DEVELOPMENT OF FREQUENCY AND PHASE MODULATED THERMAL-WAVE METHODOLOGIES FOR MATERIALS NON-DESTRUCTIVE EVALUATION AND THERMOPHOTONIC IMAGING OF TURBID MEDIA

by

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Abstract

In frequency-domain photothermal radiometry (FD-PTR) a low-power intensity-modulated optical excitation generates thermal-wave field inside the sample and the subsequent infrared radiation from the sample is analyzed to detect material’s inhomogeneities. The non-contact nature of FD-PTR makes it very suitable for non-destructive evaluation of broad range of materials. Moreover, the methodology is based on intrinsic contrast of light absorption which can be used as a diagnostic tool for inspection of malignancy in biological tissues. Nevertheless, the bottom line is that the physics of heat diffusion allows for a highly damped and dispersive propagation of thermal-waves. As a result, the current FD-PTR modalities suffer from limited inspection depth and poor axial/depth resolution. The main objective of this thesis is to show that using alternative types of modulation schemes (such as linear frequency modulation and binary phase coding) and radar matched filter signal processing, one can obtain localized responses from inherently diffuse thermal wave fields. In this thesis, the photothermal responses of turbid, transparent, and opaque media to linear frequency modulated and binary phase coded excitations are analytically derived. Theoretical simulations suggest that matched-filtering in diffusion-wave
field acts as constructive interferometry, localizing the energy of the long-duty excitation under a narrow peak and allowing one to construct depth resolved images. The developed technique is the diffusion equivalent of optical coherence tomography and is named thermal coherence tomography. It was found that the narrow-band binary phase coded matched filtering yields optimal depth resolution, while the broad-band linear frequency modulation can be used to quantify material properties through the multi-parameter fitting of the experimental data to the developed theory. Thermophotonic detection of early dental caries is discussed in detail as a potential diagnostic application of the proposed methodologies. The performance of the diagnostic system is verified through a controlled demineralization protocol as well as in teeth with natural caries.
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Chapter 1

Introduction

This chapter introduces the photothermal effect and provides a short history on the developmental milestones of photothermal radiometry. Furthermore, the shortcomings and advantages of the technique are discussed.

1 Introduction

1.1 History and definition of photothermal phenomena

Photo-thermal science is a multidisciplinary field covering a wide range of techniques and applications. In this field, an intensity modulated excitation source (photo-) is absorbed within the sample and a modulated temperature field (thermo-) is then generated at the absorption site. Several detection techniques have been proposed for monitoring the modulated temperature field. Radiometric and acoustic detections are two of the most widely used detection methods in the field. In photothermal radiometry, the modulated thermal field, henceforth referred to as the thermal-wave field, is directly measured in a non-contact manner through the infrared radiation emanating from the sample. In photo-thermo-acoustics, or simply photoacoustics, the sample response is probed through detection of sound waves that are generated at the absorption site as a result of thermoelastic volume expansion. However, this detection method requires the presence of a coupling medium between the sample and the detector.

Historically, it was the photoacoustic method that was first discovered and used by Alexander Graham Bell in 1880 while he was inventing the photophone [1]. In Bell’s photophone, Figure 1.1, the ray of sunlight was modulated through the reflection by a mirror that was vibrating with a human voice. The detection was carried out photoacoustically at the receivers end through absorption of light by crystalline selenium cells followed by thermoelastic volume expansion and sound waves generation. However, Bell did not expand his idea of the photophone as he encountered many difficulties transmitting the modulated light through open air. The next milestone in the history of photothermal phenomena was achieved almost a century later in 1976 when Allan Rosenewaig and Allen Gersho rediscovered the photoacoustic effect and placed it in a correct theoretical foundation [2]. Their revolutionary
contribution led to the rapid development of the field and resulted in the invention of several other detection schemes, some of which are shown schematically in Figure 1.2.

Figure 1.1. A schematic representation of Bell’s Photophone.

Figure 1.2. Schematic representation of different detection schemes [3].

Among the proposed detection methods, radiometry (proposed by Nordal and Kanstad in 1979, Figure 1.3 [4]) is the only method that directly measures the temperature of the sample through an infrared detector and is based on the Stefan-Boltzmann law:
\[ W = \varepsilon \sigma T^4 \]  

(1.1)

where \( W \) is the total radiant emittance from the interrogated point, \( \sigma \) is the Stefan-Boltzmann constant \((5.67 \times 10^{-12} \text{ W cm}^{-2} \text{ K}^{-4})\), and \( \varepsilon \) is the emissivity. For a given change in surface temperature, \( \delta T \), there will be a change in radiant emittance of:

\[ \delta W = 4\varepsilon \sigma T^3 \delta T \]  

(1.2)

The use of infrared radiometry in photothermal measurements was demonstrated early in the development of the field both for spectroscopy [4] and for imaging [5].

Figure 1.3. Photothermal Radiometry setup proposed by Nordal and Kanstad for measuring the local amplitude and phase of the thermal-waves through a Lock-In Amplifier [4].

1.2 Photothermal radiometry methods

Conventional photothermal radiometry can be classified in many ways. The most appropriate type of classification is based on the type of optical excitation. In general, two kinds of excitation schemes can be used:

1. Pulse excitation (P)

2. Continuous-wave excitation (CW)
In the pulse method, a few-millisecond (or few-second for low thermal conductivity materials) high-power optical pulse is applied to the specimen and its subsequent thermal evolution (cooling) is recorded either with a single infrared detector (pulse photothermal radiometry, P-PTR) or an infrared camera (pulse thermography, PT). The qualitative temporal evolution of temperature is as follows: The temperature rises at first due to the applied pulse excitation; then it decays as heat is dissipated to the bulk of the specimen as a result of the heat diffusion process. However, the heat dissipation rate is reduced over subsurface defects and as a result the defective area appears as an area of higher temperature compared to the surrounding medium, Figure 1.4.(c).

Figure 1.4. (a) Experimental setup for pulsed thermography. (b) Thermal image after optical excitation showing shallow (A), intermediate (B), and deep (C) defects as well as a semi infinite point (R). PT cooling profiles (c) before and (d) after baseline reduction.

Several signal processing methods have been proposed to improve the performance of PT. The most common one is where the temperature decay profile of a semi infinite/intact area is subtracted from those of defective areas to magnify the time-delayed energy accumulation caused by the subsurface defects, Figure 1.4.(d). In the pulse method, thermal waves are
generated at a wide range of frequencies and the sample response is investigated in the time-domain and in a transient mode, making it impossible to study the contribution of individual frequency components. Moreover, the results are of an amplitude nature and therefore very prone to non-uniformities in applied excitation, sample emissivity, ambient reflections, and sample surface conditions. However, the pulse method seems to be the most popular method in industrial non-destructive testing applications mainly due to its quickness and phenomenological ease of interpretation.

The theoretical foundation of the CW mode was initially laid by Fourier (1822) [6] who showed that heat conduction problems in solids can be solved by expanding the applied excitation as a series of waves and later by Ångström (1863) [7] who proposed a “temperature-wave” method for the determination of thermal diffusivity in a long rod. As a result, this methodology is also referred to as Fourier-domain photothermal radiometry or FD-PTR. In this case, the excitation is in the form of a low-power, continuous, amplitude-modulated pattern at a given temporal frequency and the outcome is a spatially-damped thermal-wave field inside the sample at the applied excitation frequency. The signal processing of FD-PTR is not as straight forward as its pulse counterpart and requires quadrature demodulation of the signal to retrieve the amplitude and phase information of the thermal waves. FD-PTR has unique advantages over P-PTR; the most important advantage is that it can provide an additional contrast parameter (phase channel) which is emissivity normalized [8][9] and therefore insensitive to variations in the applied excitation power and/or sample surface conditions. Furthermore, the inspection depth (thermal diffusion length) can be controlled through the modulation frequency and thermal diffusion length (1.3).

\[
\mu = \sqrt{\frac{\alpha}{\pi f}} = \sqrt{\frac{2\alpha}{\omega}}
\]  

(1.3)

Here \(\alpha, f,\) and \(\omega\) are the sample thermal diffusivity, the optical excitation modulation frequency, and the optical excitation modulation angular frequency, respectively.

### 1.3 Applications of FD-PTR

Since the introduction of photothermal radiometry, a wide range of applications has been proposed for the technique. However, the majority of the applications are intended for non-
destructive evaluation and testing (NDE & T) of industrial materials. Gerhard Busse [10] is one of the pioneers in the field of photothermal radiometry NDT. Figure 1.5(a) shows one of his early studies in 1980 for measuring the thickness of an Aluminum wedge sample through measurement of the amplitude and phase of the transmitted thermal waves. The experimental setup consisted of an Ar-ion laser emitting at 488 nm. The laser beam was modulated between 15 and 30Hz and was then focused on the sample while the transmitted thermal waves were monitored by the infrared detector (Golay cell) from the back of the sample. The amplitude (A) line scans of Figure 1.5(b) show that the transmitted thermal wave amplitude decreases exponentially as the sample thickness increases. This trend highlights the spatially damped nature of thermal waves. Moreover, the larger the sample thickness, the larger is the phase shift due to the longer path travelled by the thermal waves.

![Figure 1.5](image)

**Figure 1.5.** (a) Arrangement of photothermal radiometry setup in transmission mode. (b) Variation of photothermal amplitude and phase with sample thickness at several modulation frequencies [10].

Using the same experimental setup, it was shown that the photothermal phase can reliably be used to detect subsurface holes parallel to the interrogated surface, while the photothermal
amplitude lacks the sensitivity to detect such subsurface boundaries, Figure 1.6. The improvement in sensitivity is due to the emissivity normalized nature of the phase channel.

Figure 1.6. Photothermal subsurface structure detection. Unlike the amplitude channel, photothermal phase has enough sensitivity to detect the 1mm-diameter holes [10].

Figure 1.7. Remote probing of coating thickness and defects. Arrows indicate the presence of subsurface defects [11].
Using a similar setup to Figure 1.5, Busse then expanded the NDT applications of FD-PTR for detection of polymer coating defects on an aluminum substrate in 1983 [11]. Again, he realized that the phase channel is more sensitive to subsurface variations in thermophysical properties than the amplitude channel (Figure 1.7). He also managed to simulate coating delamination by adding a thin layer of grease underneath the polymer coating and successfully detected it using the FD-PTR Phase measurements.

One drawback of FD-PTR in terms of NDE is the extremely long measurement time required to inspect the whole sample through a surface raster scan. This problem was solved in the late 1970’s to early 1990’s by the introduction of an infrared camera to the FD-PTR setup, Figure 1.8 [13][14][15][16]. The new system was called lock-in thermography (LIT) and was later investigated by Karpen et al. for monitoring fiber orientation in laminates [17] and Wu for defect characterization in veneered wood [18]. Shortly after, LIT found its way into a wide range of applications such as measuring thickness/density/porosity of ceramic coatings [19], inspection of aircraft structural components [20], measuring thermal diffusivity of materials [21][22], detection of air bubbles and inclusions in weld parts [22], and investigating shunt defects in solar cells [23].

1.4 Motivation for use in medical diagnosis

As mentioned earlier, the majority of the applications of FD-PTR/LIT are focused on industrial materials such as metals, ceramics, polymers, and solar cells. However, FD-PTR/LIT has two unique features that make it suitable for medical diagnostic applications.
First, unlike the pulse method, in FD-PTR/LIT low excitation power is used. As a result, the subsequent temperature rise in the sample is minimal. Wu et al. have reported that the typical optical power density used in this method is comparable to that of sunshine at noon in the summer [24]. This is appealing in medical applications where a sharp temperature gradient can cause irreversible changes in structure and chemical content.

Second, in PTR methods, the sample excitation is in the form of electromagnetic radiation (i.e., light) which provides an intrinsic contrast in non-opaque materials such as hard/soft biological tissues. That is, at proper excitation wavelengths light absorption is significantly different in malignant and benign tissues. As a result, photothermal signals (i.e., energy conversion from light to heat) are selectively generated in malignant areas, giving a very low background signal from the intact areas.

Figure 1.9 plots the absorption spectra of hemoglobin (Hb) and oxygenated-hemoglobin (HbO2) at visible and near infrared wavelengths. The spectra show that light absorption by hemoglobin can be used to reveal angiogenesis (i.e., a hallmark of cancer) [25][26][27]. Moreover, taking advantage of the relatively similar absorption of hemoglobin and oxygenated-hemoglobin at 584 nm and their significantly different absorption at 600 nm, one can calculate the blood oxygen saturation in biological tissues in a non-invasive manner [25]. As a result, FD-PTR/LIT has the potential to detect another important hallmark of cancer, hyper metabolism. Figure 1.10 shows how the enhanced light absorption at 532 nm can reveal the simulated blood vessels in a tissue mimicking phantom. Consequently, light absorption provides a unique diagnostic contrast that is not available in purely optical modalities based on light scattering. Nevertheless, biological samples are turbid media and do not strongly absorb the illuminating optical radiation. As a result, the photothermal signals obtained from these samples are generally poor in terms of signal-to-noise ratio. The other complication of photothermal signals obtained from turbid media is that the emitted thermal infrared radiation is physically governed by strongly coupled diffused-photon-density and thermal-wave processes, as opposed to conventional purely thermal-wave-generated infrared radiation from opaque materials. Consequently, the term thermophotonic response was coined for the photothermal response from turbid media in our lab to highlight the contribution of both optical and thermal fields. Early dental caries detection, which is one possible diagnostic application of the thermophotonic methodologies, is discussed in detail in this thesis.
Figure 1.9. Molar extinction coefficients of oxygenated and deoxygenated hemoglobin [25].

Figure 1.10. Lock-in thermography (a) amplitude and (b) phase images of simulated blood vessels obtained at 1 Hz [26].

1.5 Physical shortcoming of FD-PTR/LIT

Despite the unique advantages of FD-PTR/LIT, the physical phenomenon is still governed by the diffusive heat diffusion equation. Unlike the wave equation (a hyperbolic equation), the heat diffusion equation is a parabolic differential equation, which explains the heat propagation in the form of a diffusive process in the direction of the field gradient. As a result, the photothermal responses maintain a poor diffusive resolution. Moreover, proper identification of the relative depth of the defects involves performance of the experiments at several modulation frequencies (i.e., to inspect different depths based on (1.3)). A complete frequency scan is a very time consuming process and therefore limits the applications of the FD-PTR/LIT technologies.
Moreover, the resolution and maximum probing depth of thermal waves are both inversely proportional to the square root of the modulation frequency [28]. These relationships imply that one has to sacrifice the resolution to be able to inspect deeply into a sample. In other words, good resolution is obtained only at shallow inspection depths.

1.6 The objectives and outlook of the thesis

As mentioned earlier, the physics of heat diffusion only allows for a highly damped and dispersive propagation of thermal-waves. As a result, current FD-PTR modalities suffer from limited inspection depth and poor axial/depth resolution. That is, in FD-PTR one sees a superposition of thermal contributions from all the features within a thermal diffusion length. The main objective of this thesis is to show that using different types of modulation schemes (such as linear frequency modulation and binary phase coding) and radar matched filter signal processing, one can improve the inspection depth and axial/depth resolution and construct depth resolved images as opposed to the conventional depth integrated images of LIT. Such an achievement not only improves the system’s dynamic range and axial resolution, but also significantly decreases the measurement time required to decipher the relative depth information of defects by avoiding a conventional FD-PTR frequency scan. It will be shown that the improvements of the proposed methodology are not limited to industrial opaque materials and are also of great importance in diagnostic inspection of biological tissues/turbid media. In this thesis, to verify the theoretical predictions, the conventional LIT method is first applied to detection of early dental caries in a thermophotonic lock-in imaging setup. Then, the thermophotonic radar theory is introduced and finally the improvements in inspection depth and depth resolution are experimentally verified for both opaque and turbid (i.e. dental samples) media.

Thermophotonic lock-in imaging of early dental caries is novel by itself as to date only FD-PTR has been applied to this application. FD-PTR is a single point detection method which provides amplitude and phase values not accessible to dental practitioners. Moreover, the thermophysical properties of teeth are significantly variable among different persons, which in return alter the photothermal amplitude and phase baselines for distinguishing between intact and carious enamel. As a result, moving to an imaging system that can show the contrast between healthy and demineralized enamel in a single experiment makes the system more appealing for clinical
use. Moreover, the sensitivity of early caries detection is improved in a diagnostic imaging setup as one can quickly realize the extent and severity of the caries compared to the surrounding healthy areas in a single image.

Chapter 2 describes the instrumentation and signal processing methodologies used in the thesis. Moreover, in this chapter the ideas behind the use of matched filtering and pulse compression techniques are described.

Since thermophotonic lock-in imaging is performed on dental samples, chapter 3 provides the dental background required for proper understanding of experiments carried out on these samples. Thermophotonic lock-in imaging of early caries is discussed in chapter 4.

Chapter 5 presents the thermal-wave radar theory for opaque and transparent media using linear frequency modulation. The unique characteristics of the proposed method are discussed through theoretical simulations and experiments in a single detector setup for both opaque and transparent media.

Finally, chapter 6 moves one step further and incorporates the thermal-wave radar theory in a fully integrated imaging system. The improvement in axial/depth resolution and the depth resolved nature of the thermal-wave radar method are experimentally verified in the imaging setup. Furthermore, the thermal-wave response of turbid media to binary phase coded optical excitation is derived analytically and verified experimentally. It is shown how the depth resolved nature of the thermal-wave radar method can be used to construct iso-depth thermal coherence tomography images.

Chapter 7 brings together the conclusions of the research presented in this thesis.
Chapter 2
Materials and instrumentation

In this chapter, the details of the instrumentation and experimental setups are discussed. The chapter opens with a conventional FD-PTR system and explains how the idea can be expanded to an integrated lock-in imaging system. Afterwards, the idea behind radar matched filtering and pulse compression techniques are explained and it is shown how the lock-in imaging system can be modified to allow for novel frequency/phase modulated thermal-wave techniques. Finally, the preparation and characteristics of a series of standard/non-biological samples are discussed.

2 Materials and instrumentation

2.1 FD-PTR system

Figure 2.1, schematically shows the experimental setup used for the FD-PTR measurements carried out in this thesis. The setup includes an 830 nm laser (power of 100 mW with beam size of 325.9 ± 25.5 µm).

![Schematic experimental setup of the FD-PTR system](image)

Figure 2.1. Schematic experimental setup of the FD-PTR system [29].

Briefly, the intensity of the laser beam is modulated at a given frequency by the internal function generator of the lock-in amplifier. Simultaneously, the infrared response from the sample is
captured by the mercury cadmium telluride (MCT) detector, amplified by the pre-amplifier, and analyzed by the lock-in amplifier to reveal the amplitude and phase of the modulated infrared response from the sample. The custom-made program designed in a LabView environment is able to run measurements both at a fixed frequency and in a range of frequencies (i.e., a frequency scan). A four-axis (XYZ and angle) positioning system allows accurate positioning of the sample at the focal point of the off-axis mirrors. The transfer function of the system was obtained by performing a frequency scan on a standard opaque sample with known thermophysical properties, fitting the experimental data to the theory, and calculating the amplitude and phase compensation gains. The calculated transfer function of the system is applied to the frequency scan data obtained from the samples to assure consistency between the experiments and theory.

The signal processing algorithm of a lock-in amplifier is relatively simple and straightforward. The device mixes the input signal with the in-phase and quadrature reference signals and, after proper weighting and low-pass filtering, calculates the amplitude and phase of the signal based on the mathematical derivation (2.1), Figure 2.2.

\[
A = \sqrt{V_1^2 + V_Q^2} \quad \text{and} \quad \phi = \tan^{-1}\left(\frac{V_Q}{V_1}\right)
\]

**Figure 2.2.** Signal processing block diagram of FD-PTR.

\[
\begin{align*}
\begin{cases}
\sin(\omega_o t) \times A \sin(\omega_o t + \phi) \\
\sin(\omega_o t + 90) \times A \sin(\omega_o t + \phi)
\end{cases}
\rightarrow
\begin{cases}
\frac{A}{2} [\cos(\phi) - \cos(2\omega_o t + \phi)] \\
\frac{A}{2} [\sin(\phi) - \cos(2\omega_o t + \phi + 90)]
\end{cases}
\rightarrow
\begin{cases}
X = \frac{A}{\sqrt{2}} \cos(\phi) \\
Y = \frac{A}{\sqrt{2}} \sin(\phi)
\end{cases}
\rightarrow
\begin{cases}
A = \sqrt{X^2 + Y^2} \\
\phi = \arctan\left(\frac{Y}{X}\right)
\end{cases}
\end{align*}
\]
where \( \sin(\omega t) \) and \( \sin(\omega t + 90) \) represent the in-phase and quadrature reference signals and \( A\sin(\omega t + \phi) \) is the captured infrared signal with amplitude \( A \) and phase \( \phi \).

### 2.2 Thermophotonic system

Figure 2.3 shows the components of the thermophotonic setup as well as an illustrative signal processing algorithm of thermophotonic lock-in imaging [30].

![Image of thermophotonic setup](image)

**Figure 2.3.** (a) A schematic of the thermophotonic setup along with (b) the thermophotonic Lock-In signal processing algorithm.

The laser source is a continuous-wave fiber-coupled 808 nm near-infrared (NIR) laser diode (JENOPTIK, Germany) with two integrated thermoelectric coolers. An appropriate laser driver (Thorlabs, LDC 3065) and a thermoelectric controller (Coherent Model 6060) are used to maintain an average optical intensity of 2.04 W/cm² on the sample surface. The distance between the laser-power delivering optical fiber and the sample is adjusted to get a 20-mm diameter beam size on the interrogated surface of the sample, while completely covering the camera’s field of view.

