Optics & Laser Technology 44 (2012) 1501–1506

Contents lists available at SciVerse ScienceDirect

Optics & Laser Technology

journal homepage: www.elsevier.com/locate/optlastec

Photoacoustic diagnosis of human teeth using interferometric detection scheme

Yasser H. El-Sharkawy [a], Ashraf F. El Sherif [b,*]

[a] Biomedical Engineering Dept., MTC, Cairo, Egypt
[b] Laser Photonics Research Group, Engineering Physics Dept., MTC, Cairo, Egypt

ARTICLE INFO

Article history:
Received 18 September 2011
Received in revised form
4 December 2011
Accepted 5 December 2011
Available online 4 January 2012

Keywords:
Photoacoustic techniques
Human teeth diagnosis
Mechanical and acoustic properties of hard tissue

ABSTRACT

We have investigated the mechanical and acoustic properties of human teeth using the laser generation of surface acoustic wave (SAW) technique. The materials investigated included normal and decayed teeth, which have similar grain sizes and different thicknesses. The tissue responds to the laser-induced stress by thermoelastic expansion. The informative features of this method allow one to determine sample thermal, optical, and acoustical properties that depend on the peculiarities of the sample compositional structure. An interferometric detection experimental scheme is applied for detection generated SAW pulses. The surface displacement curves shape of normal and decayed human teeth are shown. The dispersion curves for SAW pulses were determined by Fourier analysis. The result is an almost linear dependence of SAW velocity on frequency for a normal tooth, the magnitude of the thermoelastic expansion of the normal tooth reaches its peak at 0.344 μs, a SAW phase velocity of 2500 ms^{-1} between 0.0008 and 5 MHz was determined. For abnormal teeth, the magnitude of thermoelastic expansion of the normal tooth reaches its peak at 1.3 μs, the measured velocity was 3225 ms^{-1}. Due to the inhomogeneity of abnormal teeth perpendicular to the propagation direction, strong differences in their dispersion curves were obtained. The detection of acoustic waves is the basis of photoacoustic methods, which can be used for diagnostic purposes.

1. Introduction

Dental tissue interacts with light, e.g. from a laser, by a process called absorption [1]. The target may consist of hard tissue, including natural tooth structure, carious enamel and dentin, dental calculus, bone, or even an existing defective composite restoration within the tooth. Many different types of intraoral soft tissue targets are commonly observed upon routine examination, such as redundant gingival tissue, aberrant frenum, operculum, epulis, or benign lesions in the form of a fibroma or a papilloma.

Dental lasers function by producing waves of photons (quanta of light) that are specific to each laser wavelength [2,3]. The photonic absorption within the target tissue results in intracellular and/or intercellular changes to produce the desired result.

Diagnostic or curing lasers can be used for caries and calculus detection. Argon lasers (488 nm) can be used for curing composites, while optical coherence tomography (OCT) with lasers can be used for imaging soft and hard tissue without using any ionizing type of radiation. The 488 nm argon laser light is highly absorbed in camphorquinone, the photo-activator contained in light-activated composite materials [4]. Argon curing lasers, while clinically effective, have fallen out of use with the advent of LED curing light sources. An important distinction to be made is that LED curing light is different from laser light, as its light energy is non-coherent and inherently polychromatic. As an example, it may be directed at a peak wavelength of 655 nm toward a specific darkened groove on the occlusal surface of a patient's tooth where bacterial decay is suspected, or along the long axis of a root surface to detect the presence of bacteria-laden calculus. A diagnostic technology, in which the photons of this wavelength are absorbed in bacteria in or around areas of the patient's tooth, is called laser-induced fluorescence [5].

An alternative diagnostic is explored in this work. Frequency-domain infrared photothermal imaging is introduced as a dynamic dental diagnostic tool for quantifying sound or decayed enamel or dentin. The significance to dentistry lies in the potential of this technique to monitor dental lesions at the early stages of carious decay, where lateral and sub-surface spatial resolution on the order of the carious sizes and sub-surface depths investigated in this work (100–300 μm) may be required.

