PPTR) involves time-resolved acquisition of blackbody emission from a sample after pulsed laser exposure. With optical and thermal properties of the sample known, measured radiometric signal enables reconstruction of the laser-induced axial temperature profile, bearing information on structure and/or chromophore distribution in the sample.^{1,2} Such photothermal profiling technique has been successfully applied to strongly scattering biological tissues and tissue phantoms.^{3,4,5,6} Jacques *et al.*⁷ proposed to use PPTR for characterization of port wine stain (PWS) birthmarks in human skin. This approach was developed further and studied by Milner *et al.*^{8,9} and others,^{10,11,12} up to the point of demonstrating three-dimensional imaging of subsurface vasculature by utilizing a fast IR camera.^{13,14,15}

PWS are subsurface hypervascular lesions, located within the most superficial millimeter of human skin. Blood vessel size and depth distribution in PWS varies widely from patient to patient, with the highest fractional blood content on average 0.2–0.4 mm below the epidermal-dermal junction.¹⁶ PWS therapy currently utilizes selective photothermolysis of the ectatic vasculature using pulsed lasers (KTP/YAG, flashlamp-pumped dye). Numerous recent studies have indicated that such therapy could benefit from optimization on individual patient basis,^{17,18} in particular when it involves cryogen spray cooling.^{19,20,21} Unfortunately, none of the established medical imaging techniques offers the required combination of contrast, spatial resolution, and availability.

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A key consideration in designing a setup for temperature depth profiling using PPTR is selection of infrared (IR) detector. In order to ensure optimal system performance, application-specific details should be taken into account, in addition to common detector characteristics such as spectral response, detector noise level, and response speed. When comparing detectors with different spectral responses, blackbody emission characteristics must be analyzed in terms of the influence on radiometric signal amplitude, as well as on the amount of background shot noise. Optical penetration depth of the sample in the acquisition spectral band is also known to affect the stability of the temperature profile reconstruction.⁸ Moreover, biological tissues regularly display a pronounced variation of the absorption coefficient $\mu_{\text{IR}}(\lambda)$ in the involved mid-IR spectral range (see Fig. 1). In order to validate the approximation of a constant μ_{IR} , customarily implied to reduce the mathematical complexity of PPTR signal analysis, the detection band must be appropriately narrowed.^{22,23} Such narrowing of the detection band will reduce the signal-to-noise ratio (SNR), adversely affecting the stability and accuracy of the temperature profile reconstruction.

We present here a preliminary performance analysis of PPTR depth profiling in human skin using two commercially available IR detectors, used in three different spectral bands: InSb (3–5 μm) and MCT (i.e., HgCdTe; 7–10 μm and 10–12 μm). For each of these spectral bands, we determine first the optimal spectrally narrowed acquisition window, which maximizes the radiometric signal while allowing the use of the customary approximation of a constant μ_{IR} in temperature profile reconstruction. This selection is based on computation of the effective IR absorption coefficient μ_{eff} for each spectral window, and quantitative analysis of the profile reconstruction accuracy in simulations involving a simple, hyper-Gaussian temperature profile.

Subsequently, the profiling performance is compared between the three spectral bands, taking into account the actual spectral variation of $\mu_{\text{IR}}(\lambda)$ within each optimal acquisition window, and realistic SNR levels. The latter take into account both the noise specification of commercially available IR detectors, as well as shot noise in PPTR signals and background IR emission.