In the experimental setup, the LEGO-mounted sample is placed on a rotation stage (precision 0.5 degree) mounted on a three-axis XYZ translation stage (precision 10 \( \mu \)m). Using these four degrees of freedom, the position of the sample with respect to the camera is adjusted to yield a focused (sharp) image of the interrogated surface on the analog video output of the camera and
the XYZ and angular coordinates of the sample are recorded to ensure the identical repositioning of the sample during repeated measurements. Our camera (Cedip Titanium 520M, France) is a state-of-the-art focal plane array (FPA) infrared camera with a spectral range of 3.6 – 5.1 μm and maximum frame rate of 175 Hz at full frame (370 Hz in the sub-window mode). The camera’s detector array consists of 320x256 indium antimonide (InSb) elements with element size of 30x30 μm². Using a custom made extension tube and a 50-mm-focal-length objective lens (Cedip MW50 L0106), a magnification of one is obtained from the interrogated surface of the sample. The camera generates an integration time (IT) pulse train output at the frame rate frequency. The duration of each pulse in this pulse train is equal to the FPA integration time (1 ms). The multi-data acquisition board (National Instrument NI-6229 BNC) receives this pulse train and synchronously generates three analog outputs: flag pulse train, in-phase reference signal, and quadrature reference signal and sends them to the camera as its external trigger, lock-in signal I, and lock-in signal II inputs, respectively. The purpose of having a flag pulse train is to be able to detect the beginning of each modulation cycle for averaging purposes, Figure 2.3(b).

The frequency of the reference signals (in-phase and quadrature) is set to the laser modulation frequency and the in-phase signal leads the quadrature signal by 90°. The in-phase reference signal is used to modulate the intensity of the laser beam in either a single frequency mode (lock-in imaging) or a chirp/binary phase coded mode (thermal-wave radar imaging). The modulated laser beam illuminates the sample through the optical fiber. The camera frame headers contain the flag pulse train status (high or low) and the reference signals values (between -1 and 1) at the instance that the frame is captured. The designed data acquisition/signal processing program runs at the highest frame rate of the camera and captures an image sequence that corresponds to two modulation cycles. Then, it extracts the header information from the image sequence and finds the beginning of a modulation cycle using the flag pulse train information available in the image header. The image sequence and the reference signal values of one complete modulation cycle (starting from a zero phase value) are exported to a buffer. Depending on the signal-to-noise ratio (SNR) of the signal obtained from the sample, the above steps are repeated several times and the image sequence and its reference signals values are averaged in the buffer to suppress the stochastic noise. Adopting such an averaging methodology, instead of saving the data on the hard drive and averaging the signals after acquisition, allows for significantly less
expensive/complicated computing hardware and much faster imaging technology. Finally, based on the signal processing method chosen, the averaged image sequence is processed and the final images are calculated.

In the case of LIT, the averaged image sequence is weighted by the two reference signals (i.e. each frame is a 2-D matrix of integer numbers and is multiplied by its corresponding reference signals values, Figure 2.3(b)) and then the weighted frames are summed to low-pass filter the signals and obtain the noise reduced in-phase \( S^0 \) and quadrature \( S^{90} \) images. The amplitude and phase images are then calculated by applying (2.2) to each pixel:

\[
A = \sqrt{(S^0)^2 + (S^{90})^2} \quad \text{and} \quad \phi = \arctan\left(\frac{S^{90}}{S^0}\right)
\]  

(2.2)

The signal processing steps usually take a few seconds. Therefore, the experiment time strongly depends on the duration of the modulation cycle (i.e. lock-in frequency or chirp/binary phase code duration) as well as the SNR of the acquired signals (i.e. number of required averaging). The higher the laser intensity, the less is the averaging required to get an acceptable SNR.

2.3 Synchronous undersampling

One of the most important constraints of the current infrared cameras is their low frame rate, making them incapable of monitoring high frequency phenomena. In theory, to properly sample a waveform the sampling rate should be at least twice the highest frequency available in the waveform (i.e. at least two samples in each modulation cycle) to satisfy the Nyquist criteria and avoid aliasing artifacts [31]. However, for amplitude and phase retrieval at least four samples per cycle is required [32]. Moreover, the higher the number of samples in each cycle (over sampling), the higher is the SNR and accuracy of the amplitude and phase measurement. As a result, considering the highest frame rate of our infrared camera in the full window mode (i.e. 175Hz) and the SNR of the dental samples the highest modulation frequency for which acceptable amplitude and phase images could be formed was found to be 10Hz. Therefore, synchronous undersampling was integrated to our LabView program to allow for proper sampling of frequencies higher than 10Hz. The idea behind this sampling method is to acquire \( n \) samples in each modulation cycle of the signal. However, these \( n \) samples are collected from \( n \) consecutive cycles instead of one. Figure 2.4 graphically explains how 12 consecutive cycles can
be used to sample one modulation cycle with 12 points. Equation (2.3) shows the relationship between the modulation frequency, $f$, and the synchronous undersampling frequency, $f_s$ [32].

$$f_s = f(1 - \frac{1}{n})$$  \hspace{1cm} (2.3)

As a result, if one intends to sample 18 points per cycle with a camera having a maximum frame rate of 180Hz, using conventional sampling the highest frequency that can be sampled is 10Hz but with synchronous undersampling it can be as high as 190Hz. Importantly, the application of this sampling method requires accurate synchronization of the experimental setup.

![Figure 2.4. Synchronous undersampling of a high frequency waveform using a low sampling rate. One modulation cycle is sampled out of each 12 consecutive cycles [32].](image)

Our experimental setup is designed to use synchronous undersampling for frequencies higher than 10Hz with $n = 36$.

### 2.4 Matched filtering and pulse compression techniques

In radar science, the detection of a signal in a noisy channel has always been a challenge. As a result, a surge of research has been carried out to introduce signal recognition methodologies which implement high SNR values. In the mid 1940s Van Vleck and Middleton [33] realized that using a special kind of linear filters (known as matched filters), one can detect a known signal waveform in a highly noisy channel. Matched filters were designed based on the known
waveform, which maximized the output when a delayed replica of the waveform was passed through the channel [34][35]. The well-known cross-correlation (CC) technique is a special case of matched filters which is extensively used for signal detection, especially in continuous wave radars, Figure 2.5.

![Diagram of cross-correlation and matched filtering](image)

**Figure 2.5.** The cross-correlation operator yields a better SNR and resolution by localizing the energy of the applied long-duty excitation under a single peak.

The cross-correlation of a signal with its delayed replica (i.e. a backscattered echo) results in a “sinc” type waveform, \( Sinc(x) = \frac{\sin(\pi x)}{\pi x} \), with the main peak centered on the corresponding delay time, \( \tau_p \). In the acoustic or other propagating wave cases, the distance between the object and receiver is directly proportional to the calculated peak delay time, \( \tau_p \), and can be estimated using the wave speed in the medium. The height of the CC main peak determines the SNR and the axial/depth resolution is controlled by the width of the peak. A pulse compression technique can make the width of the peak narrower (better axial resolution) and its height larger (better SNR). Pulse compression techniques do not change the energy of the input signal; however, they encode the signal such that after matched filtering the signal energy is compressed under the main lobe of the matched-filter output (i.e. the sinc function). Compared to pulsed radars, frequency modulated CW radars transmit higher energy within their long duty cycle at much lower intensity and simultaneously attain the resolution of high-intensity pulsed radars by means of the matched filter method and pulse compression techniques. It can be shown that matched filters are the optimal linear filters for maximizing the SNR in the presence of stochastic noise [35].
CW radars use different types of pulse compression methods such as linear frequency modulation (LFM) and binary phase coding (BPC) in order to augment both the SNR and range resolution [36]. A linear frequency (chirp) modulated waveform such as a linear frequency modulated laser beam, schematically shown in Figure 2.6(a), can be mathematically defined as:

\[ f(t) = Q_0 \sin[2\pi f_1(t) t] \]  

(2.4)

where

\[ f_1(t) = f_s + Bt \]  

(2.5)

\[ B = \frac{(f_e - f_s)}{2T} \]  

(2.6)

\[ \varphi(t) = \frac{\partial [f_1(t) t]}{\partial t} = f_s + 2Bt = f_s + \frac{(f_e - f_s)}{T} t \]  

(2.7)

here \( Q_0, f_1(t), f_s, f_e, B, T, \) and \( \varphi(t) \) are modulated waveform intensity, frequency modulation function (chirp), chirp start frequency, chirp end frequency, chirp sweep rate, chirp time, and waveform instantaneous phase, respectively.

The energy within a certain frequency range of the spectrum can be obtained using the energy spectral density (ESD) function. This function has the unit of energy/hertz and can be calculated as:

\[ \Phi(\omega) = \frac{S(\omega) \cdot S^*(\omega)}{2\pi} \]  

(2.8)

where \( S(\omega) = \mathcal{F}\{s(t)\} \) is the spectrum of the received signal. Calculating the ESD of a LFM waveform, it can be seen that the energy is uniformly distributed within the chirp modulation bandwidth (from \( f_s \) to \( f_e \)), yielding a relatively wide band spectrum as shown in Figure 2.6(b).
Figure 2.6. A simulated (a) LFM (chirp) waveform and its (b) ESD. Chirp parameters: \( f_s = 0.1 \) Hz, \( f_e = 9.9 \) Hz, and \( T = 1.4 \) s.

The BPC signal, \( f(t) \), consists of a single frequency carrier, \( C(t) \), and a binary coded envelope, \( E(t) \). The signal is formed either by multiplying these components in the time-domain or alternatively by convolving their spectra in the frequency-domain, Figure 2.7. In general, an arbitrary binary sequence can be defined as \( a_j = [a_1, a_2, ..., a_N] \). In the special case where the temporal length of each code element equals the period of the carrier \( (T_o) \), the binary coded envelope can be modeled as a series of rectangular pulses of width \( T_o \) and height \( a_j \) shifted in the time axis by \( (j - 0.5)T_o \):

\[
E(t) = \sum_{j=1}^{N} a_j \text{rect} \left( \frac{t - (j - 0.5)T_o}{T_o} \right) = \sum_{j=1}^{N} a_j \text{rect} \left( \frac{t}{T_o} - (j - 0.5) \right) \quad (2.9)
\]

Therefore, starting with the Fourier transform of a pulse (2.10), the analytical spectrum of the binary coded envelope (2.12) can be obtained using the time-frequency shifting property of the Fourier transform (2.11).

\[
\mathcal{F} \left\{ \text{rect} \left( \frac{t}{T_o} \right) \right\} = \sqrt{2\pi} \cdot \text{Sinc} \left( \frac{\omega}{\omega_o} \right) \quad (2.10)
\]

\[
\mathcal{F} \{ h(t - t_o) \} = H(\omega) \exp(-i\omega t_o) \quad \text{where} \quad H(\omega) = \mathcal{F} \{ h(t) \} \quad (2.11)
\]
\[ \Im \{ E(t) \} = E(\omega) = \frac{\sqrt{2\pi}}{\omega_o} \text{Sinc} \left( \frac{\omega}{\omega_o} \right) \sum_{j=1}^{N} a_j \exp \left[ -i\omega \tau_j (j - 0.5) \right] \]  

(2.12)

where \( \omega_o \) and \( i \) are the carrier angular frequency and the imaginary unit, respectively.

Finally, based on Figure 2.7, the analytical spectrum of the binary phase coded signal (2.15) can be calculated through the convolution of (2.12) and the spectrum of the single frequency carrier waveform (2.13) using identity (2.14).

\[ C(\omega) = \sqrt{2\pi} \left( \frac{\delta(\omega - \omega_o) - \delta(\omega + \omega_o)}{2i} \right) \]  

(2.13)

\[ f(t) * \delta(t - \alpha) = f(t - \alpha) \]  

(2.14)

\[ \Im \{ f(t) \} = F(\omega) = \sum_{j=1}^{N} \left[ a_j \text{Sinc} \left( \frac{\omega - \omega_o}{\omega_o} \right) \exp \left[ -i\omega \tau_j (j - 0.5)(\omega - \omega_o) \right] \right] \]  

(2.15)

Equation (2.15) suggests that the spectrum of the BPC signal consists of a series of weighted “sinc” functions yielding a narrow-band waveform with most of its energy located at the carrier frequency, as shown in Figure 2.7. A comprehensive study on BPC can be found in [37]. The codes introduced in reference [38] have been used in our experiments.
2.5 Matched-filter signal processing

The cross-correlation of signals \( r(t) \) and \( s(t) \), i.e. reference and response signals respectively, is defined in the time-domain by (2.16). The operator basically delays the matched filter, \( r(t) \), in the time domain and registers the similarity of the input signal, \( s(t) \), to the delayed filter. The time-domain definition of CC is very computation-intensive as it uses integration and multiplication, but fortunately the operator can be expressed in the frequency domain as a simple multiplication of the signal spectra, (2.17).

\[
(s \times r)(\tau) = \varepsilon \int_{-\infty}^{\infty} s(t) r(t+\tau) dt
\]

(2.16)

\[
(s \times r)(\tau) = \varepsilon \mathcal{F}^{-1}\{S(\omega)R(\omega)^*\}
\]

(2.17)

\[
\theta_{cc}(\tau) = \frac{\varepsilon \mathcal{F}^{-1}\{R(\omega)^*\}}{\varepsilon \mathcal{F}^{-1}\{-i \text{sgn}(\omega)\}}
\]

(2.18)

\[
S(\omega) = \mathcal{F}\{s(t)\} \quad \text{and} \quad R(\omega) = \mathcal{F}\{r(t)\}
\]

Here \( \varepsilon \), *, and \( \mathcal{F} \) denote the sample emissivity, complex conjugate operator, and Fourier transform operator, respectively. In (2.18), \( \text{sgn}(\omega) \) is the signum function; the expression inside square brackets in the denominator is the Fourier transform of the quadrature reference signal. It should be noted that the emissivity is normalized in the CC phase channel.

The block diagram of Figure 2.8 shows how the CC signal and phase can experimentally be calculated using the sub-routines available in either Matlab or LabView.
The CC processing can provide three distinct contrast parameters:

1- The height of the CC peak is referred to as the amplitude channel. In a thermophotonic system, the amplitude channel shows the strength of the optical absorption in a sample. As a result, a strong absorber deep in the sample may yield higher amplitude than a weak absorber close to the interrogated surface.

2- The location of the CC peak on the delay time axis, $\tau_p$, is referred to as the peak delay time and represents the delay (i.e. phase lag) of the received response with respect to the applied excitation regardless of the severity of optical absorption. As a result, this channel is linked only to the depth of the absorber; however, the channel is sensitive to the input signal SNR and can be inaccurate when the SNR is low.

3- The phase of the CC signal can be calculated based on (2.18). Similar to CC peak delay time, the CC phase is an emissivity normalized quantity. Moreover, it is less sensitive to noise compared to the peak delay time channel.

2.6 Thermal-wave radar system

Before moving to the expensive thermophotonic radar imaging system, the developed matched filter radar theory was verified in a single detector system. Figure 2.9 illustrates the experimental setup for the thermal-wave radar (TWR) system schematically. A 1064 nm CW laser source
(IPG Photonics) is frequency modulated by an acousto-optic modulator (AOM)(Neos Technologies). Modulation is performed either at a single frequency or in a linear frequency sweep. The sample is fixed on a 2-D translation stage with its surface located at one focal point of the elliptic mirror. The black-body infrared radiation from the sample surface is observed at the other focal point of the mirror by a MCT infrared detector (Judson technologies). The MCT signal is first amplified by the pre-amplifier (PA-101, Judson technologies) and then digitized by the high speed dual channel analog to digital converter (PCI-5122, National Instruments). The synchronization of the data acquisition software and the modulation systems is carried out using an external delay generator (DG535, Stanford Research systems). The delay generator simultaneously triggers the frequency modulation system and the LabView signal recording/processing sub-routines.

Figure 2.9. Experimental setup of the thermal-wave radar system embodied in a photothermal radiometric detection scheme.

2.7 Thermophotonic radar imaging system

In the case of thermophotonic radar imaging, the experimental setup, data acquisition method, and the averaging methodology are the same as those discussed for the thermophotonic lock-in imaging (section 2.2). However, the laser excitation pattern is different from that of the
thermophotonic lock-in imaging (i.e., LFM/BPC instead of single frequency). Accordingly, matched filtering is used in the signal processing step instead of the quadrature demodulation used in thermophotonic lock-in imaging.

2.8 Multi-parameter fitting

Earlier research in our lab on FD-PTR measurements introduced a robust and complicated fitting program [39] for multi-parameter fitting of experimental data to a theory [40]. The program inputs are the theoretical model, experimental data, transfer function of the system, and fitting range of the “N” physical parameters to be fitted. The program uses a non-linear optimization algorithm called the downhill simplex method to find the best combination of “N” variables which minimizes a residual function (i.e., theory – experiment). The robustness of the algorithm, i.e. its independence of the initial estimation of parameters, is an important parameter in defining a unique solution in the multi-parameter fitting procedure, which has been evaluated elsewhere [41]. To show the possibility of extracting quantitative data from the thermophotonic imaging data, the multi-parameter fitting program was modified with the developed theory to obtain thermophysical properties of samples.

2.9 Classic samples

A series of standard/non-biological samples were prepared to evaluate/compare several features of the lock-in and thermal-wave radar methodologies.

2.9.1 Opaque sample

An AISI 1010 steel block with six 5-mm diameter blind holes, located at various depths from the surface, was prepared to simulate an opaque sample with defects at several depths. The holes were drilled to give wall thicknesses of 100, 400, 600, 1000, 1500, and 3000 µm, labeled H1 to H6 respectively, as shown in Figure 2.10.
Figure 2.10. AISI 1010 sample with six blind holes. Dimensions are in mm.

2.9.2 Perfect absorber in optically transparent medium

To simulate perfect absorbers in a non-absorbing and non-scattering medium, microscope slips with several thicknesses were painted black on the back surface prior to investigation at the front surface. Three of the glass samples were Fisher Scientific borosilicate microscope cover slips with thicknesses of 0.12, 0.15, and 0.2 mm. The 1-mm-thick glass sample used in our research was a commercial microscope cover slip with unknown physical and optical properties.

2.9.3 Absorbers in optically transparent medium

Fisher Scientific borosilicate microscope cover slips with a thickness of 120 μm were used to simulate absorbers with different absorption coefficients, but at the same depth, in an optically transparent medium. The glass was covered with commercial green (left) and black (right) paints with no paint applied to the center part to form a three-strip pattern, Figure 2.11(a). Six additional microscope cover slips were put on the painted slip to simulate two absorbers 720 μm below the surface with different absorption coefficients. No glue was used; the edges were clamped to improve the physical contact between the slips.
Figure 2.11. (a) Sampled used to simulate absorbers with different absorption coefficients at the same depth. (b) An exploded view of the sample used to simulate overlaying absorbers.

Figure 2.11(b) shows an exploded view of a cross-shaped sample made of two strip absorbers. The deeper strip is a black plastic sample, completely absorbing the optical flux, while the shallower strip is a partially absorbing phantom. All other components are transparent polyvinyl chloride-plastisol (PVCP). When viewed from the top, the cross sample covers all possible combinations: absorbers at two different depths (the end sections of the absorbing strips) and two absorbers on top of each other (the center of the sample).

2.9.4 Perfect absorbers at several depths in an optically scattering medium

A black plastic step wedge sample was placed inside a scattering medium (PVCP) with added titanium dioxide (TiO$_2$) powder for scattering [42] such that the first step was located approximately 1 mm below the phantom surface as shown in Figure 2.12. The width and height of each step were approximately 1mm and 200 $\mu$m, respectively.
Figure 2.12 (a) Black plastic step wedge sample (b) inside a scattering medium. (c) Schematic cross-section of the sample.
Chapter 3

Demineralization and early dental caries

The intention of this chapter is to provide the reader with sufficient dental background to follow the discussions made in the following chapters on the diagnostic imaging of early dental caries. It should be noted that thermophotonic imaging is not limited to early dental caries diagnosis and may be applied to any opaque/turbid medium. After discussing the dental background, the chapter closes with dental sample preparation and the artificial demineralization protocol used to simulate early dental caries.

3 Demineralization and early dental caries

Dental caries is a chronic disease identified as the leading cause of tooth loss among children and adult populations. The precursor of the disease is a minute amount of mineral loss (demineralization) from the enamel surface as a result of decomposition of hydroxyapatite crystals in the acidic environment of dental plaque [43]. Given enough time, such early caries turns into a cavity which requires surgical intervention. However, if the caries is detected early enough, not only can it be stopped (i.e. arrested) from progressing deeper into enamel, but also it can be healed (i.e. remineralized) using, for instance, oral hygiene counseling or fluoride therapy [44][45]. Unfortunately, conventional clinical diagnostic modalities, such as x-ray radiography and visual/tactile inspections, lack sufficient sensitivity and/or specificity to detect early caries [46]. There are many benefits in detecting dental carious lesions in their early stages of progression. These include:

1. Increased potential to remineralize the demineralized, non-cavitated tooth surfaces.
2. Decreased risk of progression to the cavitated stage.
3. Reduced probability of tooth sensitivity associated with deeper lesions.
5. Preservation of the natural esthetic appearance of tooth enamel.
6. Reduced treatment cost associated with premature and unnecessary surgical interventions.
3.1 Structure of human teeth

The human tooth is made up of four types of tissues: enamel, dentin, pulp, and cementum. Enamel is the hardest and most mineralized tissue in the body. Ninety-five percent (by weight) of enamel consists of mineral, with water and organic material composing the rest. Normal and sound enamel consists of hydroxyapatite crystals (Ca$_5$(PO$_4$)$_3$OH) so tightly packed that the enamel has a glass-like appearance; the enamel is translucent (refractive index of 1.62). The yellow-white color of teeth is therefore the result of dentin shining through the translucent enamel cover [47]. The enamel crystals are packed in a repeating arrangement which forms the enamel prisms. Even though crystal packing is very tight at the microscopic level, each crystal is separated from its neighbors by tiny inter-crystalline spaces that are filled with water and organic materials. The inter-crystalline spaces together form a fine network of diffusion pathways which are often referred to as micro-pores, or simply pores, in the enamel. The enamel microstructure is significantly inhomogeneous, as a result of constant interaction of enamel crystals with the oral environment through micro-pores. Enamel varies in thickness over the surface of the tooth and is often thickest at the cusp (occlusal eminence on a tooth), up to 2.5 mm, and thinnest at its border at the cementoenamel junction (CEJ).

Figure 3.1. Section of human molar [48].
Dentin is the hard tissue between enamel and the pulp, secreted by the dental pulp. In general, dentin is less mineralized than enamel and therefore softer than enamel. Dentin has microscopic channels, called dentinal tubules, which radiate outward through the dentin from the pulp cavity to the exterior cementum or enamel border. The dental pulp is the central part of the tooth filled with soft tissue. This tissue contains blood vessels and nerves that enter the tooth from a hole at the apex of the root. Along the border between the dentin and the pulp are odontoblasts, which initiate the formation of dentin. Cementum is a specialized bony substance covering the root of a tooth. Its coloration is yellowish and it is softer than either dentin or enamel [47].

3.2 Dental plaque

Glycoprotein from saliva forms an amorphous organic film (or biofilm) on the surface of enamel called “pellicle”. Pellicle is tenacious and can attract and help anchor specific types of bacteria to the tooth surface. This adherent deposit of bacteria and their product is called a “dental plaque”. There are local variations in the bacteria content and type in dental plaques, which explains why some sites experience a high caries activity while neighboring sites in the same mouth are free from caries. Some bacteria in the dental plaque are able to produce acid rapidly (1-3 minutes) from fermentable carbohydrates (i.e., the sugars sucrose and glucose) and drop the plaque pH to below 5; furthermore, they synthesize a gelatinous sticky polymer of glucose that thickens the plaque layer and prevents saliva from neutralizing the plaque pH. It takes the acidic plaque 30-60 minutes to return to its normal pH of about 7. Bacterial plaque is an essential precursor of caries and therefore sites on the tooth surface which favor plaque formation are particularly prone to caries. It is important to regard caries as an alternating process of destruction and repair. When the destructive forces outweigh the reparative powers of the saliva the process will progress (demineralization) and conversely if the reparative forces outweigh the destructive forces, the process will arrest or even reverse (remineralisation), provided it is detected in its early stages [47].

3.3 Demineralization and remineralization of tooth

As mentioned earlier, the micro-pores in the enamel are filled with water and as a result of their constant interaction with hydroxyapatite crystals of enamel, the water in the micro-pores is saturated with respect to hydroxyapatite, i.e. the mineral is in equilibrium with the ions in the solution [43]:

[47]
\[ \text{Ca}_3(\text{PO}_4)_5\text{OH} \leftrightarrow 5\text{Ca}^{2+} + 3\text{PO}_4^{3-} + \text{OH}^- \quad \text{K}_{\text{SP(HA)}} = 7.41 \times 10^{-60} \quad [\text{mol/l}] \]  

(3.1)

where \( K_{\text{SP(HA)}} \) denotes the solubility product of hydroxyapatite at 37°C. The solubility of hydroxyapatite and other calcium phosphates is greatly affected by the pH of the water in which it dissolves. As indicated by (3.1), when \( \text{PO}_4^{3-} \) and \( \text{OH}^- \) accumulate in solution, together with calcium ions, dissolution of hydroxyapatite slows and stops as the solution becomes saturated. If acid is added, \( \text{PO}_4^{3-} \) ions and \( \text{OH}^- \) ions combine with \( \text{H}^+ \) to form \( \text{HPO}_4^{2-} \) ions and \( \text{H}_2\text{O} \), respectively, thereby removing a proportion of \( \text{PO}_4^{3-} \) and \( \text{OH}^- \) ions from solution:

\[
\begin{align*}
\text{Ca}_3(\text{PO}_4)_5\text{OH} & \leftrightarrow 5\text{Ca}^{2+} + 3\text{PO}_4^{3-} + \text{OH}^- \\
& \downarrow \text{H}^+ \quad \downarrow \text{H}^+ \\
& \text{HPO}_4^{2-} \quad \text{H}_2\text{O} \\
& \downarrow \text{H}^+ \\
& \text{H}_2\text{PO}_4^- 
\end{align*}
\]

(3.2)

In this case the ion activity product of hydroxyapatite \( (\text{IAP}_{\text{HA}}) \) decreases, the solution is then said to be unsaturated \( (\text{IAP}_{\text{HA}} < K_{\text{SP(HA)}}) \) and more hydroxyapatite dissolves until saturation is re-established. Therefore, mineral is removed from the surface of the hydroxyapatite crystals to compensate for the lack of ions in the solution and the inter-crystalline spaces enlarge and, consequently, the tissue becomes more porous.

For remineralisation to occur, the solution must be supersaturated with respect to hydroxyapatite \( (\text{IAP}_{\text{HA}} > K_{\text{SP(HA)}}) \). So, it is necessary to add the constituent ions to the solution either by adding calcium and/or phosphate separately or by raising the pH (which increases \( \text{PO}_4^{3-} \) and \( \text{OH}^- \) concentrations) or both.

### 3.4 Macroscopic and microscopic features of early enamel caries

The earliest macroscopic evidence of enamel caries is known as the “white spot lesion”. It is best seen on dried extracted teeth where the lesion appears as a small, opaque, white area. The color of the lesion distinguishes it from the adjacent translucent sound enamel. The color change is based on the increased porosity of the enamel tissue (due to demineralization) which alters the way in which the light is scattered. Enamel has a refractive index of 1.62 while the refractive
index of water is 1.33. Air-drying removes the water from intercrystalline spaces which are then filled with air, refractive index of 1.0. It is these differences in refractive index which alter the scattering of light. If air drying reveals a white spot in the enamel, the change in the enamel porosity is slight. However, if the porosity is visible as a white spot without air-drying, the porosity is larger.

![Figure 3.2. Ground section of carious lesion in (a) water and (b) quinoline [47].](image)

Demineralized lesions can be seen by examining a ground section (100 μm) under a light microscope with the aid of transmitted polarized light. Sound enamel has a pore volume of about 0.1%. The pore volume of a demineralized lesion body is 5% at its periphery, increasing to 25% or more in the centre. The body of the lesion is seen particularly if the ground section is examined in water (Figure 3.2(a)). In this case, the water molecules enter the pores in the enamel tissue, and since the refractive index of water is different from that of enamel, the caries appears dark. If the ground section is observed in a clearing agent such as quinoline having the same refractive index as the enamel the lesion body appears translucent, indicating the penetration of quinoline through the porous lesion body (Figure 3.2(b)). One of the important characteristics of early enamel caries is that the small lesion remains covered by a surface layer which appears as a relatively intact area superficial to the body of the lesion (Figure 3.2(a)). The zone has a pore volume of 1%, but if the lesion progresses the surface layer is eventually destroyed and a cavity forms. The precise mechanisms behind the relative protection against further dissolution of the outer 10-30 μm of the enamel as removal of mineral from the subsurface region continues is not fully understood[43]. Some researchers have suggested that this layer is associated with the special properties of surface enamel which shows a high degree of mineralization. However, a surface zone can also be seen when the original enamel surface has been removed. Another
explanation is the dynamic processes taking place at this surface. In any case, it has been reported that the surface layer in itself forms a diffusion barrier against subsurface uptake of mineral. For this reason it is a well-known clinical phenomenon that arrested lesions (lesions with non-cariogenic plaque on the surface) with an intact surface layer remain as scars in the tissue and do not become cavities.

3.5 Dental x-ray radiography

Dental x-ray radiography is one of the clinical diagnostic modalities available to dental practitioners for detecting dental caries. However, radiography has a poor sensitivity (less than 0.5 [29]) and fails to detect early caries. Figure 3.3 shows how different stages of caries development are classified in dentistry [43]. Approximately 30%-40% mineral loss is necessary before an early enamel carious lesion is visible radiographically [46] (advanced and severe caries classifications), while incipient/early stage is when the caries can be arrested and even remineralized [44][45]. It can take 9 months or longer before further demineralization of an early caries can make it visible on a radiograph [49].

Figure 3.3. Classification of dental caries. X-ray radiography is known to successfully detect advanced and severe caries. The incipient stage is when the caries can be remineralized [50].
The radiographs of Figure 3.4 show the poor sensitivity of conventional x-ray radiography in detecting incipient and moderate caries. Moreover, radiography incorporates ionizing radiation which is not safe to biological entities in some circumstances.

Figure 3.4. X-ray radiographs showing the poor sensitivity of the method in detecting incipient and moderate caries [50].

3.6 Tooth as a turbid medium

Light-tooth interactions can be classified into four categories (schematically shown in Figure 3.5):

1. Reflection
2. Transmission
3. Scattering
4. Absorption (followed by either radiative or non-radiative energy conversion)

While light scattering simply refers to a path change of photons without any change in their energy, in an absorption event the photon energy is either completely converted to heat or partially converted to heat along with the emission of photons of lower energy (longer wavelength), referred to as fluorescence photons. Parameters used to characterize tissue optical properties include photon scattering ($\mu_s$) and absorption coefficients ($\mu_a$), which refer to the average number of scattering and absorption events per unit length of a photon propagating
though the medium [51]. Together they amount to the total attenuation coefficient, given as: 

\[ \mu_t = \mu_a + \mu_s \]

which is an important optical parameter defining the total optical penetration depth.

![Light-tissue interaction mechanisms](image)

**Figure 3.5. Light-tissue interaction mechanisms [52].**

As discussed in section 3.1, the tooth is a multilayered structure, where each layer has its own thermal/optical properties and inhomogeneities. As a result, light propagation in teeth, similar to other biological tissues, is highly random and scattered and teeth can be considered as a turbid medium. However, the amount of scattering is a function of the photon energy. That is, longer wavelengths (lower energy photons) are less scattered compared to the shorter wavelengths (higher energy photons) and can therefore penetrate deeper into the tooth structure. Light scattering by the tooth at near infrared wavelengths is highly forward directed, with a mean cosine of the scattering angle close to unity (\(g \sim 1\)). As a result, these photons can reach deep structures in the enamel and even the dentine.

Although the amount of light scattering is a key factor in determining the light penetration depth, a photothermal signal is generated only when heat is released in an absorption event. Therefore, in order to obtain optimal results from a photothermal system, one should carefully choose the excitation and detection wavelengths based on both the scattering and absorption coefficients. Figure 3.6(a) shows the attenuation coefficients of enamel and water. Based on these spectra, the
range between 700 nm to 1200 nm (near infrared) seems to be perfect for excitation as laser radiation is not absorbed by water but is slightly absorbed by enamel to generate a photothermal signal. Figure 3.6(b) shows the infrared transmission spectrum of enamel. The transmission window in the 3-7 µm range (mid-infrared) seems to be a good choice for detection of infrared radiation from subsurface layers in the tooth.

The energy fluence of the one-dimensional, uniform, collimated beam incident on a homogeneous scattering and absorbing medium is calculated by Prahl et al. [53] and can be expressed as:

\[
I(z) = \beta \exp(-\mu_{\text{eff}} z) + \gamma \exp(-\mu z)
\]

where

\[
\begin{align*}
\beta &= \frac{I_o (9 + 6w) \mu' D}{(1 + w \sqrt{4 \mu_a D}) (1 - 9 \mu D)} \\
\gamma &= \frac{-2I_o}{1 - 9 \mu D}
\end{align*}
\]

\[(3.3)\]

\[
\mu' = \mu_s (1 - g), \quad \mu_t = \mu_s^* + \mu_t^*, \quad \mu_{\text{eff}} = \sqrt{3 \mu_t^* \mu_s^*}, \quad D = (3 \mu_t^*)^{-1}, \quad w = (1 + r)/(1 - r)
\]

where \(I_o\), \(g\), and \(r\) are the optical fluence on the sample surface \((z = 0)\), the average cosine of the scattering angle, and the internal diffuse reflection coefficient, respectively.

Figure 3.6. (a) Attenuation coefficient of enamel and water at near infrared wavelengths [54]. (b) Enamel infrared transmission spectrum indicating the primary absorbers in dental enamel [55].
Equation (3.3) models the diffuse optical field inside a turbid medium such as a tooth and can therefore be coupled with the heat diffusion equation to predict the photothermal response of dental samples.

**Table 3.1. Absorption and scattering properties of sound and carious enamel, [56] with modification.**

<table>
<thead>
<tr>
<th>Wavelength (nm)</th>
<th>Absorption coefficient $\mu_a$ (m$^{-1}$)</th>
<th>Scattering coefficient $\mu_s$ (m$^{-1}$)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Sound Enamel</td>
<td></td>
</tr>
<tr>
<td>543</td>
<td>&lt; 100</td>
<td>10,500</td>
</tr>
<tr>
<td>600</td>
<td>&lt; 100</td>
<td>3,300 - 7000</td>
</tr>
<tr>
<td>633</td>
<td>40 – 97</td>
<td>110 – 6600</td>
</tr>
<tr>
<td>700</td>
<td>&lt; 100</td>
<td>2,700 – 5,500</td>
</tr>
<tr>
<td>800</td>
<td>&lt; 100</td>
<td>3,300</td>
</tr>
<tr>
<td>1000</td>
<td>&lt; 100</td>
<td>1,600</td>
</tr>
<tr>
<td>1053</td>
<td>&lt; 100</td>
<td>1,500</td>
</tr>
<tr>
<td></td>
<td>Carious Enamel</td>
<td></td>
</tr>
<tr>
<td>600</td>
<td>--</td>
<td>55,000</td>
</tr>
<tr>
<td>633</td>
<td>--</td>
<td>32,000 – 157,000</td>
</tr>
</tbody>
</table>

The absorption and scattering properties of sound and carious enamel are presented in Table 3.1. The tabulated data suggests that the dominant attenuation mechanism in enamel is light scattering, which is significantly increased in carious enamel. The other interesting point is the large variation of data reported by different sources. The reason for such large variation is that, by nature, enamel composition and crystalline quality is different amongst people of different
ages, races, social background, etc. Moreover, different teeth in the same oral cavity may be exposed to different acidic and bacterial environments leading to variations in physical properties. Such variation is the nature of the tooth [57]. The same trend can be observed for the thermal properties of enamel as shown in Table 3.2.

Table 3.2. Thermal properties of sound enamel [56].

<table>
<thead>
<tr>
<th>Thermal Conductivity (W m$^{-1}$K$^{-1}$)</th>
<th>Thermal diffusivity (m$^2$ s$^{-1}$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.65 – 1.07</td>
<td>2.27×10$^{-7}$ – 4.7×10$^{-7}$</td>
</tr>
</tbody>
</table>

Generally, light absorption and scattering is greater in dentine due to its larger organic makeup compared to enamel and the presence of dentinal tubules. The light absorption and scattering coefficients in the visible light spectrum in dentine are typically in range of 300 – 400 m$^{-1}$ and 3,000 – 120,000 m$^{-1}$, respectively [58].

3.7 Emerging technologies in the field of early caries diagnosis

Since the early 1990s, a great deal of research has been focused on the introduction of new methods for detection of early dental caries. The significance of early caries detection is discussed in section 1.5. In fact, the National Institutes of Health (U.S.) consensus statement published in 2003 has identified the development of reliable methods for detecting early carious lesions as one of the major areas in which more research is needed [59].

Among the methodologies suggested so far, optical methods have shown great potential for caries detection. This is mostly due to the intrinsic contrast associated with these methods, as discussed in section 1.4. Light is generally more scattered and absorbed within carious regions thereby producing contrast between intact and demineralized regions. Perhaps the earliest optical method used in dental diagnostics is the fiber optic transillumination (FOTI) methods which uses high intensity white light for detecting caries. However, due to the significant scattering of light in the visible region, inconsistent results have been reported, mainly suggesting low sensitivity of the method [60]. Light-to-fluorescence conversion methods are intrinsically more sensitive to the presence of dental caries than direct optical methods because they feature low background signals and enhanced dynamic range. Quantitative light-induced fluorescence (QLF) is one such method that is based on the decrease in fluorescence intensity due to increased scattering from
demineralized spots [61]. Fluorescence is caused by the excitation of fluorophores contained within the enamel-dentine junction (EDJ) using visible light. QLF is an imaging modality that is capable of detecting early carious lesions [62]. However, the need for extensive operator training, detecting approximal caries, and the masking effects of surface stains are major challenges for this technique. DIAGNODent (DD) is another caries detection device based on fluorescence which uses red illumination to excite the bacterial porphyrins found in dental caries [63]. The major downside of DD is the fact that it measures bacterial activity rather than structural changes in enamel. Moreover, DD is not an imaging device and cannot reliably detect enamel caries [64].

Recently, promising results have been reported for detection of caries using NIR transmission/reflectance methods. These methods work on the same basis as FOTI but use NIR radiation instead of visible light, resulting in a deeper penetration depth because the scattering and absorption in enamel is significantly reduced in the NIR region. Fried et al. have reported on the enhanced penetration depth of NIR radiation in teeth as well as the insensitivity of NIR radiation to surface stains and non-calcified plaque [55][65]. In the recent work of Zakian et al. differential NIR reflectance has been used to quantify lesion severity [66]. Although the extent of the literature on NIR imaging is limited, it appears that NIR light is uniquely suited for dental inspection.

Thermal infrared (Planck) radiation from teeth can also be used to detect carious lesions. This form of detection belongs to the group of energy conversion methodologies with reduced signal baselines and enhanced dynamic range advantages akin to fluorescence. Kaneko et al. were the first to use passive infrared thermography (i.e., no excitation) to correlate the changes in enamel temperature to mineral loss and lesion depth [67]. However, the methodology has major limitations because the variations in the temperature of the mouth can influence the temperature readings. In a later work, an active thermography method with pulse heating was used to overcome this problem; but the results were still not promising since visible light was used as the excitation source (highly scattering) and the carious lesions examined could clearly be detected even using visual inspection (low sensitivity) [68].
3.8 Polarized Raman spectroscopy

Polarized Raman spectroscopy (PRS), pioneered in dental caries diagnostics by Choo-Smith et al. [69], is another emerging technology in the field of early caries diagnosis which uses the degree of polarization in the backscattered light to differentiate between intact and carious enamel. Briefly, in PRS a monochromatic, linearly-polarized beam of light is focused on the sample and the Raman spectra of the backscattered photons are recorded both in the parallel and perpendicular polarization directions, as shown in Figure 3.7.

Ko et al. [69] found that the increased porosity in early caries increases light scattering and scrambles the polarization, leading to greater depolarization of the 960 cm\(^{-1}\) phosphate Raman band of hydroxyapatite, Figure 3.7(b). On the other hand, this band in intact enamel maintains the polarization of the applied optical excitation and yields smaller depolarization values, Figure 3.7(a). Consequently, as suggested by Ko et al. [69], one can use the depolarization ratio of the 960 cm\(^{-1}\) band, \(\rho_{960}\), to monitor the structural changes of enamel:

\[
\rho_{960} = \frac{I_\perp}{I_\parallel}
\]  

where \(I_\parallel\) and \(I_\perp\) are the peak to baseline intensity of the 960 cm\(^{-1}\) band in the parallel and perpendicular spectra, respectively.

![Figure 3.7](image-url)  

Figure 3.7. Parallel- (upper trace) and perpendicular- (lower trace) polarized Raman spectra of (a) sound enamel (b) carious enamel [70].
Polarized Raman spectroscopy seems to be a good candidate for early caries diagnosis, however, its sensitivity to early caries detection is still to be determined, Figure 3.8.

### 3.9 Performance of FD-PTR in early caries diagnosis

The only research group that has been actively working on detection of early dental caries using the FD-PTR method is ours (CADIFT). Therefore, the amount of literature available on this topic is limited. The method involves the application of intensity modulated optical excitation on the tooth surface and simultaneous detection of the infrared (Planck or blackbody) radiation emanating from it with an infrared detector. The amplitude and phase of the surface temperature oscillation are obtained using lock-in signal processing methods. When light enters the tooth it is scattered preferentially in the carious areas where the pore volume is larger [43][71]. In general, the greater the scattering of light in an area, the higher is the probability of radiative absorption of light in that area. As a result, the thermal-waves that are generated in the porous regions will have greater amplitude than those generated in intact enamel [71][29]. Moreover, because the carious porous areas are close to the surface, they shift the thermal-wave centroid closer to the front surface and therefore decrease the phase lag between the applied optical excitation and the surface temperature [72]. In both cases (amplitude and phase), a pronounced contrast can be observed between intact and carious spots. Figure 3.9 shows the results of a line scan that was carried out at 10 Hz over an artificially created demineralized lesion [5]. It can be seen that as the lesion (located approximately from 1 to 3 mm) is increasingly demineralized from 1 to 10 days, the amplitude and phase values increase monotonically, providing a contrast between healthy and demineralized enamel.
The infrared detector signals obtained from turbid media, such as teeth, carry subsurface information in the form of a temperature depth integral in two distinct modes: conductively, from a near-surface distance (~ 50-500 µm) controlled by the thermal-diffusion length and thermal diffusivity of enamel, equation (1.3); and radiatively, through blackbody infrared emission from considerably deeper regions commensurate with optical penetration of the diffusely scattered laser-induced optical field (several millimeters). While the conductive mode only contributes to the phase of the detected infrared emission, the amplitude channel receives contributions from both the conductive and radiative modes [30]. As a result, the FD-PTR phase channel is the most interesting channel since its probing depth can be controlled according to equation(1.3) and allows for depth profilometry of the dental sample through a frequency scan [72][73]. Jeon et al. [29] carried out a detailed research on 52 extracted human teeth and reported that the sensitivity of FD-PTR to early caries detection was significantly higher than the conventional diagnostic methods as well as that of the DIAGNODent. The research carried out in CADIFT has shown that FD-PTR can successfully detect pit and fissures caries (located on the occlusal surface) [29], detect interproximal caries (located between adjacent teeth) [74], and monitor the early demineralization and remineralization of enamel [71]. However, FD-PTR is a single point measurement technique, not perfectly suitable for clinical practice.
3.10 Dental laser safety limit

As mentioned in section 2.2, in the developed thermophotonic system a collimated 20-mm diameter laser beam with average optical intensity 2.04 W/cm\(^2\) is applied to the sample. Although this optical intensity may seem to be above the permissible clinical diagnostic threshold, reflection at the enamel surface and the poor absorption of NIR light by both water and enamel, Figure 3.6(a), significantly reduce the effective power deposited in the tooth. To evaluate the safety of the developed system, a resistance temperature detector (RTD) was inserted into the pulp chamber from the back surface of a tooth while the thermophotonic imaging was carried out on the front surface. The 2.04 W/cm\(^2\) optical power density was found to increase the pulp chamber temperature by less than 5°C. According to the dental literature [75], an intrapulpal temperature increase of more than 5°C may cause irreversible trauma to the soft tissue inside the pulp chamber. Although the intrapulpal temperature rise caused in our system is marginally within the dental guidelines, it is possible to reduce the optical power density at the cost of longer measurement time (i.e., more averaging). Moreover, the optical power density required for removing enamel/dentine layers using thermal damage is significantly more than that used in our system (~ 7.5×10\(^8\) W/cm\(^2\) is required for dental ablation processes [76]).