2. BACKGROUND

2.1. PPTR profiling - basics

Detailed discussion of PPTR profiling in semi-infinite medium can be found elsewhere.^{8,10} In short, one-dimensional analysis is justified when the laser-irradiated spot is much larger than the involved optical penetration depths and thermal diffusion lengths. Due to the finite IR absorption coefficient (μ_{IR}), radiometric emission from a non-uniformly heated sample is composed of correspondingly attenuated contributions from all depths z . For modest temperature rises, Planck's law of radiation, describing the surface density of locally emitted power at wavelength λ , is linearized around the initial skin temperature ($B_{\lambda}(T) \approx B_{\lambda}(T_0) + B'_{\lambda}(T_0) \Delta T$), and the radiometric signal $S_{\lambda}(t)$ is expressed as

$$S_{\lambda}(t) = C B_{\lambda}(T_0) + C B'_{\lambda}(T_0) \mu_{\text{IR}} \int_{z=0}^{z=\infty} \Delta T(z,t) e^{-\mu_{\text{IR}} z} dz \quad (1)$$

where constant C accounts for sample emissivity, numerical aperture and losses of the collection optics, and detector specifics (such as sensitivity and spectral bandwidth).

The temperature field evolution $\Delta T(z,t)$ after pulsed laser heating can be calculated from the initial laser-induced temperature rise $\Delta T(z',0)$ using the Green's function approach. The transient part of the radiometric signal $\Delta S(t)$ is thus expressed as

$$\Delta S_{\lambda}(t) = C B'_{\lambda}(T_0) \mu_{\text{IR}} \int_{z'=0}^{z'=\infty} \Delta T(z',0) \int_{z=0}^{z=\infty} K_T(z',z,t) e^{-\mu_{\text{IR}} z} dz dz'. \quad (2)$$

where $K_T(z',z,t)$ represents the thermal Green's function of the sample. This function is known in advance, and the integration over z can be performed, leading to

$$\Delta S_{\lambda}(t) = \int_{z=0}^{z=\infty} K(z,t) \Delta T(z,0) dz . \quad (3)$$

PPTR signal $\Delta S_{\lambda}(t)$ is thus linearly related to the initial temperature profile $\Delta T(z,0)$. For all practical purposes, PPTR signals and temperature profiles can be treated as sets of discrete values, and so the linear mapping (3) can be represented by multiplication of the initial temperature profile vector $\mathbf{T}$ ($T_j = \Delta T(z_j,0)$) with a kernel matrix $\mathbf{K}$:

$$\mathbf{S} = \mathbf{K} \cdot \mathbf{T}; \quad K_{i,j} = K(z_i, t_j) \Delta z . \quad (4)$$

Definition of the kernel function $K(z,t)$ is evident from (2) and (3). In the examples below, we model human skin as thermally homogeneous semi-infinite medium. The kernel function then has a form:⁸

$$K(z',t) = \frac{1}{2} C B_{\lambda}'(T_0) \mu_{\text{IR}} e^{-\frac{z'^2}{4Dt}} \left\{ \text{erfcx}(u_-) + \text{erfcx}(u_+) - \frac{2h}{(h - \mu_{\text{IR}})} [\text{erfcx}(u_+) - \text{erfcx}(u_1)] \right\}, \quad (5)$$

where $\text{erfcx}(u) = \exp(u^2) \cdot [1 - \text{erf}(u)]$, $u_{\pm} = \mu_{\text{IR}} \sqrt{Dt} \pm z'/(2\sqrt{Dt})$, $u_1 = h\sqrt{Dt} \pm z'/(2\sqrt{Dt})$, D is the heat diffusivity ($D = 1.1 \times 10^{-7} \text{ m}^2 \text{ s}^{-1}$),²⁴ and h is reduced heat transfer coefficient, describing convective and radiative heat transfer at the skin surface ($h = 0.02 \text{ m}^{-1}$).

2.2. Spectral variation of μ_{IR}

Expressions (1)–(5) are valid rigorously only for the case of monochromatic detection, or when the absorption coefficient μ_{IR} is constant throughout the detection band. However, most biological tissues display a pronounced variation of the absorption coefficient $\mu_{\text{IR}}(\lambda)$ in the involved mid-IR spectral range (inherited from both water and protein constituents - see Fig. 1).^{25,26} Realistic radiometric signals are thus computed by summing up spectral contributions (2) within the detection band (λ_l to λ_h):

$$\Delta S(t) = C' \int_{\lambda_l}^{\lambda_h} R(\lambda) \int_{z=0}^{\infty} B_{\lambda}'(T_0) \mu_{\text{IR}}(\lambda) \Delta T(z,t) e^{-\mu_{\text{IR}}(\lambda)z} dz d\lambda , \quad (6)$$

where spectral response of the detector $R(\lambda)$ is also taken into account. (Note the re-defined constant C').