3.11 Dental samples and controlled demineralization

To investigate the capabilities of the developed thermophotonic imaging system in detecting early enamel caries, several extracted human teeth were obtained from nearby dental offices. Healthy looking teeth with no visible stains, cracks, or any other defects were chosen for the artificial demineralization study, while relatively healthy teeth with suspicious white spots were chosen to test the system performance for detecting natural caries. Irrespective of sample type, samples were carefully cleaned and then mounted on LEGO blocks (15.8mm (W) x 15.8mm (D) x 9.5mm (H)) using a commercial epoxy putty adhesive. The putty was only applied to the root part of the samples and the surface of the LEGO block. The non-liquid putty state of the adhesive assured no diffusive contamination of the enamel. Mounting the teeth on LEGO blocks allowed them to be remounted in the same position in the experimental setup during repeated measurements. Samples were stored in an air-tight humid container before measurements. Inside the humid box, a small rectangular dish containing distilled water was used to maintain constant
humidity. The surfaces of the samples were not polished or altered in any way prior to the experiments.

In order to apply controlled demineralization of the dental samples, a demineralizing solution was prepared by our collaborators at the university of Texas, department of Dentistry. The solution was an acidified gel, consisting of 0.1 M lactic acid gelled to a thick consistency with 6% w/v hydroxyethylcellulose and the pH was adjusted to 4.5 with 0.1 M NaOH. Demineralization with acidified gel approximates the natural lesion, as it mimics the properties of actual dental plaques in the oral cavity [77]. Previous studies carried out in our lab show that this solution can produce a subsurface lesion in enamel with a sound surface layer, similar to those shown in Figure 3.2[71][57].

![Figure 3.10. A tooth sample during the artificial demineralization in the acidic gel.](image)

Since the goal of the controlled demineralization protocol was to see the contrast between demineralized and healthy areas in a whole tooth, a treatment protocol was followed in which the interrogated surface of the sample was covered with two coats of commercial transparent nail polish except for a rectangular window, henceforth referred to as the treatment window. The demineralization in the window was carried out by submerging the sample upside down in a polypropylene test tube containing 30 mL of demineralizing solution, Figure 3.10. After the treatment period the sample was removed from the gel, rinsed under running tap water for approximately 1 minute, and dried in air for 5 minutes. Then the transparent nail polish was
removed from the interrogated surface with acetone and the sample was again rinsed, dried, and stored in the air-tight humid box for at least 24 hours before thermophotonic imaging.

After the imaging, the sample was again covered with the transparent nail polish (except for the treatment window) and demineralized for an additional period of several days to investigate the progression of demineralization with time. The samples with natural caries were directly imaged after being mounted on LEGO blocks and no artificial demineralization was applied to them.

3.12 X-Ray radiography and transverse micro-radiography

Using standard dental x-ray equipment, radiographs were taken from a selection of samples before and after the treatment. Each tooth was mounted on a jig that fixed the distance between the film, tooth and x-ray tube head, and radiography was carried out on both the proximal and buccal sides. Radiography was not performed for all samples because the methodology is not sensitive enough to detect early caries, as discussed in section 3.5. However, transverse micro-radiography (TMR) was used as the gold standard to verify the diagnostic images obtained from the thermophotonic system.

![Figure 3.11. A snapshot of the TMR software calculating the lesion depth and mineral loss of artificially created early enamel caries (dark area).]
TMR experiments were carried out by our dental collaborator at the university of Texas where tooth slices approximately 100-μm thick were prepared from several regions of interest on samples using a water-cooled diamond-coated wire saw (Well, Le Locle model 3242, Switzerland). Both the sample slices and a standard aluminum step wedge (10 steps of 24.5 μm high) were microradiographed on type 1A high-resolution glass x-ray plates (IMTECH CA, USA) with a Phillips x-ray device using a nickel-filtered Cu-Kα target. This system generates monochromatic x-ray radiation at 184 Å, which is suitable for hydroxyapatite radiography. The plates were exposed for 10 minutes at 20 kV/10 mA prior to processing. Processing entailed five minutes in a developer (Kodak HR) and 15 minutes in a rapid-fixer (Kodak) before a final 30-minute wash period. After drying, the microradiographs were visualized using an optical microscope (Leica DMR) linked via a closed-circuit television camera (Sony, XC-75CE) to a computer. The enhanced image of the microradiograph was analyzed under standard conditions of light intensity and magnification, then processed, along with data from the image of the step wedge, using the TMR software (TMRW version 2.0.27.2, Inspektor Research Inc., Amsterdam, The Netherlands) to quantify the lesion parameters of integrated mineral loss (Δz, vol% μm) and lesion depth (LD, μm). The mineral loss was computed as the difference in volume percent of mineral between sound and demineralized tissue integrated over the LD. The LD was assessed as the distance from the measured sound enamel surface to the location in the lesion at which the mineral content was larger than 95% of the mineral content in sound enamel, Figure 3.11.

### 3.13 Polarized Raman spectroscopy

To compare the sensitivity of our thermophotonic method with other emerging technologies in the field of early caries diagnosis, polarized Raman spectroscopy measurements were carried out by our collaborators at Natural Resources Canada in Devon (Alberta).

Raman spectra of the tooth samples were acquired using a fully automated Renishaw inVia confocal Raman microspectrometer. Excitation was provided by a linearly polarized 633 nm Spectra Physics He-Ne laser with an output power of 30 mW and beam size of 1x1 mm² yielding an optical fluence of 3 W/cm² at the laser output. In this system a set of mirrors, positioned on automated stages, directs the laser beam to the microscope compartment where the sample is illuminated through a 50x objective lens. Scattered light is collected using a 180° backscattering geometry and directed to an optical notch filter that eliminates the intense Rayleigh scattering. A
rotary stage holds a half-wave plate (HWP) and an analyzer in the beam path at normal incidence; these components are used to determine the polarization characteristics of the Raman spectra of tooth enamel. The scattered light is dispersed by an 1800 line/mm holographic diffraction grating and focused on a CCD camera with a concave lens.

Parallel- and cross-polarized components of the scattered light were recorded at a series of points on the enamel of the tooth samples. The HWP was employed only for the parallel polarization measurements. All spectra were recorded using 50% of the available laser power, with a 50-s acquisition time and five accumulations. The Raman band at 960 cm$^{-1}$, due to the totally symmetric phosphate (PO$_4^{3-}$) vibration within hydroxyapatite, exhibits strong polarization dependence; accordingly spectra were acquired only in the 800–1100 cm$^{-1}$ range. Peak intensities were obtained by numerically fitting the data to Lorentzian functions. Several experiments showed that the background intensity in the spectra has negligible effects on the parallel-/perpendicular- polarization intensity ratios for healthy and demineralized enamel.
Chapter 4
Thermophotonic lock-in imaging

After a short review of the thermal-wave principles, the chapter focuses on FD-PTR and thermophotonic lock-in imaging of early dental caries. The sensitivity of the method to early caries detection is compared to that of the polarized Raman spectroscopy as a competing emerging technology in the field. This chapter is based on a provisional patent filed on the developed technology [78], a paper published in Journal of Biomedical Optics [30], and a paper currently under publication in Journal of Biomedical Optics [79].

4 Thermophotonic lock-in imaging

4.1 Basic principles of thermal-waves

Diffusion wave science has received much attention in recent years, ranging from thermal waves [80] to carrier plasma diffusion waves, diffuse photon density waves, and mass-transport waves [3]. One interesting application of diffusive waves is that of thermal waves. As discussed in the first chapter, following the absorption of modulated excitation, a periodic thermal field is formed inside the sample. The periodic field resembles a wave field and is therefore referred to as the thermal-wave field. Unlike hyperbolic travelling waves such as acoustic or optical waves, the physics of diffusive thermal waves is governed by the parabolic heat diffusion equation and in accordance to Fickian principles. The thermal wave field in a solid is a scalar field governed by the thermal diffusion equation,

\[ \nabla^2 T(x,y,z,t) - \frac{1}{\alpha} \frac{\partial T(x,y,z,t)}{\partial t} = - \frac{H(x,y,z,t)}{\kappa} \quad (4.1) \]

Here \( T(x,y,z,t) \) is the excess temperature at a point \((x,y,z)\) and time \( t \) relative to a reference temperature \( T_o \); \( H(x,y,z,t) \) is the source term, and is the rate of heat input per unit volume, \( \alpha \) [m\(^2\)/s] is thermal diffusivity, and \( \kappa \) [W/(m.K)] is thermal conductivity. When the source term is in the form of pulse excitation, (4.1) is solved in the time-domain but when the source is modulated (FD-PTR), it is more convenient to solve the frequency domain problem:

\[ \nabla^2 \theta(x,y,z,\omega) - \sigma^2 \theta(x,y,z,\omega) = - \frac{\tilde{H}(x,y,z,\omega)}{\kappa} \quad (4.2) \]
\[ \sigma = \sqrt{\frac{i\omega}{\alpha}} = \frac{1}{\mu}(1+i) \]  

(4.3)

where \( \theta(x,y,z,\omega) \) and \( \tilde{H}(x,y,z,\omega) \) are the frequency domain counterparts of \( T \) and \( H \), mathematically obtained through temporal Fourier transform. \( \sigma \) is the complex wavenumber and \( \mu \) is the thermal diffusion length defined by (1.3).

Many of the properties of thermal waves are embodied in equations (4.2) and (4.3). Equation (4.2) is very similar to the Helmholtz equation of hyperbolic wave-fields. However, the complex character of \( \sigma \) and the gradient driven nature of heat diffusion do not allow the generation of hyperbolic waves. An immediate consequence of the gradient-driven principle is the lack of wave fronts and the diffuse depth-integrated (rather than localized) transport of power through the medium, leading to poor diffusive axial resolution. This diffusive limitation has tempted several researchers to treat the frequency-domain heat diffusion equation as a hyperbolic Helmholtz equation with a complex wavenumber, defining wavefronts and applying Snell’s refraction laws to thermal waves [81][82][83][84]. While these interpretations seem to allow for the presence of wavefronts and localized transfer of power by thermal waves, they violate the second law of thermodynamics since they admit the propagation of energy against the field gradient (i.e., a temperature field in which heat would appear to move from cold to hot) [85].

In photothermal science most of the applications involve the use of wide beam excitation where the excitation source covers an area on the sample that is much larger than the thermal diffusion length, (1.3). In these situations, thermal-waves are effectively planar and the three dimensional equation (4.2) is reduced to a one dimensional problem.

\[ \frac{\partial^2}{\partial z^2} \theta(z,\omega) - \sigma^2 \theta(z,\omega) = -\frac{\tilde{H}(z,\omega)}{\kappa} \]  

(4.4)

Therefore, solving (4.4) for an opaque homogeneous semi-infinite medium (e.g., a thick steel block) whose surface is subjected to plane harmonic heating of the form \((I_0/2)[1+\cos(\omega t)]\) and transforming back to the time-domain using an inverse Fourier transform yields the equation of thermal-waves for a simple geometry:
$$T(z,t) = \frac{Q_o}{2\sqrt{2}\kappa} \mu \exp(-\sigma z) \exp\left[i\left(\omega t - \frac{\pi}{4}\right)\right]$$  \hspace{1cm} (4.5)

Several basic properties of thermal waves can be understood from (4.5). For a given sample and at a given laser modulation frequency ($\omega$), the term “$\exp(-\sigma z)$” suggests that the thermal waves are exponentially damped as they travel inside the medium. As a result, the thermal diffusion length is defined as the depth at which the thermal wave amplitude is reduced to $e^{-1}$ of its initial value. Moreover, the damping constant ($\sigma$) is directly proportional to the modulation frequency ($\omega$). This suggests that the low frequency thermal waves are less damped and can propagate deep into the sample, while the high frequency thermal waves are heavily damped and limited to the near-surface depths.

The radiometric signal detected by the infrared detector originates from the sample surface ($z = 0$) and can be calculated as:

$$S(t, \omega) = \overline{\mu}_{IR} \int_{0}^{\infty} T(z,t) \exp(-\overline{\mu}_{IR} z) dz$$  \hspace{1cm} (4.6)

where $\overline{\mu}_{IR}$ is the average infrared absorption coefficient of the sample in the spectral range of the detector. For opaque materials, light is absorbed within a few microns of the interrogated surface. Therefore, $\overline{\mu}_{IR}$ is very large and can effectively be taken as infinite.

$$\overline{\mu}_{IR} \rightarrow \infty \Rightarrow S(t, \omega) = \frac{Q_o}{2\sqrt{2}\kappa} \times \mu \times \exp\left[i\left(\omega t - \frac{\pi}{4}\right)\right]$$  \hspace{1cm} (4.7)

Equation (4.7) shows that for a semi-infinite opaque material, the amplitude of the radiometric signal is proportional to $\mu$ or alternatively proportional to $1/\sqrt{\omega}$ and, regardless of the modulation frequency ($\omega$), the phase is always $-45^\circ$. However, the presence of a defect below the surface violates the semi-infinite assumption and results in a higher photothermal amplitude as well as a photothermal phase value other than $-45^\circ$ to show the contribution of the reflected energy by the defect.
4.2 Physics of dental samples

To date, thermal-wave fields have been derived analytically for a wide range of geometries and materials [80]. One of the most interesting class of materials is the turbid media where the applied optical excitation is gradually absorbed within the medium, as suggested by (3.3). As a result, the physics of the problem in turbid media is quite complicated and is governed by coupled diffuse-photon-density and thermal-wave fields. Here, using the basic optics equations (i.e. Snell’s and Beer’s laws) the coherent light field, \( \Psi_c(z) \), is first modeled. The modeled coherent field is then used to solve the light diffusion differential equation and find the diffuse optical field, \( \Psi_d(z) \). The sum of coherent and diffuse fields provides the total optical field intensity inside the turbid medium, \( \Psi_t(z) \).

\[
\begin{align*}
\frac{d^2}{dz^2} \Psi_d(z) + 3 \mu_a \mu_s' \Psi_d(z) &= -\frac{1}{D} \mu_s \left( \frac{\mu_t + g \mu_a}{\mu_t - g \mu_s} \right) \Psi_c(z) \\
\Psi_t(z) &= \Psi_c(z) + \Psi_d(z)
\end{align*}
\]

(4.8)

Subsequently, the spatially attenuated total optical field is used as the source term of the heat diffusion equation (4.9) to calculate the thermal-wave field and the corresponding radiometric signal at the modulation frequency, \( \text{Signal}(\omega) \).

\[
\begin{align*}
\frac{d^2}{dz^2} T(z; \omega) - \sigma^2 T(z; \omega) &= -\frac{\mu_a}{\kappa} \Psi_t(z; \omega) \\
\text{Signal}(\omega) &= \mu_{th} \int_0^l T(z; \omega) \exp(-\mu_{th} z) dz
\end{align*}
\]

(4.9)

The above equations have been analytically solved in our lab for both intact and carious enamel [40]. While the analytical solutions help to explain the phenomenon, their use is quite limited since the solutions depend on a significant number of thermal and optical properties of enamel that have not been accurately measured and most importantly are significantly variable in enamel.
4.3 Justification of one-dimensional solution

A one-dimensional solution to the heat diffusion problem is widely used in the thermal-wave field, because in most applications the optical excitation beam size is much larger than the thermal diffusion length. One simple way to experimentally verify this assumption is to compare the frequency scan phase plots (phase vs. frequency) of a sample obtained at several excitation beam sizes. Figure 4.1 shows such a plot for a standard glassy carbon sample. It can be seen that at beam sizes larger than 700 µm, the effects of lateral heat diffusion become minimal and the phase plots become independent of the beam size, suggesting the formation of planar thermal-waves (i.e., 1-D heat flow).

Figure 4.1. Frequency scan phase plots of a standard glassy carbon sample at several beam sizes.

Figure 4.2 shows the results of a similar investigation carried out on a tooth sample. Likewise, the phase plots merge together for beam sizes above 310 µm to confirm the formation of planar thermal-waves, independent of the beam size. The reason for the formation of planar thermal-waves in a tooth at smaller beam sizes than those of the glassy carbon sample is that light photons are highly scattered in turbid media and the beam size is further increased as light penetrates into the tooth. It should be noted that the beam sizes used in our thermophotonic experiments (~ 20mm) were significantly larger than the beam sizes of Figure 4.1 and Figure 4.2 and therefore the one-dimensional solution to the heat diffusion problem is justified.
4.4 Justification of the controlled demineralization protocol

As mentioned in section 3.11, a controlled demineralization protocol has been adapted to investigate early caries formation and progression in enamel. In order to understand the effects of the protocol’s non-demineralizing parameters on the thermophotonic images, two experiments were carried out. In the first experiment the effect of nail polish was investigated. A healthy tooth was chosen and thermophotonically imaged, Figure 4.3(b). Then, the sample was covered with two coats of transparent nail polish except for the treatment window shown in Figure 4.3(a). The nail polish was removed after 24 hours and the sample was imaged again, Figure 4.3(d). Figure 4.3(c) shows the normalized phase values along row 70 of the thermophotonic images taken before and after application and removal of nail polish. It can be seen that the nail polish has no significant effect on the thermophotonic phase values. However, the images and plots suggest that the phase image becomes more uniform after the nail polish removal. The reason for this uniformity may be that the applied acetone has removed the dirt and organic material on the surface of enamel. The results of Figure 4.3 suggest that application of nail polish does not mark the treatment window and therefore does not affect the thermophotonic diagnostic images.
Figure 4.3. (a) A DC infrared image of the sample showing the position of the window. Normalized thermophotonic phase image of sample (b) before and (d) after applying the nail polish. (c) Normalized phase values along the horizontal lines shown in parts (b) and (d).

The next experiment was to see if the chemicals present in the demineralization gel, other than the H⁺, mark the treatment window. To achieve this goal, a gel with the exact chemical contents of the original gel was prepared; however, the pH was set to 7 to avoid any demineralization. In order to see the effect at all frequencies, a photothermal frequency scan was performed on a point on the interrogated surface using the experimental setup depicted in Figure 2.1. It can be seen that the exposure to the gel can affect the amplitude channel since the gel can change the emissivity of the surface. But, the results of the photothermal phase values clearly show that the effect of non-demineralizing chemicals in the gel on the emissivity-normalized phase channel is minimal and the phase values obtained from 1 to 300 Hz are more or less the same before and after 2 and 5 days of exposure to the neutral gel.
Figure 4.4. Photothermal (a) amplitude and (b) phase frequency scans of a point on the sample surface before and after 2 and 5 days of exposure to the pH7 gel.

Figure 4.5. (a) Radiographic image of the cross section of the tooth and (b) TMR mineral profile of the sample after 5 days of exposure to the pH7 gel showing no demineralization.

Comparison of Figure 4.5(a) and Figure 4.5(b) with Figure 3.11 proves that the chemicals in the neutral gel do not remove mineral from the enamel.

4.5 Thermophotonic imaging of caries initiation and progression

A healthy tooth with no visible stain or crack was prepared according to the procedure mentioned in section 3.11. The study on this sample (sample A1) was carried out using the FD-
PTR and thermophotonic lock-in imaging systems, Figure 2.1 and Figure 2.3 respectively. Experimental baselines were acquired by running the experiments before any demineralization. Subsequently, a rectangular treatment window of size 1.2mm (W) × 4mm (H) was located on the sample and FD-PTR frequency scans and thermophotonic imaging were carried out on the sample after 2, 4, 8, and 10 days of exposure to the acidic gel. Finally, the thermophotonic imaging results were compared with those of the conventional diagnostic modalities (i.e. radiography and visual inspection) and verified through the destructive TMR experiments.

When modulated optical excitation is absorbed inside the tooth, the subsequent heat generation gives two contributions to the infrared camera signal. First, an oscillatory heat distribution (thermal-wave) is formed at or near the surface within a thermal diffusion length which will conductively reach the surface of the tooth and contribute to the camera signal in the form of a depth integral through infrared emission. Second, direct thermal infrared (Planck) emission occurs from all absorption locations (surface and subsurface) with IR photon back-propagation through the enamel due to the transmittance of enamel in the mid-infrared region [86]. Considering the speed of light and the thickness of enamel, the latter contribution is effectively instantaneous and therefore there will be no phase shift between the direct infrared (Planck) emission responses received from different depths in the tooth. Consequently, there will be no contrast contribution from this type of direct emission in the TPLI phase image; however, direct emission will contribute to the amplitude image contrast as these images are concerned with the amplitude of infrared emission (number of IR photons) received at a specific modulation frequency, a function of the local absorption coefficient and optical-to-thermal energy conversion efficiency [72].

On the other hand, the thermal-wave contribution to the infrared camera signal is not instantaneous because the speed of propagation of the heat is significantly smaller than that of light. Consequently, absorption at different depths will result in different phase values at a fixed modulation frequency. Unlike pure thermal waves generated in opaque media, optically non-saturated photothermal waves carry optical as well as thermal information. The most important features of thermal waves are that their effective penetration depth can be controlled through the modulation frequency based on (1.3) and that the signal carries optical absorption information from depths $d \leq \mu(f)$ [80]. Since the thermal-wave generated infrared emission is the dominant
source of the contrast in thermophotonic lock-in phase images, one can control the imaging depth by adjusting the modulation frequency according to (1.3). Due to the small thermal diffusivity, $\alpha$, of enamel the maximum thermal diffusion length (i.e., at the minimum modulation frequency $\sim 1$ Hz) is on the order of $\sim 300$ μm. Therefore, phase images are best used for detecting inhomogeneities at short subsurface distances within the enamel such as the early demineralization and carious lesions, while amplitude images can be used to detect deep features due to the large optical absorption depth of enamel. Moreover, surface stains and non-calcified plaque are not expected to produce artifacts in the TPLI images due to their transparency to the 808 nm NIR excitation source used in the setup [65].

![Graph showing amplitude and phase frequency scans](image)

**Figure 4.6.** FD-PTR (a) amplitude and (b) phase frequency scans obtained at the center of the treatment window of sample A1 at several stages of controlled demineralization. Insets: demineralization times (days, D).

PTR frequency scans using a focused laser beam allow physical insights into the signal generation process, not directly available through thermophotonic images. Preferential absorption of incident light at carious and demineralized regions is the intrinsic source of contrast in photothermal dental techniques and is directly linked to the mineral content of the enamel. Figure 4.6(a) shows the PTR amplitude frequency scans of a spot at the center of the treatment window of sample A1. It can be seen that at any given treatment time the photothermal amplitude decreases as the laser modulation frequency increases, a typical dental photothermal response [87]. Furthermore, at any given modulation frequency the photothermal amplitude
monotonically increases with the progression of enamel demineralization caused by the acidic treatment gel. This is due to the fact that demineralization decreases the mineral density of enamel, thereby making it more porous and degrading its thermophysical properties while enhancing light scattering and absorption [71]. In general, the greater the optical scattering in a dental region, the higher will be the probability of light absorption in the region. Therefore, both the direct and the thermal-wave-generated emissions yield greater contributions in the porous/carious regions than those generated for intact enamel, increasing the photothermal amplitude [29][72][74].

The PTR phase frequency scans of the same spot at several treatment times are plotted in Figure 4.6(b). The dominant feature of this plot is the appearance of a maximum point in the curves for the treated (demineralized) cases. The post-demineralization maximum has also been observed in previous FD-PTR studies [71] and was linked to the formation of a demineralized surface layer which supports standing thermal waves. Moreover, as treatment time increases, the phase maximum shifts toward lower frequencies as a result of a thicker surface layer, supporting standing waves of increased thermal wavelength \( \lambda = 2\pi \mu \) and decreased frequency [71]. As a result of these maxima, it can be seen that the trends between intact and demineralized enamel are reversed by increasing the modulation frequency from 1 to 120 Hz. Unlike the phase curves of the treated samples, the phase of the intact enamel is a monotonically increasing function of modulation frequency.

Trapping of light in demineralized regions of poorer optical and thermal properties than intact enamel confines its penetration and shifts the subsequently generated photothermal temperature distribution closer to the surface as compared with a healthy region. These confinement effects result in a PTR phase lead at carious spots at a given modulation frequency [72]. Generally, the size of the phase channel error bars decreases as demineralization proceeds since demineralization increases the PTR signal amplitude and improves the signal-to-noise ratio. The results of Figure 4.6 suggest that both amplitude and phase channels can be used to detect demineralization in teeth and monitor lesion evolution.

Figure 4.7(a) shows an optical image of sample A1 before application of demineralization in the treatment window. The dashed rectangle in this figure shows the area that was imaged using our TPLI system while the solid rectangle depicts the location of the treatment window. The optical
image shows traces of a discontinuity on the surface of the enamel (feature 1). The two vertical lines on the root area show the approximate lateral position of the treatment window (solid rectangle). These lines were meant to aid the operator with sample alignment and are of no scientific importance. Figure 4.7(b) “F” shows the x-ray radiograph of the untreated sample (“before”) at the same view as the optical image and Figure 4.7(b) “S” shows the side view indicated by the arrow in the optical image. Based on these radiographs, sample A1 was a relatively healthy tooth before application of artificial treatment and it is interesting to note that the discontinuity on the enamel surface (feature 1) could not be resolved in either of the x-ray radiographs.

Figure 4.7(c) and Figure 4.7(d) are the x-ray radiographs and the optical image taken from sample A1 after 10 days of treatment, respectively. None of these images show even a trace of mineral loss in the treatment window, demonstrating the insensitivity of conventional clinical diagnostic methods to early demineralization. Figure 4.7(e) depicts the TPLI phase image of sample A1 taken at 10 Hz before any treatment. This image not only shows the enamel discontinuity observed in the optical image (feature 1) but also reveals the presence of a vertical crack (feature 2) and the CEJ (feature 3). Neither the x-ray radiographs (Figure 4.7(b)) nor the optical image (Figure 4.7(a)), could resolve this vertical crack (feature 2). The reason of the high sensitivity of thermophotonic images to cracks is their fissured nature, which enhances the photothermal temperature field through thermal-wave flux localization that generates high contrast.

The appearance of the enamel discontinuity in the phase image (feature 1 in Figure 4.7(e)) has a similar physical origin to that of cracks. We postulate that feature 1 was caused by the excessive force applied during tooth extraction. The effects of cracks and discontinuities on the photothermal phase were first reported in the focused-laser-based FD-PTR experiments of Nicolaides et al.[88] and later verified by Jeon et al. [72]. The appearance of a dark band in the phase images of Figure 4.7 (feature 3) at the CEJ level suggests the presence of natural caries at this position. According to the dental literature [43], enamel of the cervical margin of the tooth is one of the locations which favors plaque formation and is therefore prone to demineralization. It can be seen that all the thermophotonic phase images of Figure 4.7 detect feature 3 with high sensitivity (contrast). In fact, one of the most important advantages of TPLI imaging compared to single point FD-PTR measurements is the significant improvement in the wealth of data resulting
in excellent contrast and reliability of the results in real time, in direct comparison with conventional radiographs. Point-by-point measurements would require much longer time spans to produce surface images, which would be impractical in clinical applications.

Figure 4.7(f) to Figure 4.7(i) show the phase images taken at 2, 4, 8, and 10 days of treatment (mineral loss only within the treatment window), respectively. It should be noted that the same contrast mapping (linear, with identical thresholds) has been used in all TPLI images of Figure 4.7 to ensure the validity of comparison between the images. It can be observed that as treatment time increases the treated window becomes more apparent while the other features in the images remain more or less the same. The mean phase values within the treatment window (empty rectangle in Figure 4.7(e)) for the untreated, 2D, 4D, 8D, and 10D samples are found to be $-7.47^\circ$, $-25.22^\circ$, $-31.75^\circ$, $-42.07^\circ$, and $-49.66^\circ$, respectively. This monotonic decrease in the phase lag is due to the progression of the lesion into the enamel. As the lesion thickness increases, light is more strongly absorbed and the thermal-wave centroid shifts closer to the surface, thereby decreasing the phase lag between the applied optical excitation and the received infrared response. The phase lag also decreases with respect to the intact state (“before”). Feature 4 in Figure 4.7(f) is most probably a material inhomogeneity formed as a result of incomplete removal of nail polish from the enamel surface after the second day of demineralization. This feature disappeared after the next demineralization cycle. It appears that feature 5 is stress-induced cracks that were formed during tooth extraction. The cracks become more apparent toward the later stages of demineralization (Figure 4.7(e) to Figure 4.7(i)) due to successive nail polish penetration. It is known that nail polish can penetrate tens of micrometers into dental enamel [89].
Figure 4.7. (a) Optical image and (b) front (F) and side (S) x-ray radiographs of sample A1 before treatment. (c) Front (F) and side (S) x-ray radiographs and (d) optical image of sample A1 after 10 days of mineral loss within the treatment window. TPLI phase images of sample A1 (e) before treatment and after (f) 2, (g) 4, (h) 8, and (i) 10 days of demineralization within the treatment window. (j) Phase profiles along the dashed line shown in part i for samples at several demineralization stages. The dashed vertical lines show the location of the treatment window. (k) TMR mineral profile along the center of the treatment window of the 10-day demineralized sample.
Figure 4.7(j) is a plot of transverse profiles of the phase images along the dashed line shown in Figure 4.7(i). The dashed vertical lines represent the location of the treatment window (centered at 6.27 mm). It can be seen that as demineralization progresses, the absolute thermophotonic phase values increase within the treatment window but remain approximately the same outside the treatment window. Furthermore, examination of these phase profiles reveals that the demineralization has not only propagated vertically into the enamel but has also spread out laterally. However, a part of this lateral spread may be due to the error of repositioning the treatment window in exactly the same place in repeated measurements. The two dips in the phase values at ~3.9 mm and ~8.8 mm lateral positions are related to the vertical defects at those locations (features 2 and 1, respectively).

Figure 4.7(k) represents the TMR mineral profile vertically along the center of the treatment window. A mean lesion depth of 326.4 μm and mineral loss of 5,710 vol%. μm was reported by the TMR software for the lesion produced at the center of the treatment window. It can be seen that the lesion retains a relatively well-preserved surface layer with a moderate mineral loss over a large depth. It is somewhat surprising to find such a deep lesion formed after only 10 days of demineralization, but it is a well-known fact that the rate of demineralization can vary greatly among different teeth [57].

Figure 4.8 shows how the effective imaging depth can be controlled by changing the modulation frequency. In Figure 4.8(a) to Figure 4.8(d) the modulation frequency is increased from 3 to 100 Hz without any change in the sample. Based on (1.3), increasing the modulation frequency decreases the thermal diffusion length and allows for investigation of surface features. The other interesting feature is that the axial/depth resolution is significantly reduced at low frequencies, making the image blurry as seen in Figure 4.8(a). The poor resolution at low modulation frequency is one of the well-known challenges of FD-PTR/LIT.

Figure 4.9 shows the non-emissivity normalized thermophotonic lock-in amplitude images. As mentioned in section 4.4, these images are affected by the gel and most importantly by the transfer function of the thermophotonic system. The presence of a bright circular area at the center of Figure 4.9(c) and Figure 4.9(d) is due to the fact that the custom made extension tube allows more photons to reach the central parts of the camera’s detector array, whereas the corners receive less photon fluence.
Figure 4.8. Thermophotonic lock-in phase images of sample A1-10D at several frequencies.

Figure 4.9. Thermophotonic lock-in amplitude images of sample A1-10D at several frequencies.
In general, while amplitude images have significantly higher signal-to-noise ratio compared to phase images, the latter are the dominant diagnostic and NDT images used in the literature. The information of the amplitude images, however, can be used to complement that of the phase images.

### 4.6 Thermophotonic imaging of natural caries

To evaluate the performance of the developed TPLI imaging system to real-scenario problems, samples with natural caries were selected and mounted on LEGO blocks. Samples A2 and A3 (Figure 4.10(a) and Figure 4.11(a), respectively) had stained fissures suspected to be caries.

Figure 4.10. (a) Optical image of sample A2. Thermophotonic (b) amplitude and (c) phase images of sample A2 obtained at 10 Hz. Thermophotonic (d) amplitude and (e) phase images of sample A2 obtained at 100 Hz. (f)-(i) TMR mineral profiles of points f-i, respectively, indicated in part e.
Figure 4.10 presents the results obtained from sample A2. Unlike Sample A1, the occlusal surface was investigated for this sample. Figure 4.10(b) shows the TPLI amplitude image taken at 10 Hz. The amplitude image shows the presence of caries at several locations but the image is rather diffuse. Using this image, four regions of interest (i, f, g, and h) were identified within the area of the optical image (Figure 4.10(a)). The TPLI phase image taken at 10 Hz is shown in Figure 4.10(c). No features can be resolved in the blurry phase image. The reason for such poor resolution is the relatively long, diffusion-limited thermal wavelength at 10 Hz. In fact, the contributions from deeper features in these images have interfered with those from features closer to the surface. Consequently, to avoid the interfering effects of deep features, we reduced the thermal wavelength by generating images at 100 Hz in order to effectively detect the areas of mineral loss in the pits and fissures of the occlusal surface and the near-subsurface regions. The resolution improvement at higher frequencies is a well-known phenomenology of thermal-waves also reported by Jeon et al. [72], and is clearly visible in the amplitude and phase images obtained at 100 Hz (Figure 4.10(d) and Figure 4.10(e)). The pits labeled f, g, and h in Figure 4.10(e) are shown as dark carious regions (similar to the treatment window of Figure 4.7(c) to Figure 4.7(f)) but the groove i is shown as a healthy bright spot (similar to the intact regions in Figure 4.7(b) to Figure 4.7(f)). The TMR profiles obtained at these points clearly show the presence of mineral losses at points f, g, and h, while no significant mineral loss is observed in the mineral profile obtained at point i. The TMR profiles fully validate the results of our non-contact, non-destructive imaging method and underscore the importance of TPLI as a dynamic modality as opposed to dc IR imaging [67].

Figure 4.11(a) and (b) depict the optical images of occlusal and approximal views of sample A3, respectively. While the occlusal (interrogated) surface appears relatively healthy, the presence of caries is obvious on the approximal side (shown by an arrow in Figure 4.11(b)). The inset in Figure 4.11(b) shows that the center of the natural approximal caries (feature 7) is closer to the occlusal surface than the rest of the lesion (feature 6), which remains below lesion 7. The TPLI amplitude image taken from the occlusal surface at 10 Hz (Figure 4.11(c)) not only detects the approximal caries, but also shows the depth difference between features 6 and 7 through differences in contrast levels. The feature closer to the occlusal surface (feature 7) has higher contrast compared to the deeper feature (6). The fact that such features cannot be resolved in the
TPLI phase image taken at 10 Hz (Figure 4.11(d)) is due to the poor resolution associated with low frequency thermal waves.

**Figure 4.11.** Sample A3, (a) Occlusal and (b) approximal optical images, TPLI (c) amplitude and (d) phase images obtained at 10 Hz, TPLI (e) amplitude and (f) phase images obtained at 100 Hz, and (g) TMR profile along the depth of the approximal natural caries.

The depth profilometric capability of TPLI can be understood by comparing Figure 4.11(c) and Figure 4.11(e). According to the definition of thermal diffusion length, increasing the modulation frequency from 10 to 100 Hz decreases the thermal diffusion length. This, in return, decreases the effective detection depth and results in the near-fading of the deeper feature (feature 6), while the closer feature to the surface (feature 7), is clearly resolved. However, due to the well-known inverse relationship between the modulation frequency and the photothermal signal amplitude [80] both features have less contrast at 100Hz (Figure 4.11(e)) compared to that at 10Hz (Figure
Therefore, generating TPLI images at various frequencies can help to detect sub-surface lesions with improved spatial resolution at increased frequency and also reveal depth information about the features.

The TPLI phase image obtained at 100 Hz (Figure 4.11(f)) yields more precise depth profilometric information when compared to the 10-Hz phase image, Figure 4.11(d). Finally, Figure 4.11(g) presents the TMR profile taken vertically along the depth of the natural approximal caries lesion shown by the arrow in Figure 4.11(b). The profile exhibits classic profile of an enamel lesion with a well preserved intact surface layer. In dental practice, proximal caries is very challenging to detect as it cannot be easily probed. One of the advantages of the experimental setup used in this thesis is that when the occlusal surface of the tooth is examined, proximal surfaces can be probed just as well as the approximal surfaces. As a result, the study carried out on sample A3 suggests that TPLI has the potential to detect proximal caries when viewed photothermally from the accessible occlusal surface and underscores two outstanding features of TPLI imaging: subsurface detection and depth profilometry.

4.7 TPLI imaging vs polarized Raman spectroscopy

To compare the sensitivity of the developed thermophotonic system with that of PRS two samples were prepared and mounted on LEGO blocks. Sample S1 had two treatment windows on the interrogated surface while sample S2 had only one treatment window. The left and right treatment windows of S1, Figure 4.12(a), were treated for 2 and 4 days, respectively, and that of S2, Figure 4.12(c), was treated for 20 days. Thermophotonic lock-in imaging was carried out using the 808 nm laser at 1Hz. PRS measurements were performed using a 633 nm laser based on the principles and instrumentation discussed in sections 3.8 and 3.13. The use of short laser wavelengths in Raman spectroscopy sometimes gives rise to significant fluorescence backgrounds that mask weak Raman bands. In these experiments the 633 nm excitation yielded acceptable backgrounds which did not adversely affect intensity measurements, Figure 4.12(d) and Figure 4.12(e). Specifically, baseline effects were minimal as there were no stains on the samples and therefore the capability of Raman spectroscopy with regard to caries detection was not affected [90]. The wavelength of the laser used in PRS study is slightly shorter than that used in the thermophotonic system, an advantage in PRS since the intensity of Raman scattering is inversely proportional to the fourth power of the excitation wavelength ($I \propto 1/\lambda^4$).
Figure 4.12. Photographs of S1 (a) before and (b) after the treatment. (c) Photograph of sample S2 after the treatment. The dotted and solid rectangles show the imaged area and the locations of the treatment windows. The cross (solid) and parallel (dashed) Raman spectra of (d) a healthy spot on S1 and (e) the 20-day-treated caries in sample S2.

Figure 4.13 represents the thermophotonic phase images of S1 and S2 before and after the treatment using identical linear contrast mapping. Figure 4.13(a) and Figure 4.13(b) show that even before applying the artificial demineralization, thermophotonic imaging can detect the inhomogeneities in the sample. The phase image of S1 after the treatment, Figure 4.13(c), clearly reveals the 2 (left) and 4 (right)-day-treated windows, 2-D and 4-D respectively, which are not visible in the photograph in Figure 4.12(b). The phase values at the center of the 2-D and 4-D regions (indicated by the squares in Figure 4.13(c)) were found to be $61.64 \pm 1.23^\circ$ and $57.89 \pm 1.17^\circ$, respectively, while the phase values in the exact same regions before treatment (i.e., healthy) were measured to be $71.05 \pm 3.96^\circ$ (indicated by the squares in Figure 4.13(a)). Such a good contrast between the intact and infinitesimally demineralized enamel comes from the way light interacts with enamel. Intact enamel is semi-transparent to NIR radiation and as a result light is gradually absorbed as it penetrates deeply into the enamel, yielding a thermal-wave
field with a deep centroid (large phase values). Demineralization, on the other hand, always starts from the enamel surface and replaces the hydroxyapatite crystals of this close-to-surface enamel with micro-cavities, increasing light scattering and absorption near the surface and therefore shifting the thermal-wave centroid towards the surface, yielding smaller phase values. Therefore, the mean phase value of the 20-day-treated window, Figure 4.13(d), was found to be smaller than those of the 2-D and 4-D regions (44.50 ± 0.93°) as the additional treatment days resulted in more light absorption closer to the enamel surface.

![Figure 4.13](image)

*Figure 4.13. Thermophotonic phase images of (a) S1 and (b) S2 before demineralization treatment and (c) S1 and (d) S2 after treatment using the linear contrast mapping shown in the figure. The squares (20×20 pixels²) indicate the regions used for statistical analysis.*

Figure 4.12(d) and Figure 4.12(e) plot the cross and parallel polarized Raman spectra of healthy and carious spots on S2, respectively. According to the literature [69][91][92], the dominant Raman band is the 960 cm⁻¹ phosphate band (PO stretching) of hydroxyapatite. Previous studies have shown that this is the most sensitive Raman band with regards to the structural changes in dental enamel during early demineralization. Moreover, the Raman spectra of healthy and 20-D regions clearly show that the fluorescence background did not obscure the PO band, as the samples were artificially demineralized and no stain was present on the surface [90]. While the
Raman response of the healthy enamel, Figure 4.12(d), maintains the polarization of the applied excitation, the increased scattering by the caries scrambles the polarization and yields dominant 960 cm$^{-1}$ bands in both parallel and perpendicular polarized spectra, Figure 4.12(e). Consequently, as suggested by Ko et al. [69], one can use the depolarization ratio of the 960 cm$^{-1}$ Raman band, (3.4), to monitor the structural changes in enamel. However, the question remains as to how sensitive the $\rho_{960}$ is to early structural changes.

![Figure 4.14](image)

**Figure 4.14.** The mean (a) Raman depolarization ratio and (b) thermophotonic phase values obtained in healthy, 2-D, 4-D, and 20-D regions of samples S1 and S2 along with their standard deviations. (c) The mean transverse micro-radiography mineral profiles of 2-D, 4-D, and 20-D treatment windows.

The bar plot of Figure 4.14(a) represents the mean $\rho_{960}$ obtained from S1 and S2 along with the standard deviation of these measurements over the healthy, 2-D, 4-D, and 20-D regions. The number next to each error bar depicts the number of Raman experiments carried out on that region. The 47 Raman measurements suggest that one cannot statistically differentiate the 2-D and 4-D treatment windows from healthy areas, nor can one garner any statistical difference between the 2-D and 4-D early caries. However, the well-developed 20-D caries of S2, which as
shown in Figure 4.12(c) can even be detected by visual inspection, is detectable by polarized Raman spectroscopy. In analogy to this figure, the bar plot of Figure 4.14(b) shows the mean thermophotonic phase values and their standard deviations over the regions of interest in S1 and S2 (shown by the squares in Figure 4.13). It can be seen that the phase value decreases as more mineral is removed from the enamel shifting the thermal-wave centroid closer to the interrogated surface of the enamel. Moreover, it can be seen that thermophotonic imaging reliably differentiates between the healthy, 2-D, 4-D, and 20-D regions. To summarize, the bar plots suggest that polarized Raman spectroscopy is not sensitive enough to detect the onset or the very early stages of demineralization, while thermophotonic imaging reliably detects both. The basis for the enhanced sensitivity of thermophotonic imaging originates in the fact that, unlike Raman spectroscopy, this form of detection belongs to the group of energy conversion methodologies with reduced signal baseline and enhanced dynamic range advantages, where excitation is optical and detection is thermal. By contrast, in Raman spectroscopy excitation and detection are both optical. Finally, Figure 4.14(c) shows the TMR profiles obtained at the center of the artificial caries created in S1 and S2 as the verification of the comparison made in this section. It can be seen that both of the artificially generated caries of S1 are truly in the initial stages of formation and can easily be arrested and remineralized, but the caries in S2 has a significantly large mineral loss near the surface.
Chapter 5

Thermal-wave radar

One major drawback of the frequency-domain photothermal radiometry, or alternatively in 2-D the thermophotonic lock-in imaging, is the compromise one has to make between dynamic range (probing depth) and the depth resolution. While low-frequency thermal waves can “see” deep into the sample (on the order of millimeters), they lack the desired depth resolution as seen in Figure 4.10(c). High-frequency thermal waves, on the other hand, display exactly the opposite behavior due to the reduced thermal diffusion length and thermal wavelength. The thermal-wave radar technique incorporates linear frequency modulation (chirp) excitation through radar matched filtering to maintain good resolution and depth range inside a sample. The first application of correlation and spectral analysis to a photothermal-wave system using frequency chirps was introduced by Mandelis et al. in a series of papers in 1986 [93][94][95]. More recently, Mulaveesala and Tuli [96] and Tabatabaei and Mandelis [97][98] applied this methodology to thermal-wave non-destructive testing. This chapter introduces the TWR theory in a wide range of materials and experimentally demonstrates the advantages of chirped modulation and introduces a modality of thermal-wave radar based on an emissivity-normalized, higher-dynamic-range contrast parameter known as cross-correlation phase. All the experimental data presented in this chapter were acquired in a single detector element system (TWR system) shown in Figure 2.9. Applications of the TWR theory to a thermophotonic imaging system are discussed in chapter 6. This chapter is based on a provisional patent filed on the developed technology [78] and a paper published in Review of Scientific Instruments [97].