Figure 1: Absorption coefficient of human skin in mid-IR spectral region, computed as 75% of μ_{IR} for water (lighter line),²⁵ increased by the contribution from 25 wt.% of collagen (darker line).²⁶ Within the customary detection band of the InSb detector (3–5 μm), μ_{IR} varies by two orders of magnitude, and by nearly one order of magnitude within the detection band of the MCT (7–12 μm).

The effective value of the absorption coefficient, which should be used in PPTR image reconstruction algorithms that neglect the spectral dependence $\mu_{\text{IR}}(\lambda)$ within the detection band, can be obtained by inserting a spectrally constant value μ_{eff} into (6). From the requirement that the obtained signal expression equals the r.h.s. of (6) for an arbitrary temperature profile $\Delta T(z,0)$ follows the implicit equation for the effective absorption coefficient μ_{eff} .^{22,23}

$$\mu_{\text{eff}} e^{-\mu_{\text{eff}} z} = \frac{\int_{\lambda_l}^{\lambda_h} R(\lambda) B_{\lambda}'(T_0) \mu_{\text{IR}}(\lambda) e^{-\mu_{\text{IR}}(\lambda) z} d\lambda}{\int_{\lambda_l}^{\lambda_h} R(\lambda) B_{\lambda}'(T_0) d\lambda}. \quad (7)$$

3. METHODS

The effective absorption coefficient values μ_{eff} for various spectral acquisition windows (λ_l to λ_h) are obtained as follows: The IR absorption coefficient of human skin is computed first at 100 equidistant points within the spectral band, by applying a 4-point interpolation to the absorption data in Figure 1 (75% of μ_{IR} for water²⁵ plus 25% of that for collagen²⁶). Both integrals in (7) are then evaluated using the Simpson's rule.²⁷ Finally, the implicit equation (7) is solved for μ_{eff} using the Jacobi iterative method (with predefined maximum difference between consecutive solutions), implemented as an ANSI C code. The solutions $\mu_{\text{eff}}(z)$ are assessed at 100 equidistant depths ($z = 1\text{--}100 \mu\text{m}$).

The simulated PPTR signals are calculated using a custom code written in C++ programming language (Microsoft). The spectral acquisition window is divided into 100 intervals, and monochromatic signals are calculated for each wavelength using (4). The obtained monochromatic components $\Delta S_{\lambda}(t)$ are then summed up into a spectrally composite signal by using appropriate weights $R(\lambda)$, according to (6). For each example, 1000 signal values are computed at equidistant time points within a time interval of 1 s (acquisition rate: 1000 s⁻¹).

Reconstruction of the initial temperature profile $\Delta T(z,0)$ from the PPTR signal $\Delta S(t)$ is a severely ill-posed inverse problem. We solve it by iterative minimization of the residual norm $\|\mathbf{K} \cdot \mathbf{T}^{(n)} - \mathbf{S}\|$ using the conjugate-gradient algorithm with a nonnegativity constraint.⁸ A custom reconstruction algorithm was written in MatLab (The MathWorks, Inc.) using Optimization Toolbox. We use function *lsqlin*, with kernel matrix $\mathbf{K}$, signal vector $\mathbf{S}$, and nonnegativity constraint as arguments. The temperature profile values are sought at 100 equidistant points within the 1 mm thick superficial layer of "skin".