5 Thermal-wave radar

5.1 Theory

5.1.1 Opaque materials

The concept of matched filtering and the experimental setup for thermal-wave radar are presented in sections 2.4 and 2.6, respectively. In this section, the analytical response of an opaque material to linear frequency modulated excitation is presented. A linear frequency modulated waveform \( f(t) \), such as an LFM laser beam, is shown in Figure 2.6(a) and can be mathematically expressed as (2.4).
Figure 5.1. A schematic model of the steel sample discussed in section 2.9.1 as an example of an opaque material.

For a finitely thick opaque sample, if one assumes an insulated rear surface and a linear frequency modulated heat flux on the front surface (Figure 5.1), a simplified version of the associated heat transfer problem can be obtained. The justification of one dimensional heat flow is discussed in section 4.3. In order to solve the boundary value problem, we apply the infinite integral Fourier transform to the time domain problem, assuming a zero value for the AC temperature for $t < 0$. As a result, the frequency domain boundary-value problem can be formulated as follows:

$$\frac{\partial^2 \theta(z, \omega)}{\partial z^2} - \frac{i \omega}{\alpha} \theta(z, \omega) = 0$$  \hspace{1cm} (5.1)

$$-\kappa \frac{\partial \theta(z, \omega)}{\partial z} \bigg|_{z=0} = \mathcal{F}\{f(t)\} = F(\omega)$$  \hspace{1cm} (5.2)

$$-\kappa \frac{\partial \theta(z, \omega)}{\partial z} \bigg|_{z=L} = 0$$  \hspace{1cm} (5.3)

$$\theta(z, \omega) = T(z, \omega) - T_\infty$$
where $\alpha$, $\kappa$, $F(\omega)$, and $T_e$ are thermal diffusivity, thermal conductivity, the incident LFM laser beam spectrum, and sample equilibrium temperature, respectively. Equation (5.1) is a homogeneous second order ordinary differential equation which can be solved by finding the solutions of the corresponding characteristic equation. Furthermore, by applying the boundary conditions (Eqs. (5.2) and (5.3)) to the general solution (5.4), one can obtain the final solution to the frequency domain heat diffusion problem as:

$$\theta(z, \omega) = A(\omega, l) \exp(\sigma z) + B(\omega, l) \exp(-\sigma z)$$

(5.4)

$$A(\omega, l) = \frac{F(\omega)}{\kappa \sigma} \frac{\exp(-2l \sigma)}{1 - \exp(-2l \sigma)}$$

(5.5)

$$B(\omega, l) = \frac{F(\omega)}{\kappa \sigma} \frac{1}{1 - \exp(-2l \sigma)}$$

(5.6)

The time domain thermal-wave field inside a finite thickness sample can be obtained by taking the numerical inverse Fourier transform of (5.4). The thermal diffusion length plays a key role in thermal-wave methodologies since it corresponds to the approximate maximum imaging depth at a specific modulation frequency. By definition, the thermal diffusion length decreases as the modulation frequency increases. As a result, when using an LFM laser beam, we do not expect the high frequency components of a chirp to penetrate as deeply as their low frequency counterparts. The radiometric signal measured at the interrogated surface of a finitely thick opaque sample subjected to LFM excitation can finally be calculated as:

$$s(t) = \mathcal{F}^{-1} \left\{ \bar{\mu}_{IR} \int_0^l \theta(z, \omega) \exp(-\bar{\mu}_{IR} z) dz \right\}$$

(5.7)

where $\bar{\mu}_{IR}$ is the average infrared absorption coefficient over the detection wavelength range. Inserting (5.4) into (5.7) and integrating through the sample thickness yields:

$$s(t) = \mathcal{F}^{-1} \left\{ S(\omega) \right\} = \mathcal{F}^{-1} \left\{ \left( \frac{A(\omega, l) \mu_{IR}}{\sigma - \mu_{IR}} \right) e^{i(\sigma - \mu_{IR})} - 1 \right\} + \left( \frac{B(\omega, l) \mu_{IR}}{\sigma + \mu_{IR}} \right) \left[ 1 - e^{-i(\sigma + \mu_{IR})} \right] \right\}$$

(5.8)
For an opaque material laser absorption takes place at the sample surface. Therefore, letting $\mu_{ir} \to \infty$, the infrared detector signal in a sample having a finite thickness $l$, can be obtained as:

$$S(t) = \mathcal{F}^{-1}[A(\omega, l) + B(\omega, l)]$$

(5.9)

As a result of thermal-wave damping, even a finite thickness sample can be taken as semi-infinite with high frequency thermal-waves. The simplest thermal-wave case of a semi-infinite solid can be obtained analytically from the boundary-value problem of (5.4) upon replacing the boundary condition at $z=l$ with boundedness at $z \to \infty$, i.e. $\Theta(\infty, \omega) = 0$. Subsequently, the resulting frequency domain temperature field $\Theta(z, \omega)$ and radiometric signal spectrum can be obtained as:

$$\Theta(z, \omega) = C(\omega) \exp(-\sigma z)$$

(5.10)

$$C(\omega) = \frac{F(\omega)}{\kappa \sigma}$$

(5.11)

$$S(\omega) = \mu_{ir} \int_0^\infty \Theta(z, \omega) \exp(-\mu_{ir} z)dz = \frac{C(\omega) \mu_{ir}}{\sigma + \mu_{ir}}$$

(5.12)

Letting $\mu_{ir} \to \infty$ yields the well-known radiometric signal for metallic semi-infinite solids:

$$S(\omega) = C(\omega)$$

(5.13)

Note that by letting $l \to \infty$ in (5.9) the solution for the finitely thick opaque sample reduces to that of the semi-infinite sample (5.13), as expected.

### 5.1.2 Transparent materials

The case of sub-surface optical absorbers in the presence of absorbing or non-absorbing overlayers is an important and challenging one for thermal-wave studies because the damped nature of the diffusive-waves from the underlayer tends to be dominated or completely masked by the conductive pathway across the overlayer and/or overlayer absorption contributions to the
thermal-wave detected at the front surface. To assess the performance of the thermal-wave radar in this configuration, the heat diffusion problem was solved for a transparent solid (a glass) which is illuminated at its front surface \((z=0)\) and coated with an opaque thin layer at its rear surface.

**Figure 5.2. A schematic model of transparent samples discussed in section 2.9.2.**

In this case, optical radiation is absorbed on the back surface of the solid \((z=l)\), Figure 5.2. Because it is delayed with respect to the excitation beam, a (phase shifted) thermal-wave is detected at the surface. The boundary-value problem of (5.4) must be modified by assuming thermal insulation at the front surface and frequency modulated heat flux at \(z=l\). The frequency domain temperature field and radiometric signal spectrum can then be obtained as:

\[
\theta(z,\omega) = E(\omega,l)\left[\exp(\sigma z) + \exp(-\sigma z)\right]
\]

\[(5.14)\]

\[
E(\omega,l) = \frac{F(\omega)}{\kappa \sigma} \cdot \frac{1}{\exp(-l\sigma) - \exp(l\sigma)}
\]

\[(5.15)\]
\[
S(\omega) = \frac{E(\omega,l) \bar{\mu}_{\text{IR}}}{\sigma - \bar{\mu}_{\text{IR}}} \left\{ \exp\left[l(\sigma - \bar{\mu}_{\text{IR}})\right] - 1 \right\} + \frac{E(\omega,l) \bar{\mu}_{\text{IR}}}{\sigma + \bar{\mu}_{\text{IR}}} \left\{ 1 - \exp\left[-l(\sigma + \bar{\mu}_{\text{IR}})\right] \right\} \quad (5.16)
\]

### 5.1.3 The matched filter/cross-correlation response

The radiometric signal spectra are modeled for the special cases of finite, semi-infinite, and transparent solids (\(S(\omega)\) in equations (5.9), (5.13), and (5.16)). The thermal-wave radar output, \(CC(\tau)\), can be obtained by substituting the filter frequency response term in equation (2.17), \(R(\omega)\), with the applied optical heat source spectrum, \(F(\omega)\), and the received signal spectrum term, \(S(\omega)\), with the modeled radiometric signal spectrum of each foregoing solid sample and geometry:

\[
CC(\tau)_{\text{finite}} = \mathcal{Z}^{-1}\left\{ A(\omega,l) + B(\omega,l) \cdot F(\omega)^* \right\} \quad (5.17)
\]

\[
CC(\tau)_{\text{semi-infinite}} = \mathcal{Z}^{-1}\left\{ C(\omega) \cdot F(\omega)^* \right\} \quad (5.18)
\]

\[
CC(\tau)_{\text{transparent}} = \mathcal{Z}^{-1}\left\{ \frac{E(\omega,l) \mu_{\text{IR}}}{\sigma - \mu_{\text{IR}}} \left( \exp\left[l(\sigma - \mu_{\text{IR}})\right] - 1 \right) \right. \\
+ \left. \frac{E(\omega,l) \mu_{\text{IR}}}{\sigma + \mu_{\text{IR}}} \left( 1 - \exp\left[-l(\sigma + \mu_{\text{IR}})\right] \right) \right\} \cdot F(\omega)^* \quad (5.19)
\]

### 5.1.4 Thermal-wave radar subtractive mode response

Unlike propagating hyperbolic wave fields, parabolic diffusive waves do not possess wavefronts. Furthermore, in these wave fields the diffusive energy transport is strongly dispersive [80]. As a result, in opaque solids, the coherent energy accumulation due to the presence of underlayers can be effectively masked by the dominant front surface generated diffusive waves. One way to enhance the effect of underlayers is by omitting the effect of the dominant front surface
generated diffusive waves through subtraction of the full radiometric signal of a semi-infinite opaque sample from that of a finite sample, provided that the same optical excitation waveform is used for both samples. Equation (5.20) illustrates this concept mathematically:

\[ \Delta S(\omega) = [A(\omega, l) + B(\omega, l)] - C(\omega) = \frac{F(\omega)}{\kappa \sigma} \cdot \frac{2 \exp(-2l\sigma)}{1 - \exp(-2l\sigma)} \tag{5.20} \]

One can find the TWR cross-correlation curve due to the coherent energy accumulation of underlayers as:

\[ \Delta CC(\tau) = \Im^{-1} \left\{ \frac{F(\omega)}{\kappa \sigma} \cdot \frac{2 \exp(-2l\sigma)}{1 - \exp(-2l\sigma)} \cdot F(\omega)^* \right\} \tag{5.21} \]

Mathematically, (5.21) is the definition of the TWR subtractive mode. It will be shown in the following section that compared to TWR, the TWR subtractive mode is a more sensitive method for detecting deep inhomogeneities since the dominating fundamental forward diffusive term is eliminated.

5.2 Theoretical simulations

5.2.1 Opaque materials

Figure 5.3 depicts the energy spectral density of simulated radiometric signals, (5.9), from steel samples with various thicknesses (l) along with the energy spectral density of the simulated radiometric signal of a semi-infinite steel, (5.13). For comparison, the energy spectral density of the optical excitation waveform, \( F(\omega) \), is also included.
The simulated ESD of the radiometric signal from a steel sample with several thicknesses and a semi-infinite surface. The ESD of the incident LFM excitation is also included for comparison.

The spectral energy of the radiometric response from the steel sample surface can be divided into two parts: The energy due to excitation-generated diffusive waves at the sample surface (surface energy absorption) and the coherently accumulated energy due to back-interface interacting thermal-waves. Surface energy absorption increases the sample surface temperature and therefore produces a radiometric signal whose spectral energy distribution exhibits the damped behavior expected under the envelope of diffusion (a low pass filtering effect). On the other hand, the energy due to back-interface interacting thermal-waves in finitely thick opaque solids is delayed (phase-shifted) with respect to the surface absorption energy part. In a semi-infinite sample, the radiometric signal energy (Figure 5.3, curve 7) is only due to surface energy absorption and no delayed back-interface interacting thermal-wave contribution is expected. As a result, a relatively uniform ESD is observed, characterized by the diffusive low-pass filtering profile (damping).

The surface energy absorption part is also present in the ESD of finite thickness opaque samples. However, the interface interacting thermal-wave part of the ESD in these samples (both back and front interfaces) contribute to the resulting diffusion-wave frequency spectrum. In this sense, the delayed portion of the ESD (subtractive part) becomes a function of sample thickness and can be used as a sensitive parameter to distinguish various thicknesses. According to Figure 5.3 as the
sample thickness decreases the subtractive mode energy density increases. The energy increase is more pronounced at lower frequencies as low frequency components are less damped during their propagation.

**Figure 5.4.** The steel sample simulated surface temperature profile due to contributions of diffusive-waves from several thicknesses with the fundamental forward thermal wave, subtracted. The semi-log figure inset magnifies the concept of delayed contribution due to an increase in sample thickness. The inset axes are the same as those of the figure.

The concept of interface interacting (depleted or accumulated) thermal-waves can be better understood from Figure 5.4. This figure is the result of subtraction of the surface thermal-waves of several finite thickness samples from that of a semi-infinite sample. It can be observed that due to the damping behavior of thermal-waves, as the sample thickness ($\ell$) increases the coherent energy accumulation from the interface confinement to the detected thermal-wave at the sample surface decreases; moreover, this energy contribution becomes more delayed (Figure 5.4 inset), since increasing the sample thickness results in increased transport distance for thermal waves. The concept of delayed contributions from subsurface thermophysical discontinuities is an important feature that can be used to distinguish various thicknesses and subsurface defects, and lies at the heart of the thermal-wave radar.
Figure 5.5. Theoretical TWR signals for 100-μm thick and semi-infinite steel samples, using equations (5.17) and (5.18). Chirp data: T=10 s, $f_1=1$ Hz, $f_2=5$ Hz. The inset magnifies the time interval checked in the main plot.

Figure 5.5 shows the simulated TWR/cross-correlation output signals for 100 μm thick and semi-infinite steel samples. It can be observed that as the steel sample thickness increases from 100 μm to infinity, the cross-correlation peak delay time ($\tau_p$) decreases from 83 to 39 ms, as magnified in the figure inset. In addition, since the radiometric signal amplitude decreases as the sample thickness increases, the actual cross-correlation peak values are significantly different between these samples (32% difference). The theoretical TWR cross-correlation peak delay times are plotted as a function of sample thickness in Figure 5.6 (solid line). In this plot, it is clearly seen that delay time decreases as the thickness increases. This is due to the fact that increasing the sample thickness dampens the contribution of thermal-waves from depths other than the surface, such as confined thermal-waves at the steel-air back interface. As a result, the ESD of the TWR signal becomes more similar to that of the semi-infinite sample (Figure 5.3) and the TWR cross-correlation delay time approaches that of the semi-infinite sample (Figure 5.6 dotted line), as expected. The cross-correlation peak delay time of a semi-infinite sample has a non-zero value (39 ms) due to the well-known time shift of the surface temperature oscillation with respect to the incident thermal-wave flux [80].
Figure 5.6. Steel sample simulated TWR cross-correlation peak delay times for several thicknesses using TWR (solid line) and TWR in subtraction mode (dashed line). The dotted base line indicates the TWR delay time corresponding to a semi-infinite steel sample. Circles denote the normalized experimental data. Chirp bandwidth = 1-5 Hz, chirp duration = 10 s.

The dashed line in Figure 5.6 represents the theoretical variation of the TWR subtractive mode cross-correlation $\tau_p$ with sample thickness. Compared to the shallow saturation of the full TWR mode, this simulation clearly shows the capability of TWR in resolving delayed contributions from much deeper regions using the subtractive mode to enhance interface-interacting thermal-waves. It can be observed that as the sample thickness increases, the contribution is more delayed due to the increase in thermal-wave conduction distance.

Figure 5.7 shows the effect of excitation chirp properties on the TWR depth detection range. This plot suggests that, at a fixed starting frequency ($f_s$), as frequency modulation sweep rate decreases the theoretical detection range increases; however, at the same time the depth resolution between shallow inhomogeneities decreases. This is manifested by smaller slopes at smaller thicknesses in Figure 5.7. Therefore, the TWR chirp parameters can be tailored to the application to yield optimal results.
Figure 5.7. Theoretical dependence of cross-correlation peak delay time, $\tau_p$, on steel sample thickness using several chirp sweep rates.

5.2.2 Transparent materials

Figure 5.8 shows the simulated front and back surface TWR temperature profiles in a 1-mm-thick back-surface painted glass sample.

Figure 5.8. Simulated AC temperature profile on front (solid line) and back (dashed line) surfaces of a 1-mm-thick back-surface painted glass sample. The figure inset has the same axes as the main figure and illustrates the concept of delayed contribution of the two thermal-waves.
The front surface AC temperature profile lags behind the back surface AC temperature profile (Figure 5.8 inset), as expected. This phase lag is directly related to the sample thickness and thermal properties of glass. Furthermore, not all frequency components in the optically generated heat flow can reach the front surface (detection plane) because the overlayer acts as an efficient low-pass filter. Moreover, the absolute (DC) temperature of the sample constantly increases as a result of the continuing inflow of thermal energy accumulating in the solid according to the theoretical model.

![Image: Figure 5.9. Simulated TWR cross-correlation peak delay time for various glass sample thicknesses. Chirp data: \( f_s = 0.1 \) Hz, \( f_c = 5 \) Hz, duration = 10 s, \( \alpha = 0.45 \times 10^{-6} \) \( m^2/S \). The diffusive equation is fitted to the simulation data.]

Open circles in Figure 5.9 depict the theoretically simulated TWR cross-correlation peak delay times as a function of the thickness of the glass sample in the configuration of Figure 5.2. Comparison with the solid line in Figure 5.6 shows that a completely different trend is observed for the glass samples. In fact, the curve is more similar to that of the TWR subtractive mode (Figure 5.6, dashed curve). The reason for this similarity is that in the case of transparent solids the ESD of the radiometric signal is only due to the contribution of thermal-waves which diffuse away from the absorbing back surface; the fundamental contribution from front-surface generated forward thermal-waves is missing entirely due to transparency.
The finite delay time $\tau_p$ at $l=0$ is a consequence of the heat flux-temperature rise delay discussed in conjunction with the baseline of Figure 5.6. The theoretical simulations (Figure 5.6 and Figure 5.9) confirm that the matched filtering signal processing method used in the TWR is capable of measuring the delay time between the optical excitation waveform and the PTR response in both opaque and transparent solids.

### 5.2.3 FD-PTR vs. TWR

A different set of simulations was designed to verify the higher dynamic range of the TWR method compared to the conventional FD-PTR method. In these simulations, the FD-PTR (both phase and amplitude channels) and TWR responses of steel samples having thicknesses between 100 μm and 5 mm were simulated. To make the simulations more realistic the simulated radiometric signals were superposed with stochastic noise (SNR=1) before the signal processing step. Ten accumulations were used for each simulation to evaluate the robustness of each method. Finally, the mean of the outputs (FD-PTR amplitude and phase, and cross-correlation peak delay times in both full TWR and TWR subtractive modes) were plotted against the sample thickness, where the error bars indicate the standard deviation for each simulation.

---

**Figure 5.10.** Simulations of the variation in FD-PTR (a) amplitude and (b) phase values with steel sample thickness. The input radiometric signal was superposed with stochastic noise (SNR=1) prior to signal processing. Each point was simulated 10 times to measure the standard deviation. The maximum detection depth is indicated on the figure.
Figure 5.10(a) and Figure 5.10(b) show the simulation results of FD-PTR amplitude and phase channels, respectively. It can be observed that the amplitude channel can detect thicknesses unambiguously only up to 1 mm. On the other hand, the phase channel exhibits higher dynamic range and saturates to the −45° phase of the semi-infinite mode at thicknesses greater than 1.79 mm.

![Graphs of TWR delay time vs. thickness](image)

**Figure 5.11.** Simulations of the variation in cross-correlation peak delay time, $\tau_p$, with steel sample thickness: (a) TWR; (b) TWR subtractive mode. Simulation conditions are included in the figure insets.

Figure 5.11(a) and Figure 5.11(b) depict the simulation results of the TWR and TWR subtractive mode, respectively. In the case of the full TWR method, the dynamic range is improved by 112% compared with the FD-PTR amplitude channel and 18.43% compared to the FD-PTR phase channel. The TWR subtractive mode shows dynamic range enhancement of 171% over the FD-PTR amplitude channel and 51.4% over the FD-PTR phase channel. The dynamic range enhancement of TWR over FD-PTR can be attributed to the high SNR in the matched-filtering method when used together with pulse compression techniques such as the LFM. Using the conventional radar SNR expressions for calculating the TWR signal-to-noise ratio will result in a very high SNR value (4 orders of magnitude greater than that of FD-PTR) which is not realistic. The reason for this discrepancy is that the conventional radar formulas do not account for the damped nature of thermal-waves. Therefore, a more realistic formula for both FD-PTR and TWR is used as suggested by other investigators [83]:
where the output of FD-PTR can be either amplitude or phase and the output of TWR is the cross-correlation peak delay time. Using (5.22), the following average SNR values can be obtained for the simulation data depicted in Figure 5.10 and Figure 5.11:

\[ SNR_{FD\text{-PTR}}^{\text{Amp}} = 27.96 \]
\[ SNR_{FD\text{-PTR}}^{\text{Phase}} = 28.95 \]
\[ SNR_{TWR} = 77.90 \]

The SNR of TWR is 178% higher than that of the FD-PTR amplitude channel and 169% higher than that of the FD-PTR phase channel. The experimental results further demonstrate the enhanced depth-resolution and dynamic range of TWR compared to FD-PTR.

5.3 Experiments

5.3.1 Opaque sample

Figure 5.12 shows the results of the frequency-scanned PTR experiments carried out on the steel sample over several holes.