4. RESULTS – OPTIMAL ACQUISITION BANDS

4.1. InSb detector, $\lambda = 3.0\text{--}5.3 \mu\text{m}$

Using (7), we compute first the effective absorption coefficient values μ_{eff} of human skin for detection with InSb radiation detector in the customary spectral band of 3–5 μm . We use the $\mu_{\text{IR}}(\lambda)$ data from Figure 1, and the detector spectral response is modeled as $R(\lambda) = \lambda/\lambda_p$ between λ_l and λ_h , and $R(\lambda)=0$ elsewhere. (Note that evaluation of absolute signal amplitude will not be required in this step, since the calibration constant cancels out in (7).) The wavelength of peak detector response is set to $\lambda_p = 5.3 \mu\text{m}$, typical of InSb photovoltaic detectors cooled with liquid nitrogen (e.g., Hamamatsu P5968-100).

Figure 2 presents a series of results, where the lower limit of the acquisition window runs across the spectral band ($\lambda_l = 3.0\text{--}5.1 \mu\text{m}$), while the upper limit is fixed at $\lambda_h = \lambda_p = 5.3 \mu\text{m}$ (note the wavelength range written in each graph). The obtained effective values μ_{eff} are double-valued and in general vary with depth (z). This indicates that no fixed value μ_{eff} yields a correct PPTR signal for an arbitrary temperature profile.

The variation of optimal μ_{eff} with source depth (z) is obviously smaller for narrower acquisition windows. These will, however, yield lower signal-to-noise ratios (SNR) of the PPTR signals, adversely affecting the temperature profile reconstruction. The question is therefore, which is the widest acquisition window that will permit the use of a constant μ_{IR} in the reconstruction algorithm (while not sacrificing too much of the detected radiometric power). To answer this

question, we look first at the average μ_{eff} within the 100 μm thick superficial layer as a function of the lower limit λ_l (Fig. 3). The standard deviation of μ_{eff} within each acquisition window (represented by bars in Fig. 3) exhibits a substantial increase towards lower λ_l . However, there is no sharp transition from small to large standard deviation, which would indicate an obvious choice for the optimal acquisition window.

Figure 2: Effective absorption coefficient values $\mu_{\text{eff}}(z)$, calculated from (7) for human skin and different acquisition windows within the InSb detection band (3–5.3 μm). The lower wavelength limit λ_l runs from 3.0 μm to 5.1 μm , with the upper limit fixed at $\lambda_h = 5.3$ μm (note the values written in each graph). The solution is double-valued and in general varies with depth z .

Figure 3: Average effective absorption coefficient μ_{eff} for detection bands with lower limit λ_l from 3.0 μm to 5.1 μm , and upper limit fixed at $\lambda_h = 5.3$ μm . Bars represent standard deviation of μ_{eff} within the 100 μm thick superficial layer of skin.

Consequently, we perform a number of test reconstruction examples involving an imaginary temperature profile. We use a hyper-Gossip profile: $\Delta T(z,0) = \Delta T_0 \cdot \exp[-2(z-z_0)^4/w^4]$, with $\Delta T_0 = 1$ K, $z_0 = 300$ μm , $w = 100$ μm ,^{22,23} vaguely resembling a laser-heated PWS layer. With no noise added to the simulated PPTR signals, the convergence is very good, and the temperature profiles are typically reconstructed in 50–150 iteration steps – depending on the size and spread of the involved IR absorption coefficients.

Figure 4: Reconstructed temperature depth profiles (black lines) as obtained for acquisition bands of: (a) 3.0–5.3 μm , (b) 4.1–5.3 μm , and (c) monochromatic detection at 5.3 μm . The actual temperature profile is plotted for comparison (gray line).

Figure 4 presents reconstructed temperature depth profiles for the examples involving the full (3.0–5.3 μm) and narrowed spectral bands (4.1–5.3 μm), as well as monochromatic detection at 5.3 μm (Figs. 4 a, b, and c, respectively). The actual hyperGaussian temperature profile is plotted for comparison (gray line). It is evident that narrowing of the acquisition band considerably improves the accuracy of the reconstructed profile. On the other hand, even the monochromatic acquisition does not allow an arbitrarily accurate reconstruction of the actual temperature profile.

In Figure 5, we analyze the accuracy of the reconstructed profiles as a function of the lower wavelength limit λ_l . Figure 5a shows relative errors of the profile peak temperature (T_p), depth at half-maximum (z_1), and width at half-maximum (w). Quadratic norm of difference between the reconstructed and actual depth profile (presented relative to norm of the actual profile in Figure 5b) decreases almost monotonically with narrowing of the spectral acquisition window. From both graphs, a lower wavelength limit of $\lambda_l = 4.1$ μm seems to present an optimal compromise.

Figure 5: (a) Relative error of the reconstructed temperature depth profile as a function of the lower wavelength limit λ_l . The analyzed profile parameters are: depth at half-maximum (z_1), peak temperature (T_p), and width at half-maximum (w). (b) Quadratic norm of difference between the reconstructed and actual depth profile (relative to norm of the actual profile) for different lower wavelength limits λ_l . The upper wavelength limit is fixed at $\lambda_h = 5.3$ μm .

4.2. MCT detector, $\lambda = 10\text{--}12\ \mu\text{m}$

The same analysis as above is repeated next for a typical MCT (i.e., HgCdTe) photoconductive detector. Here, the upper wavelength limit (λ_h) is fixed at the peak of detector response, $\lambda_p = 12.0\ \mu\text{m}$ (e.g., Hamamatsu P3257-10).

Figure 6: Average effective absorption coefficient μ_{eff} for detection bands with lower limit λ_l from $10.0\ \mu\text{m}$ to $11.8\ \mu\text{m}$, and upper limit fixed at $\lambda_h = 12.0\ \mu\text{m}$. Standard deviations of μ_{eff} for each band are shown as bars.

Figure 7: (a) Relative error of the reconstructed profile parameters: peak temperature (T_p), depth at half-maximum (z_1), width of profile at the half of the peak temperature (w). (b) Quadratic norm of difference between the reconstructed and actual depth profile (relative to norm of the actual profile). Lower wavelength limit λ_l is varied, the upper limit is fixed at $\lambda_h = 12.0\ \mu\text{m}$.

Similarly to the previous example, the standard deviation of μ_{eff} diminishes substantially with narrowing of the spectral acquisition window (Fig. 6). Due to the strong absorption of water in the involved spectral band, however, all the μ_{eff} values are considerably higher. Analysis of the test reconstruction results is presented in Figure 7. By focusing on the depth of the upper edge of the heated layer (z_1), which is the most important parameter to extract in target clinical application, we conclude that the lower wavelength limit λ_l should not be smaller than $11.2\ \mu\text{m}$. This choice is supported also by the quadratic norm of the image error (Figure 7b).

4.3. MCT detector, $\lambda = 7\text{--}10\ \mu\text{m}$

Bearing in mind that the selection of acquisition window for PPTR depth profiling is governed by the spectral variation of $\mu_{\text{IR}}(\lambda)$, it seems plausible to consider using the MCT detector at $7\text{--}10\ \mu\text{m}$, where the μ_{IR} of skin is comparatively flat over a broad wavelength range (see Fig. 1). In the ensuing analysis, the upper wavelength limit (λ_h) is fixed at $\lambda_h = 10.0\ \mu\text{m}$.

Based on the results (Figs. 8, 9), we select $\lambda_l = 7.2\ \mu\text{m}$ for the optimal lower wavelength limit.

Figure 8: Average effective absorption coefficient μ_{eff} for detection bands with lower limit λ_l from 6.8 μm to 9.8 μm , and upper limit fixed at $\lambda_h = 10.0 \mu\text{m}$. Standard deviations of μ_{eff} for each band are shown as bars.

Figure 9: (a) Relative error of the reconstructed temperature profile parameters as a function of the lower wavelength limit λ_l (see Fig. 8 for explanation). (b) Quadratic norm of difference between the reconstructed and actual profile (relative to norm of the actual profile). The upper limit is fixed at $\lambda_h = 10.0 \mu\text{m}$.