![Graphs showing PTR amplitude and phase variation](image)

Figure 5.12. (a) Log-log amplitude and (b) semi-log phase variation of experimental FD-PTR values with modulation frequency over subsurface steel sample holes with various depths (as indicated in the inset).
Figure 5.12(a) shows that for thicknesses above 600 μm \((l>600\mu m)\), the FD-PTR amplitudes are straight lines coinciding with the semi-infinite sample amplitude. For \(l = 1000 \mu m\), the photothermal amplitude at the lowest modulation frequency (1 Hz) deviates from the straight semi-infinite line, but is within a standard deviation of that line. The overlap is more clearly seen in the phase plots, Figure 5.12(b). It is concluded that the FD-PTR amplitude channel cannot detect inhomogeneities located deeper than 600 μm. Deviations from the semi-infinite sample at lower frequencies over sample thicknesses \(l \leq 600\mu m\) confirm the presence of an inhomogeneity in these samples. As the subsurface hole distance to the surface decreases \((l \to 0)\), the frequency at which the FD-PTR graph deviates from that of a semi-infinite steel shifts toward higher values. A similar trend can also be seen in the FD-PTR phase response curves. For \(l > 1000\mu m\), at all scanned frequencies the FD-PTR phase hovers around \(-45^\circ\) which is the theoretical value for a semi-infinite sample; indicating that the FD-PTR phase channel cannot differentiate between these holes and a semi-infinite sample. On the other hand, for \(l < 1000\mu m\), the FD-PTR phase channel has a better dynamic range than the FD-PTR amplitude channel since it can detect samples with identical thicknesses at higher modulation frequencies (shorter thermal diffusion lengths). For example, for \(l = 100\mu m\), the amplitude channel can only recognize the hole at \(f \leq 30Hz\) but the phase channel can resolve the hole at \(f \leq 150Hz\). The better dynamic range of the FD-PTR phase was first reported by Busse and Eyerer [11]. The large error bars in the phase channel at 1 Hz are due to signal cut-off by the AC-coupled pre-amplifier in our experimental setup. Therefore, despite the ability of the phase channel to resolve the \(l = 1000\mu m\) sample, the experimental results at this frequency may be distorted. In summary, the experimental results suggest that neither of the FD-PTR channels can reliably detect subsurface defects (holes) located deeper than 600 μm.

Figure 5.13(a) and Figure 5.13(b) show the results of a conventional FD-PTR spatial scan carried out at 5 Hz along the centerline of shallow \((l = 100, 400, \text{and } 600 \mu m)\) and deep \((l = 1000, 1500, \text{and } 3000 \mu m)\) holes, respectively. Due to the AC-coupled pre-amplifier cut-off, the spatial scan at 5Hz was found to yield optimal results in terms of inhomogeneity detection. The spatial scan clearly detects the shallow but not the deep subsurface holes. These line-imaging results are consistent with Figure 5.12, and show the inability of conventional FD-PTR to detect inhomogeneities deeper than 600 μm.
Figure 5.13. Experimental amplitude and phase of an FD-PTR spatial scan carried out at 5 Hz over thicknesses of (a) 100, 400, and 600 μm; (b) 1000, 1500, and 3000 μm.

Figure 5.14. Experimental TWR spatial scan over thicknesses of (a) 100, 400, and 600 μm; (b) 1000, 1500, and 3000 μm.

Figure 5.14 shows a TWR spatial scan along the same lines as Figure 5.13. The enhanced dynamic range of TWR can be clearly seen when comparing Figure 5.13(b) and Figure 5.14(b). After normalization of the FD-PTR data with theory (matching the theoretical and experimental delay times of the semi-infinite sample), the TWR delay time can be fitted to the theoretical curve, to measure the subsurface depth of the holes (Figure 5.6; solid line, and open circles, respectively).
Figure 5.15. TWR cross-correlation peak delay time, $\tau_p$, image of the steel sample obtained by running a surface raster scan. Scan resolution: $1\text{mm} \times 1\text{mm}$.

Figure 5.15 is a TWR image of the steel sample that was obtained through a 2-D TWR raster scan of the interrogated surface (scan resolution: $1\text{mm} \times 1\text{mm}$). Each pixel represents one laser beam location. The contrast parameter in this image is the cross-correlation peak delay time, $\tau_p$. The image is consistent with Figure 5.14 in that the TWR can detect holes at subsurface depths $1500 \text{ μm}$ or less. This considerable improvement in the dynamic range over conventional FD-PTR is due to the higher SNR of TWR compared to FD-PTR, in agreement with the foregoing theoretical predictions in section 5.2.3.

5.3.2 Transparent samples

Figure 5.16 shows the experimental TWR results for the back-painted glass samples. It should be noted that, for a clear glass sample (not painted on either side), a straight line close to zero indicates that no photothermal signal was obtained from this sample. Therefore, it was concluded that the borosilicate glass used in this experiment was effectively transparent to the wavelength of the excitation beam. On the other hand, for the front-surface painted glass sample, the presence of the TWR cross-correlation maximum at $\tau = 0$ indicates purely surface absorption (no delayed conductive contribution from underlayers). For the back-surface painted glass samples, as the thickness of the glass increases, the corresponding peak delay time also increases (as predicted by theory, Figure 5.9). Furthermore, with an increase in glass thickness the TWR signal is strongly attenuated as thermal-waves decay exponentially throughout the glass thickness.
The TWR peak delay time, $\tau_p$, is characteristic of the depth of the absorber and as a rough approximation can be fitted to the diffusive equation:

$$\tau_p(l) = k_1 l^2 + k_2$$  \hspace{1cm} (5.23)

where $k_1$ and $k_2$ are unknown constants that can be obtained by fitting the diffusive equation to the simulation data. Equation (5.23) indicates that imaging performed at fixed delay times, $\tau_p$, using the TWR will yield a thermal-wave image from a fixed depth, $l$. This feature of the TWR is not matched by conventional FD-PTR: fixed $\tau_p$ spatial scans using the thermal-wave radar can lead to depth-selective subsurface images, whereas FD-PTR scans at fixed frequency only lead to depth-integrated subsurface imaging down to a distance controlled by the thermal diffusion length. The solid line in Figure 5.9 represents the fitted diffusive equation, (5.23), to the theoretically simulated data (Figure 5.9 - ○). It can be observed that the normalized experimental data (Figure 5.9 - ▽) closely follow the simulated trend. As a result, the simulation fitted diffusive equation can be used to decode TWR images and obtain depth information of subsurface absorbers.
5.4 Advantages and limitations of the TWR setup

The first application of cross correlation and spectral analysis to a photothermal-wave system using frequency chirps was introduced by Mandelis et al. in a series of papers in 1986 [93][94][95]. In this research the photothermal response of opaque media to LFM excitation was investigated through the deflection of a probe beam using the mirage effect, Figure 1.2. Although this research laid the conceptual basis of TWR, applications of the method were limited as the probe beam could only be applied to flat surfaces and samples had to be placed in water to create higher refractive index gradients. On the contrary, the developed TWR methodology does not require a coupling medium and is non-contact, since inspection is carried out on the infrared radiation emanating from the sample (radiometry). More recently, in 2006 Mulaveesala and Tuli [96] applied LFM matched filtering to thermal-wave non-destructive testing of industrial opaque samples. However, their research was only limited to the cross correlation amplitude and therefore was prone to variations in the sample emissivity and/or laser intensity. In our system, this problem is resolved by using the cross correlation peak delay time and phase channels, section 2.5.

Despite the promising experimental results of the TWR system, the experimental setup designed for TWR had certain instrumentational limitations. In analogy to the theoretical ESD plots of the TWR responses of the steel sample holes (Figure 5.3), Figure 5.17 presents the ESD of the radiometric signals obtained in the TWR setup. It can be seen that the spectral energy distribution exhibits the damped behavior expected under the diffusion envelope (the low pass filtering action), as expected from the theory. However, the signals are highly distorted. The distortion of the ESD is due to the following causes:

1. Poor SNR of the radiometric signal
2. The low cutoff frequency of the AC preamplifier used in the setup

Since an external analog function generator was used in the TWR setup, the initial phase of the reference chirp signal was not fixed. Therefore, dynamic signal averaging, similar to that used in the thermophotonic lock-in imaging setup, was not possible and the acquired radiometric signal had a poor SNR. Moreover, the AC-coupled preamplifier used in the setup is an effective high-pass filter with a low cutoff frequency and as a result the low-frequency components of the
radiometric signals were attenuated more than their high-frequency counterparts. To overcome these limitations and optimize the experimental setup, the thermophotonic lock-in imaging system was modified to allow for thermal-wave radar imaging, henceforth referred to as thermophotonic radar imaging.

Figure 5.17. The experimental ESD of the radiometric signal from the steel sample with blind holes. The legend shows the wall thickness of each hole.
In order to overcome the instrumental limitations of the TWR setup and investigate the performance of the method in an imaging platform, the thermophotonic lock-in imaging system was modified to enable thermophotonic radar imaging. This chapter presents the key advantages of the thermal-wave radar method as a depth selective imaging technique and demonstrates the possibility of obtaining thermal-wave coherence tomographic images. The key advantages are verified using non-biological samples as well as dental samples. Furthermore, binary phase coding, as an alternative pulse compression technique to linear frequency modulation, is introduced, theoretically modeled, and experimentally investigated to show how it gives rise to depth selective thermal coherence tomography. This chapter is based on a provisional patent filed on the developed technology [78] as well as papers published in Applied Physics Letters [98] and Physical Review Letters [99].

6 Thermophotonic radar imaging

6.1 Opaque sample

Figure 6.1, Figure 6.2, and Figure 6.3 show the thermophotonic radar images of holes H2, H3, H4, and H5 using 10s long chirps. In these figures, while the start frequency and duration of the chirp is kept the same ($f_1 = 0.1$ Hz), the chirp end frequency, or alternatively the chirp sweep rate, is increased to experimentally validate the theoretical predictions of Figure 5.7. It can be seen that, in agreement with the theory, at low sweep rates (Figure 6.1) neither the CC peak delay time nor the CC phase has sufficient depth resolution to differentiate between H2 and H3 (poor axial resolution at shallow depths). At high sweep rate, on the other hand, the depth resolution is improved but at the cost of reduced maximum probing depth, Figure 6.3. The intermediate sweep rate, however, seems to be optimal for the steel sample as all holes are detected in the CC phase image with acceptable depth resolution, Figure 6.2. The other interesting point is the better sensitivity of CC phase compared to CC peak delay time due to more robustness with respect to noise, as explained in section 2.5. The spectra of the radiometric signals over the holes and the semi infinite part are in excellent agreement with the theoretical predictions of Figure 5.3 and can therefore be used for quantification of material properties.
through multi-parameter fitting. The comparison of these spectra with those obtained in the TWR system, Figure 5.17, proves the low SNR and the signal high-pass filtering are completely resolved in the thermophotonic radar imaging system.

Figure 6.1. Steel sample thermophotonic radar CC (a) peak amplitude, (b) peak delay time, and (c) CC phase images using 10s long chirps with $f_1 = 0.1\text{Hz}$ and $f_2 = 2\text{Hz}$ over H2 (top-right), H3 (top-left), H4 (bottom-left), and H5 (bottom-right) holes. Panels (d), (e), and (f) are the corresponding profiles along the lines passing through the center of the top (blue line) and bottom (green line) row holes. (g) ESD of the thermophotonic signal over the holes and the semi infinite part (center of the image).
Figure 6.2. Steel sample thermophotonic radar CC (a) peak amplitude, (b) peak delay time, and (c) CC phase images using 10s long chirps with $f_1 = 0.1$Hz and $f_2 = 5$Hz over H2 (top-right), H3 (top-left), H4 (bottom-left), and H5 (bottom-right) holes. Panels (d), (e), and (f) are the corresponding profiles along the lines passing through the center of the top (blue line) and bottom (green line) row holes. (g) ESD of the thermophotonic signal over the holes and the semi infinite part (center of the image).
Figure 6.3. Steel sample thermophotonic radar CC (a) peak amplitude, (b) peak delay time, and (c) CC phase images using 10s long chirps with $f_1 = 0.1\text{Hz}$ and $f_2 = 10\text{Hz}$ over H2 (top-right), H3 (top-left), H4 (bottom-left), and H5 (bottom-right) holes. Panels (d), (e), and (f) are the corresponding profiles along the lines passing through the center of the top (blue line) and bottom (green line) row holes. (g) ESD of the thermophotonic signal over the holes and the semi infinite part (center of the image).
6.2 Quantification of the thermophotonic radar images

To show the possibility of extracting material properties from the thermophotonic images, thermophotonic radar imaging was carried out on the steel sample. Subsequently, the temporal data of a semi-infinite point were used together with equation (5.13) to calculate the transfer function of the system. As discussed in section 2.8, a pre-existing multi-parameter fitting program, based on the simplex downhill algorithm, was modified to account for the thermal-wave radar theory for finitely thick opaque materials, (5.9). Finally, the experimental spectra of the radiometric signals over the H2, H3, H4, and H5 holes were convolved with the calculated transfer function and fed to the fitting program to estimate the hole wall thickness, \( l \), as well as the sample’s thermal diffusivity and thermal conductivity.

![Figure 6.4. Experimental (points) and fitted (solid lines) spectra of the radiometric signals over the H2 (400 µm), H3 (600 µm), H4 (1000 µm), and H5 (1500 µm) holes.](image)

Figure 6.4 shows the experimental (points) and fitted (lines) spectra of the radiometric signals obtained over the blind holes. The fitting accuracy is improved as the wall thickness increases...
from 400 (H2) to 1500µm (H5). The same trend is observed for higher frequencies in each spectra. Table 3.1 includes the results of the multi-parameter fitting. The actual wall thickness values were measured with a vernier caliper. The fitting limits provided below the table were broken into 100 divisions and the best combination was found by minimizing a residual function ( |Experimental value – Theoretical value| ).

Table 6.1. Material properties estimated through multi-parameter fitting of the experimental data to the theory.

<table>
<thead>
<tr>
<th>Hole</th>
<th>Actual thickness $l_A$ (µm)</th>
<th>Estimated thickness $l_E$ (µm)</th>
<th>Error $(l_A-l_E)/l_A$</th>
<th>Estimated $\alpha$ (m²s⁻¹)</th>
<th>Estimated $\kappa$ (Wm⁻¹K⁻¹)</th>
<th>Fitting time (hrs)</th>
</tr>
</thead>
<tbody>
<tr>
<td>H2</td>
<td>410</td>
<td>550</td>
<td>-34.15%</td>
<td>1.89365×10⁻⁵</td>
<td>58.6423</td>
<td>56.72</td>
</tr>
<tr>
<td>H3</td>
<td>580</td>
<td>466</td>
<td>+19.66%</td>
<td>1.03287×10⁻⁵</td>
<td>52.7667</td>
<td>38.88</td>
</tr>
<tr>
<td>H4</td>
<td>980</td>
<td>785</td>
<td>+19.90%</td>
<td>1.28951×10⁻⁵</td>
<td>49.1624</td>
<td>35.87</td>
</tr>
<tr>
<td>H5</td>
<td>1480</td>
<td>1218</td>
<td>+17.70%</td>
<td>1.26319×10⁻⁵</td>
<td>44.8141</td>
<td>33.57</td>
</tr>
</tbody>
</table>

* The following limits were used for the fitting: $l$ [20 - 2500], $\alpha$ [0.8×10⁻⁵ - 8×10⁻⁵], $\kappa$ [20 80]

Comparison of the actual and estimated wall thicknesses show that except for H2 the fitting program yields acceptable estimates. The tabulated thermal diffusivity and thermal conductivity of AISI 1010 steel, 1.26319×10⁻⁵ (m²s⁻¹) and 49.8 (Wm⁻¹K⁻¹) respectively[100], are also consistent with the estimated values. Although the quantitative aspect is not the focus of this thesis, the quantitative study shows that a single thermophotonic radar experiment can be used to estimate material properties and defect depths.

6.3 CC amplitude vs. CC phase

As mentioned earlier, in frequency-domain photothermal radiometry/thermography the most reliable contrast parameter is the phase, or alternatively the delay, channel. The reason for the improved reliability is the fact that the phase/delay channel is emissivity and laser-intensity-fluctuation normalized. Figure 6.5 shows the thermophotonic radar images for the transparent
The sample simulates two iso-depth absorbers with different light absorption coefficients. Parts a, b, and c are the images obtained using cross correlation amplitude, peak delay time, and phase, respectively, and parts d, e, and f show their horizontal mean profiles, respectively. It can be seen that the amplitude channel is representative of the amount of energy absorption by the two equally deep absorbers (green and black paints), yielding significantly different CC amplitude values. Consequently, the amplitude channel is not a true measure of the depth of the absorber. However, the CC peak delay time and phase values are linked to the true depth of the absorbers since they maintain the same value over the two absorbers regardless of their absorption coefficients. Nevertheless, in terms of SNR the CC amplitude channel is significantly stronger than the peak delay time and phase channels and therefore the amplitude images should always be used to complement the information obtained from the phase and peak delay time images.

Figure 6.5. Thermophotonic radar imaging of the transparent sample shown in Figure 2.11(a) using (a) CC amplitude; (b) CC peak delay time; (c) CC phase and their mean horizontal profiles; (d), (e), and (f), respectively. Chirp parameters: 0.01-1 Hz in 6s.

6.4 Improving the performance in dental diagnostics

As mentioned earlier, the motivation behind the TWR technique was to overcome the compromise one has to make between dynamic range (probing depth) and depth resolution in a conventional FD-PTR/LIT system through generation of depth-selective results. Figure 6.6 includes the thermophotonic radar and lock-in images of sample T1. Similarly to sample S1, the
controlled demineralization protocol was used to create two treatment windows on the interrogated surface of this sample. However, the left and right treatment windows were treated for 10 and 20 days, respectively.

Figure 6.6. Thermophotonic radar and phase lock-in imaging of early dental caries. (a) CC amplitude image; (b) CC peak delay time image; (c) CC phase image (Chirp parameters: 0.01-1 Hz in 6s). LIT phase images at (d) 0.01 Hz and (e) 1 Hz; (f) TMR mineral profiles at points 1-3 indicated in figure (c), figure inset depicts the amount of mineral loss.

Figure 6.6(a) to Figure 6.6(c) show the CC amplitude, peak delay time, and phase images using a 6 second chirp starting from 0.01 Hz and ending at 1Hz. In general, tooth demineralization results in higher porosity and leads to more light scattering and shallower absorption compared to intact enamel, thereby increasing the amplitude of thermal waves and shifting the thermal-wave centroid closer to the surface than those of the healthy spots. As a result, the artificially created caries is clearly detectable in all CC images. However, due to the emissivity-normalized nature of peak delay time and phase channels more details can be resolved in Figure 6.6(b) and Figure 6.6(c) compared to Figure 6.6(a). The higher contrast of the right treatment window indicates greater mineral loss due to the additional treatment days. These results suggest that the thermophotonic radar imaging can probe deeply and at the same time maintain a good depth resolution. The thermophotonic lock-in phase image performed at the chirp’s start frequency (0.01 Hz), on the other hand, is too diffuse and suffers from low depth resolution, Figure 6.6(d), while the phase images obtained at the chirp’s end frequency (1Hz) cannot “see” deeply enough
to show the additional mineral loss in the right treatment window, Figure 6.6(e). Figure 6.6(f) depicts the TMR mineral profiles along points 1-3 marked in Figure 6.6(c), provided as proof of relative mineral loss within the treatment windows.

6.5 Binary phase coding - thermal coherence tomography

The intention of this section is to show how non-propagating (parabolic) diffusion-wave energy fields can exhibit energy localization normally encountered in propagating hyperbolic wavefields. To achieve this goal, a new thermal-wave binary phase code generation scheme is used together with the matched-filter signal processing method. Obtaining localized responses from a diffusion wave field not only results in an improvement in axial resolution, but also allows the deconvolution of individual responses of superposed axially discrete sources and leads to depth-selective rather than depth-integrated images. These concepts are not limited to thermal waves, but are of very broad interest to all parabolic diffusion-wave fields.

6.5.1 Theory

The role of pulse compression techniques in matched filtering has been discussed in section 2.4. Thermal-wave radar theory incorporates linear frequency modulation as the pulse compression technique to generate depth selective images; however, other pulse compression techniques are also available. Binary phase coding is one such alternative that is widely used in communication and radar sciences due to its simplicity and low side-lobe level. While the frequency chirp modulation maintains a relatively constant power within the chirp frequency range (wide-band detection as shown in Figure 2.6(b)), BPC modulation results in distinct energy localization advantages, especially with diffusive fields.