5. PERFORMANCE COMPARISON

If we compare the PPTR profiling simulation results obtained with near-optimal spectral acquisition windows bands as determined for the three spectral bands under considerations above, the most accurate profiling is achieved using the MCT detector at 11.2–12 μm , despite the largest spread of $\mu_{\text{eff}}(z)$. This effect can be attributed to the highest value of μ_{eff} in this spectral band, which is known to reduce the ill-posedness of the involved inverse problem.⁸

However, for a trustable comparison of the three experimental designs, it is essential to take into account realistic SNR levels. These are estimated by assessing the lumped detector noise (N_d), and shot noise in radiometric signal (N_{sn}):²⁸

$$N_d = \frac{R}{D^*} \sqrt{A} \sqrt{B} \quad (8)$$

$$N_{sn} = \sqrt{2 B e_0 \varepsilon \alpha A \int R(\lambda) B_\lambda(T) d\lambda} \quad (9)$$

Here, R is detector sensitivity, D^* is detectivity (both at the wavelength of peak response), A is detector area, B is frequency bandwidth, e_0 is electron charge, and ε is skin emissivity. α describes collecting power of the optics and for typical setups approaches $1/(4F^2)$ where F is the “F-number” of the lens. After combining both noise contributions (by adding their squares), the result is compared with the amplitude of expected radiometric signal:

$$S = \varepsilon \alpha A \Delta T \int R(\lambda) B_{\lambda}'(T_0) d\lambda \quad (10)$$

The integrals in (9) and (10) run across the corresponding acquisition spectral band (λ_l – λ_h).

We use the following InSb detector characteristics as an example (Hamamatsu P5968-100): $R = 2 \text{ A/W}$, $D^* = 1.5 \times 10^{11} \text{ cm (Hz)}^{1/2}/\text{W}$, $A = 0.78 \text{ mm}^2$. By setting $B = 1 \text{ kHz}$, $\varepsilon = 1$, $F = 2$, and $\Delta T = 1 \text{ K}$, we obtain $\text{SNR} = 152$ for the optimal acquisition band of 4.1–5.3 μm . For an example MCT detector we take Hamamatsu P3257-10: $R = 25 \text{ A/W}$, $D^* = 4 \times 10^{10} \text{ cm (Hz)}^{1/2}/\text{W}$, $A = 1.0 \text{ mm}^2$. The obtained noise, signal, and SNR values for all acquisition bands under considerations are presented in Table 1.

Table 1: Optimal acquisition spectral band (λ_l – λ_h), effective absorption coefficient value μ_{eff} , and standard deviation of $\mu_{\text{eff}}(z)$ in top 0.1 mm of the skin (in parenthesis) for the three cases under consideration; followed by combined noise (N), signal level for $\Delta T = 10 \text{ K}$ (S), and resulting signal to noise ratio (SNR). In the last three columns are parameters of the profiles reconstructed from corresponding noisy PPTR signals: z_1 - depth at half-maximum (and relative error), T_p - peak temperature, $\|T-T'\|/\|T'\|$ - quadratic norm of the difference between the reconstructed and actual profile, relative to the norm of the actual profile. (Parameters of the actual profile: $z_1' = 0.223 \text{ mm}$, $T_p' = 1.000 \text{ K}$)

Detector, spectral band	μ_{eff} [mm^{-1}]	N [A]	S [A]	SNR	z_1 [mm]	T_p [K]	$\ T-T'\ /\ T'\ $
InSb, 4.1–5.3 μm	25.3 ± 1.1	3.79×10^{-11}	5.75×10^{-9}	152	0.226 (+1%)	1.050	0.094
MCT, 11.2–12 μm	123.5 ± 6.2	1.98×10^{-9}	1.77×10^{-7}	90	0.229 (+3%)	1.110	0.104
MCT, 7.2–10 μm	49.5 ± 0.4	1.98×10^{-9}	6.20×10^{-7}	313	0.219 (–2%)	1.015	0.061

Figure 10: Reconstructed temperature depth profiles (black lines) as obtained using the acquisition bands of: (a) 4.1–5.3 μm , (b) 11.2–12 μm , and (c) 7.2–10 μm ; with white noise added to the simulated PPTR signals according to SNR values in Table 1. The actual temperature profile is plotted for comparison (gray line).

Finally, we use the obtained SNR values to add a corresponding amount of normally distributed white noise to the multispectral PPTR signals computed from our test temperature profile, for each of the three “optimal” acquisition bands under considerations. Temperature depth profiles, reconstructed from these signals using the corresponding values μ_{eff} are presented in Figure 10 (black lines; the actual temperature profile is plotted in gray line for comparison). In the last three columns of Table 1, we compare the parameters of the reconstructed profiles: z_1 - depth at half-

maximum (and relative error), T_p - peak temperature (parameters of the actual profile are $z_1' = 0.223$ mm, $T_p' = 1.00$ K). A quadratic norm of the difference between the reconstructed and actual profile, relative to the norm of the actual profile ($\|T - T'\|/\|T'\|$) serves as objective criterion of the accuracy of the simulated PPTR depth profiling.

6. DISCUSSION

The presented simulation, involving multispectral composition of radiometric emission and realistic SNR values, confirms that PPTR depth profiling is a viable approach to noninvasive characterization of vascular lesions in human skin, providing a sufficient accuracy for real-time guidance of laser therapy on individual patient basis. Due to large spectral variation of mid-IR absorption coefficient in biological tissues, the acquisition band must be narrowed to ensure the validity of the customary approximation of a constant absorption coefficient value in signal analysis. Since this leads to reduced SNR in radiometric signals, a trade-off between the two effects must be carefully analyzed to determine the optimal spectral acquisition band.

In present study, we performed a number of PPTR measurement simulations involving a hyper-Gaussian temperature profile (imitating laser-heated vasculature in skin), taking into account also spectral sensitivity of commercial radiation detectors. By analyzing the accuracy of reconstructed profiles we determined near-optimal spectral acquisition windows within two popular (3–5 μm , 10–12 μm) and one potentially promising spectral band (7–10 μm) (see Table 1). In order to isolate the effect of spectral variation $\mu_{\text{eff}}(\lambda)$, these tests were performed without any noise in the simulated PPTR signals. A more accurate and objective determination of optimal acquisition bands might result from including realistic noise levels in this analysis, and we are currently performing such tests.

When comparing the simulation results between the near-optimal spectral acquisition windows, as determined for the three spectral bands under considerations above, the most accurate profiling was achieved using the MCT detector at 11.2–12 μm , despite the largest spread of $\mu_{\text{eff}}(z)$ in this example (see Table 1). We attribute this effect exclusively to the highest value of μ_{eff} in this spectral band, which reduces the ill-posedness of the reconstruction process.⁸ However, after augmenting the simulated signals by realistic levels of random noise, the best performance was predicted for the same detector used at 7.2–10 μm . This statement is based on comparison of errors in T_p (1.5% vs. 11%) and quadratic norm of the profile error (6.1% vs. 10.4%). This advantage of the 7.2–10 μm detection band evidently results from the largest SNR value (due to both the largest spectral width and highest values of $B_\lambda'(T_0)$), combined with the least amount of spread in $\mu_{\text{eff}}(z)$ (both absolute and relative). On the other hand, the latter is significantly larger than at 4.1–5.3 μm (49.5 vs. 25.3 mm^{-1}), where SNR is also relatively high, leading to better conditioned reconstruction problem.

We believe that the obtained conclusion can be generalized to photothermal measurements beyond PPTR depth profiling (e.g., frequency domain PTR, measurements of optical or thermal properties) in water- and protein-rich biological samples. On the other hand, the exact optimal acquisition windows, and perhaps even the selection of optimal radiation detector, might depend on application specifics, such as signal acquisition rate, detector size requirements, sample chemical composition, or even temperature profile amplitude and shape. To that end, we are developing more detailed simulations, involving digitized histology of PWS, Monte Carlo optical transport model, and finite difference thermal transport model, to analyze the expected system performance.²⁹

7. ACKNOWLEDGEMENTS

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