Figure 6.7. Schematic model of a black body absorber in a turbid medium.
The thermophotonic response (Planck radiation emission) of a non-opaque turbid medium to a BPC excitation can be obtained through coupling of the optical and thermal-wave fields, where the total optical field (coherent + scattered) is the source of thermal-wave generation. As a simple example, this section analytically investigates the response of black body absorbers in a turbid medium. That is, the BPC excitation is applied to the surface \( z=0 \) of a turbid medium with known scattering and absorption coefficients \( (\mu_s \text{ and } \mu_a) \), respectively, where, after interaction with the turbid medium, the attenuated light is completely absorbed at \( z=l \) and thermal waves are generated, Figure 6.7. The energy fluence of the one-dimensional, collimated beam incident on a homogeneous scattering and absorbing medium can be expressed by (3.3). Therefore, the thermal-wave problem can be formulated by adding a depth dependent source term to the heat diffusion differential equation (due to absorption by the medium) as well as an attenuated heat source at \( z=l \) through a boundary condition:

\[
\begin{align*}
\left( \frac{\partial^2 \theta(z;\omega)}{\partial z^2} - \sigma^2 \theta(z;\omega) \right) &= -\frac{\mu_a}{\kappa} I(z) F(\omega) \\
-\kappa \frac{\partial \theta(z;\omega)}{\partial z} \bigg|_{z=0} &= 0 \quad \text{(I)} \\
-\kappa \frac{\partial \theta(z;\omega)}{\partial z} \bigg|_{z=l} &= I(l) F(\omega) \quad \text{(II)} \\
\left\{ \theta(z;\omega) = \mathcal{F}(T(z,t)-T_\infty) \right\}
\end{align*}
\]

where \( \kappa \) and \( F(\omega) \) are thermal conductivity and the spectrum of the applied BPC excitation (2.15), respectively, and \( \sigma = \sqrt{i\omega/\alpha} \) is the complex wavenumber. Note the similarity of (6.1) to the Helmholtz equation of hyperbolic wave-fields. Equation (6.1) is an ordinary non-homogenous differential equation whose solution can be found through summation of its general and particular solutions, \( y = y_c + y_p \).

\[
y_c: \quad \theta(z;\omega) = A \exp(\sigma z) + B \exp(-\sigma z) \\
y_p: \quad \theta(z;\omega) = C \exp(-\mu_{eq} z) + D \exp(-\mu z)
\]

Inserting \( y_p \) terms one by one into (6.1) we get:
\[ C = \frac{\mu_c \beta F(\omega)}{\kappa \left( \sigma^2 - \mu_{\text{eff}}^2 \right)} \quad \text{and} \quad D = \frac{\mu_c \gamma F(\omega)}{\kappa \left( \sigma^2 - \mu_i^2 \right)} \quad (6.3) \]

Applying the first boundary condition, we get:

\[ B = A - \frac{\mu_{\text{eff}} C}{\sigma} - \frac{\mu_i D}{\sigma} \quad (6.4) \]

Applying the second boundary condition we get:

\[ A = \frac{C \mu_{\text{eff}} \left[ \exp(-\mu_{\text{eff}} l) - \exp(-\sigma l) \right] + D \mu_i \left[ \exp(-\mu_i l) - \exp(-\sigma l) \right] - \frac{I(l)}{\kappa} F(\omega)}{\sigma \left[ \exp(\sigma l) - \exp(-\sigma l) \right]} \quad (6.5) \]

Therefore, the spectrum of the radiometric signal can be calculated as:

\[ S_c(l; \omega) \propto \bar{\mu}_{\text{IR}} \int_0^l \theta(z; \omega) \exp(-\bar{\mu}_{\text{IR}} z) dz = \]

\[ \bar{\mu}_{\text{IR}} \left[ \frac{A}{\sigma - \bar{\mu}_{\text{IR}}} \left( \exp[(\sigma - \bar{\mu}_{\text{IR}}) l] - 1 \right) - \frac{B}{\sigma + \bar{\mu}_{\text{IR}}} \left( \exp[-(\sigma + \bar{\mu}_{\text{IR}}) l] - 1 \right) \right. \]

\[ - \left. \frac{C}{\mu_{\text{eff}} + \bar{\mu}_{\text{IR}}} \left( \exp[-(\mu_{\text{eff}} + \bar{\mu}_{\text{IR}}) l] - 1 \right) - \frac{D}{\mu_i + \bar{\mu}_{\text{IR}}} \left( \exp[-(\mu_i + \bar{\mu}_{\text{IR}}) l] - 1 \right) \right] \quad (6.6) \]

However, the infrared emission captured by the detector is the superposition of (6.6) and the direct Planck emission from the black-body absorber, attenuated through the turbid medium:

\[ S(l; \omega) \propto S_c(l; \omega) + \theta(l; \omega) \exp(-\bar{\mu}_{\text{IR}} l) \quad (6.7) \]

Equations (2.15) and (6.7) formulate the spectra of the applied binary phase coded excitation (i.e. the matched filter) and the thermophotonic response of a subsurface absorber at depth \( l \) to such excitation, respectively. Consequently, one can calculate the matched-filter cross-correlation signal and its phase analytically using (2.17) and (2.18), or experimentally via the algorithm depicted in Figure 2.8.
6.5.2 Theoretical simulations

Figure 6.8(a) plots the theoretical CC phase of two extreme $\mu_{IR}$ cases for a 7-bit binary code as a function of the subsurface absorber depth, $l$, at several carrier frequencies. For both $\mu_{IR}$ values, at a given carrier frequency, the deeper the absorber the larger the amount of phase shift, showing a more delayed contribution from the deeper absorber. Moreover, an increase in the carrier frequency reduces the maximum probing depth due to the reduced thermal diffusion length [80], as expected.

Figure 6.8. Theoretical cross-correlation (a) Phase, (b) Peak delay time (left and bottom axes) and amplitude (right and top axes) curves as a function of the subsurface absorber depth using properties of dental enamel: $\mu_a=100 \text{ [m}^{-1}\text{]}$, $\mu_s=6000 \text{ [m}^{-1}\text{]}$, $g=0.96$, $r=0.65$, $k=0.9 \text{ [Wm}^{-1}\text{k}^{-1}\text{]}$, $\alpha=5\times10^{-7} \text{ [m}^2\text{s}^{-1}\text{]}$. The numbers accompanying the curves determine the carrier frequency according to the legend in part (a).

The key point in this figure is the effect of $\mu_{IR}$. The infrared radiation captured by the detector is composed of a delayed conductive thermal-wave portion and an instantaneous direct Planck emission, (6.7). While an increase in the direct emission improves the signal-to-noise ratio of the amplitude channel, it deteriorates the maximum probing depth of the phase channel. This occurs because the direct emissions from absorbers at different depths are instantaneous and therefore all in phase, dominating the depth-dependent phase information of the conductive portion and consequently limiting the maximum probing depth, Figure 6.8(a). Figure 6.8(b) shows how the
height (peak amplitude channel) and location (peak delay time channel) of the CC signal peak behaves as a function of the absorber depth at various carrier frequencies. Generally, a shallower absorber results in a higher amplitude and shifts the CC peak location to shorter delay times to manifest a less attenuated thermal-wave source closer to the interrogated surface. In other words, matched-filtering localizes the energy of the long-duty BPC excitation under a narrow peak whose location on the delay time axis is linked to the depth of the source and allows one to construct iso-delay or alternatively iso-depth (depth selective) images. This is the diffusion equivalent of optical coherence tomography (OCT). Moreover, while the amplitude channel has by far the highest SNR, it is less reliable because, unlike phase and peak delay time, it is not an emissivity normalized quantity. Another interesting feature of BPC imaging is the effect of code length. While increasing the code length does not alter the CC phase and peak delay time curves of Figure 6.8, it increases the pulse compression ratio (i.e., it increases the amplitude of the CC peaks, SNR, and axial resolution).

6.5.3 Frequency vs. phase modulation (LFM vs. BPC)

Although linear frequency modulation (chirp) and binary phase coding are both pulse compression techniques used for improving the axial/depth resolution of the matched filter, their implementation for thermophotonic fields yields different characteristics. The key difference between these two modulation schemes is that LFM provides a wide-band signal while BPC yields a narrow-band signal. Figure 6.9 plots the simulated cross correlation signals of absorbers at several depths inside dental enamel using LFM modulation (0.1Hz – 4.9Hz, 6.4 s). Figure 6.10 shows similar cross correlation signals while using a 6.4s long BPC modulation at 2.5 Hz (center frequency of the LFM). Comparison of these figures clearly shows that the narrow band nature of BPC can yield more localized responses with minimal side lobes. As a result, it can be concluded that under similar conditions BPC yields more depth resolved images than LFM.
Figure 6.9. Simulated thermophotonic radar CC signals of absorbers at several depths using chirp modulation (0.1Hz – 4.9 Hz, 6.4 s). Turbid medium properties: \( \mu_a = 100 \text{ [m}^{-1} \text{]}, \mu_s = 6000 \text{ [m}^{-1} \text{]}, g = 0.96, r = 0.65, k = 0.9 \text{ [Wm}^{-1}\text{k}^{-1} \text{]}, \alpha = 5 \times 10^{-7} \text{ [m}^2\text{s}^{-1} \text{]} \).

Figure 6.10. Simulated thermophotonic radar CC signals of absorbers at several depths using BPC modulation (2.5Hz, 16-bit coding). Turbid medium properties: \( \mu_a = 100 \text{ [m}^{-1} \text{]}, \mu_s = 6000 \text{ [m}^{-1} \text{]}, g = 0.96, r = 0.65, k = 0.9 \text{ [Wm}^{-1}\text{k}^{-1} \text{]}, \alpha = 5 \times 10^{-7} \text{ [m}^2\text{s}^{-1} \text{]} \).
Figure 6.11. PTR, TWR, and BPC phase values of absorbers at several depths at low modulation frequencies. The simulated signals were modified with white noise (SNR = 1) and the phase values were calculated 10 times, yielding the error bars. Turbid medium properties: $\mu_a=100$ [m$^{-1}$], $\mu_s=6000$ [m$^{-1}$], $g=0.96$, $r=0.65$, $k=0.9$ [Wm$^{-1}$k$^{-1}$], $\alpha=5\times10^{-7}$ [m$^2$s$^{-1}$].

Figure 6.12. PTR, TWR, and BPC phase values of absorbers at several depths at high modulation frequencies. The simulated signals were modified with white noise (SNR = 1)
and the phase values were calculated 10 times, yielding the error bars. Turbid medium properties: $\mu_a=100$ [m$^{-1}$], $\mu_s=6000$ [m$^{-1}$], $g=0.96$, $r=0.65$, $k=0.9$ [Wm$^{-1}$k$^{-1}$], $\alpha=5\times10^{-7}$ [m$^2$s$^{-1}$].

Figure 6.11 lies at the heart of this thesis and simulates the performance of the thermophotonic system using single frequency modulation (FD-PTR/LIT), LFM, and BPC modulation. The spectra of the simulated signals were superposed with white noise (SNR=1) to obtain more realistic results and simulations were repeated 10 times at each absorber depth to produce the error bars. All other simulation conditions were kept the same. Figure 6.11 clearly shows the improvement of axial/depth resolution (the slope of the lines) when changing the modulation scheme from single frequency to LFM and finally to BPC modulation. Moreover, due to the higher SNR of matched filtering the sizes of the error bars are smaller in the LFM and BPC compared to that of the FD-PTR.

For the case of dental applications, since the early dental caries is typically very close to the surface (~300 µm), increasing the modulation frequency/chirp center frequency can help to improve the axial/depth resolution even more (Figure 6.12, note the increase in the slope of the lines). Therefore, it can be concluded that application of BPC modulation in turbid media yields optimal depth resolution. Nevertheless, due to the narrow band nature of the BPC signal, fitting the experimental data to the theory becomes very sensitive to noise and less accurate compared to the wide band LFM data.

6.5.4 Experimental verifications

Figure 6.13(a) schematically shows the cross-section of the black plastic step wedge sample discussed in section 2.9.4. Figure 6.13(b) shows the conventional TPLI phase image obtained at 3Hz along with its mean profile over the steps. It can be seen that although LI imaging detects all the steps, it loses depth resolution over deeper steps due to the diffuse nature of thermal waves. On the other hand, BPC imaging at the same modulation frequency and experimental conditions (averaging, laser power, etc.), maintains excellent resolution down to the deepest step in both peak delay time and phase images, Figure 6.13(c) and Figure 6.13(d) respectively. This experiment clearly shows how BPC matched-filtering can result in a more localized response in a diffusive field and improve the axial resolution while probing deeply into the sample.
Figure 6.13. (a) Cross section of the step wedge sample. (b) Conventional LIT phase, (c) BPC peak delay time, and (d) BPC phase images of the step wedge sample using 16-bit [1 1 1 1 1 -1 1 -1 -1 1 1 -1 -1] code. The curve in each image shows the mean horizontal profile of the corresponding contrast parameter.

Figure 6.14(a) shows the optical image of the interrogated surface of a goat bone along with its cross section. The optical image shows that the spongy trabecular bone is covered by the more dense cortical bone on the surface. Figure 6.14(b) shows the BPC phase image obtained at 10Hz. Due to the relatively short thermal diffusion length at this frequency the phase image reveals the structure of the cortical bone. However, reducing the excitation frequency to 1Hz the depth selective nature of thermophotonic radar imaging clearly reveals the underlying trabecular structure, Figure 6.14(c).

Figure 6.14. (a) Optical image of goat bone. The rectangle shows the images area. Thermophotonic radar phase images at (b) 10Hz and (c) 1Hz using a 7 bit binary code.
Figure 6.14 shows that another potential diagnostic application of the depth-selective thermophotonic radar imaging technology may be early bone osteoporosis imaging. Direct expenditures for treatment of osteoporotic fracture in the U.S. are estimated at $10 – $15 billion annually and no imaging methodology currently exists for the early diagnosis of osteoporosis.

6.5.5 Thermal coherence tomography

Figure 6.15 depicts another well-known limitation and challenge of diffusion fields: resolving stacked overlapping defects. Figure 6.15(a) shows an exploded view of a cross-shaped sample made of two strip absorbers. The deeper strip is a black plastic sample that completely absorbs the optical flux, while the shallower strip is a partially absorbing phantom. All other components are transparent polyvinyl chloride-plastisol. When viewed from the top, the cross sample covers all possible combinations: absorbers at two different depths (the end sections of the absorbing strips, points 1 and 2 in Figure 6.15(b)) and two absorbers on top of each other (point 3 in Figure 6.15(b)).

Figure 6.15. (a) Exploded view of the cross-shaped sample with two absorbers at different depths. Conventional LIT (b) Amplitude and (c) phase image. BPC peak delay time image matched to the camera temporal data of (d) point 1 and (e) point 2, as indicated in part (b), using a 3-bit [1 1 -1] code.

Figure 6.15(b) and Figure 6.15(c) show the conventional lock-in amplitude and phase images of this sample, respectively. It can be observed that the amplitude image cannot distinguish between the deeper and shallower absorbers (vertical and horizontal strips, respectively). However, the
amplitude information can be used to extract and process only those pixels having a high value (i.e., a subsurface absorber). Although the lock-in phase image yields significantly different values over the single absorbers (points 1 and 2) it cannot detect the layered structure of point 3. The axial resolution of both amplitude and phase images is limited by the depth-integrated nature of thermal waves.

Figure 6.15(d) and Figure 6.15(e) are the BPC peak delay time images using the temporal data of points 1 and 2 as the matched filter, respectively. When the temporal camera data of point 1 are used as the matched filter, all the diffusion waves originating at the same depth as that of point 1 will have their CC peaks located at $\tau = 0$ (correlographic image). A similar situation arises in Figure 6.15(e) for diffusion waves from the shallower absorber using the temporal camera data of point 2 as the matched filter. The fact that the layered structure of point 3 shows a maximum correlation to both data of points 1 and 2 ($\tau_p = 0$) shows that matched-filter BPC imaging can resolve the layered structure axially. This depth resolution phenomenon is normally a property of propagating hyperbolic wave-fields and not of parabolic (diffusive) fields. The implications of correlographic imaging can open a new field of generating layer-by-layer subsurface thermal coherence tomography (TCT) using naturally incoherent diffusion waves.

![Figure 6.15](image)

Figure 6.16. (a) Teeth matrix with hidden inter-proximal early caries. The rectangle shows the imaged area. (b) Conventional LIT and (c) BPC phase images. Correlographic BPC images obtained at (d) 2.7ms, (e) 29.7 ms, and (f) 45.9 ms. 7-bit code [1 1 1 1 1 1 -1] was used for the BPC images. White color depicts the pixels coherent to the matched filter. The arrow indicates the hidden interproximal caries.
An important shortcoming of dental thermophotonic lock-in imaging is its inability to detect interproximal (between teeth) caries when inspected from the accessible buccal (front) surface. A comparison of LIT and BPC phase images of Figure 6.16(b) and Figure 6.16(c) under identical experimental conditions, shows how the enhanced axial resolution of the BPC imaging can resolve deep interproximal caries.

Figure 6.16(d), Figure 6.16(e), and Figure 6.16(f) show thermal coherence tomographic images obtained through mathematically delaying the matched filter by 2.7, 29.7, and 45.9 ms and registering signals coherent to such filters ($\tau_p = 0$), respectively. No heat source is observed in the first (i.e. shallowest) slice, while the deepest slice shows healthy enamel areas. Defects and caries shift the thermal-wave centroid closer to the interrogated surface (compared to the surrounding healthy enamel) and as a result are revealed in the intermediately deep slice, Figure 6.16(e).
Chapter 7

Conclusions

This chapter provides the conclusions and the important results obtained in this research. The results are provided in the order of their appearance in the thesis.

7 Conclusions

7.1 Thermophotonic lock-in imaging

Extending the idea of FD-PTR into an imaging system, a thermophotonic lock-in imaging setup was designed and developed using a state-of-the-art mid-infrared camera. The system incorporates two high power NIR lasers at 808 and 1120 nm to maintain a large beam size (~20 mm) on the sample surface. The complicated software developed in the LabView platform allows for online averaging of the infrared signals with no need to save large amounts of data on the computer hard drive. To overcome the limited sampling rate of the camera, a synchronous undersampling method was added to the software to allow for inspection of high frequency thermal-wave fields. The developed system can be used for non-contact detection of flaws and inhomogeneities in a wide range of materials. To verify the diagnostic capabilities of the system, detection of early dental caries in human teeth was investigated. The increased porosity of early caries traps light photons, resulting in more light scattering and absorption inside the caries. As a result, the thermal waves generated in early caries have higher amplitude and are less delayed compared to those of intact areas. The experiments carried out on the controlled demineralization samples show that thermophotonic lock-in imaging can reliably detect early demineralization of dental enamel. Moreover, the methodology can successfully monitor the progression of the early caries. Other physical defects such as cracks and discontinuities are also detectable in the thermophotonic images. Other than simulated caries, teeth with natural caries were examined and the ability of the imaging system to detect occlusal caries was also verified. It was shown that thermophotonic lock-in imaging can reveal the proximal/interproximal caries through inspection of the accessible occlusal surface. The experimental results show that mid-infrared thermophotonic lock-in imaging at appropriate frequencies can detect very early demineralization lesions that are impossible to diagnose by dental radiographs or visual/tactile inspections. Moreover, unlike radiography, thermophotonic lock-in imaging uses safe non-
ionizing laser illumination. A comparison of the performance of thermophotonic lock-in imaging with that of polarized Raman spectroscopy proves the unique sensitivity of the developed system to very early caries detection. The new imaging modality is a promising candidate of superior contrast and sensitivity to carious lesions, very suitable for replacing, or at least supplementing, today’s ubiquitous dental x-ray technology.

7.2 Thermal-wave radar

Combining the ideas behind linear frequency modulated continuous wave radars and FD-PTR, a novel photothermal radiometry method is introduced to overcome the limitations of conventional FD-PTR/LIT systems. Analytical solutions to the thermal-wave radar heat diffusion problem for both opaque and transparent solids are provided. Simulations and experimental results suggest a significant improvement in the dynamic range when using the TWR method instead of conventional FD-PTR in opaque materials. The basis of the enhanced performance lies in the depth selective nature of the matched filtering which localizes the energy of the applied long duty optical excitation under a single lobe, improving SNR and the dynamic range. Theoretical simulations show that LFM parameters can be optimized to obtain optimal dynamic range and depth resolution in a given sample. Furthermore, it was found that using the TWR method, the dynamic range is improved by 112% compared to the FD-PTR amplitude channel and 18.43% compared to the FD-PTR phase channel. The SNR of TWR is 178% higher than that of the FD-PTR amplitude channel and 169% higher than that of the FD-PTR phase channel. The experimental results for a steel sample with several blind holes verified the theoretical predictions of better dynamic range.

7.3 Thermophotonic radar imaging & thermal coherence tomography

The TWR theory was implemented in the thermophotonic imaging system. The experiments carried out on several classical and dental samples showed that LFM together with matched filter signal processing can improve the axial/depth resolution. Moreover, the result of the multi-parameter fitting carried out on the steel sample with blind holes suggests that the raw data of a single TWR experiment can be fitted to the derived theory to estimate thermophysical properties of materials. Such quantification of material properties in an FD-PTR system requires performance of several experiments at a range of frequencies. Furthermore, the cross correlation
phase was introduced as an emissivity normalized quantity of superior sensitivity in the TWR theory. As an extension of the TWR theory, binary phase coding was introduced in the thermophotonic field as an alternative narrow-band pulse compression technique for further improvement of depth/axial resolution. The thermophotonic response of absorbers in turbid media to BPC excitation was analytically derived. Theoretical simulations of turbid media responses to BPC, LFM, and FD-PTR showed that under similar conditions BPC has optimal depth resolution. It was found that matched-filtering in diffusion-wave field acts as constructive interferometry, localizing the energy of the long-duty excitation under a narrow peak and allowing one to construct iso-depth images. The developed technique is the diffusion equivalent of optical coherence tomography and is named thermal coherence tomography. It was shown that by progressively delaying the BPC matched filter and registering signals coherent to it one can construct depth-selective thermal coherence tomographic images. Experiments showed that the highly depth resolved nature of thermal coherence tomography can be used to resolve overlaying absorbers.
References


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<table>
<thead>
<tr>
<th>Acronym</th>
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<tbody>
<tr>
<td>AISI</td>
<td>American Iron and Steel Institute</td>
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<tr>
<td>AOM</td>
<td>Acousto-optic modulator</td>
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<td>BPC</td>
<td>Binary phase coding</td>
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<tr>
<td>CC</td>
<td>Cross-correlation</td>
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<td>CCD</td>
<td>Charge-coupled device</td>
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<tr>
<td>CEJ</td>
<td>Cementoenamel junction</td>
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<tr>
<td>CW</td>
<td>Continuous wave</td>
</tr>
<tr>
<td>DD</td>
<td>DIAGNODent</td>
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<tr>
<td>EDJ</td>
<td>Enamel-dentine junction</td>
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<tr>
<td>ESD</td>
<td>Energy spectral density</td>
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<tr>
<td>FD-PTR</td>
<td>Fourier domain photothermal radiometry</td>
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<tr>
<td>FFT</td>
<td>Fast Fourier transform</td>
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<tr>
<td>FOTI</td>
<td>Fiber optic transillumination</td>
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<tr>
<td>FPA</td>
<td>Focal plane array</td>
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<tr>
<td>HT</td>
<td>Hilbert transform</td>
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<tr>
<td>HWP</td>
<td>Half wave plate</td>
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<tr>
<td>IAP</td>
<td>Ion activity product</td>
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<tr>
<td>IFFT</td>
<td>Inverse fast Fourier transform</td>
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<tr>
<td>IT</td>
<td>Integration time</td>
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<td>LD</td>
<td>Lesion depth</td>
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<td>LFM</td>
<td>Linear frequency modulation</td>
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<td>LIT</td>
<td>Lock-in thermography</td>
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<tr>
<td>MCT</td>
<td>Mercury cadmium telluride</td>
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<tr>
<td>NDE&amp;T</td>
<td>Non-destructive evaluation and testing</td>
</tr>
<tr>
<td>NIR</td>
<td>Near infrared</td>
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<tr>
<td>P</td>
<td>Pulse</td>
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<tr>
<td>PRS</td>
<td>Polarized Raman spectroscopy</td>
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<tr>
<td>PT</td>
<td>Pulse thermography</td>
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<tr>
<td>PVCP</td>
<td>Polyvinyl chloride plastisol</td>
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<tr>
<td>P-PTR</td>
<td>Pulse photothermal radiometry</td>
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<tr>
<td>QLF</td>
<td>Quantitative light-induced fluorescence</td>
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<tr>
<td>RTD</td>
<td>Resistance temperature detector</td>
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<tr>
<td>SNR</td>
<td>Signal-to-noise ratio</td>
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<td>Thermal coherence tomography</td>
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