An important step in deciding whether or not lasers can be used as diagnostic tools for human teeth is to determine the behavior and physical properties of human teeth following laser irradiation. We wish to determine if certain properties of teeth

* Corresponding author. Tel.: +2 0122967331.
E-mail address: ashraf.alsharif@staff.aast.edu (A.F. El Sherif).

0030-3992/$ - see front matter © 2011 Elsevier Ltd. All rights reserved.
doi:10.1016/j.optlastec.2011.12.009

are capable of allowing distinction to be made between normal tissue and tissue with degenerative changes. Such methods might provide precise information on detection, and guidance on removal by cutting or ablation for carious matter.

When a tissue sample absorbs a laser pulse of a short duration, an increased temperature distribution is created. If the laser pulse duration is shorter than both the mechanical relaxation and thermal relaxation times, then the system is mechanically and thermally confined. The parameters of effective attenuation coefficient, longitudinal sound speed, thermal diffusivity and geometrical factors determine the effectiveness of the interaction. Generally, photomechanical or photothermal processes will dominate the laser tissue interaction. The resulting enhanced temperature distribution creates internal stresses, which propagate as acoustic waves. The material responds to these stresses through thermoelastic deformation, which can be modeled. In this study we have used an interferometric technique to measure the surface expansion of human teeth following the absorption of a 10 ns laser pulse. This method allows measurement of the longitudinal speed of sound, the effective optical penetration depth and elastic properties. Such a technique is found to be useful to differentiate between normal and carious teeth on a microscopic scale.

In many cases the elastic properties of the tissue sample can give information about their degree of normality and microstructure. In principle these can be determined by interaction of either bulk or surface acoustic waves with the sample. Methods employing bulk acoustic waves can often be used to determine sound velocities in a relatively straightforward manner, but they require sound pulses with aduration of a few picoseconds, which propagate perpendicular to the investigated surface and are reflected from layer boundaries. Such pulses can be generated and detected optically. Since surface acoustic waves (SAWs) propagate parallel to the investigated surface, they interact significantly within a thin tissue depth as thin as one-hundredth of their wavelength. Hence, surface acoustic waves are valuable for determining the acoustic properties of near-surface layers. A non-contact measurement of the acoustical properties of a near-surface layer is the combination of photoacoustic generation and interferometric detection of surface acoustic wave pulses [4–8]. We present observations on SAW pulses whose shapes can be analyzed in an appropriate manner by Fourier domain analysis. In this paper a description of the optical technique of tissue analysis is given and its applicability to the determination of the elastic constants of a variety of normal and abnormal tissues.

2. Theoretical studies

2.1. SAW generation on layered media

When electromagnetic radiation interacts with matter, sound waves can be generated. This process is known as the optoacoustic, or photoacoustic effect. Photoacoustic sound waves are generated through the thermoelastic effect when short laser pulses are absorbed by a medium. When the laser pulse interacts with the medium, a fraction of the energy is absorbed and converted into heat. The subsequent thermal expansion will produce a pressure wave that will propagate through the medium, as shown in Fig. 1).

When a tissue sample absorbs a laser pulse of short duration, a non-zero temperature distribution is created. If the laser pulse duration, τ_p, is shorter than both acoustic relaxation and thermal relaxation times, then the system is mechanically and thermally confined. The acoustic relaxation time is $1/\mu_{eff}C_l$, and the thermal relaxation time is $a/k\mu_{eff}^2$, where μ_{eff} is the effective attenuation coefficient, C_l is the longitudinal sound speed, k is the thermal diffusivity and a is a numerical constant dependent on geometry.

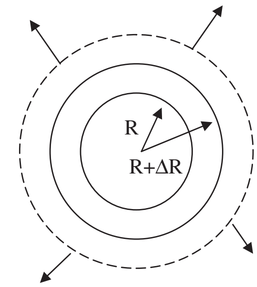

Fig. 1. A sphere of radius R expands to radius $R+\Delta R$ in time Δt, creating a thermoelastic wave.

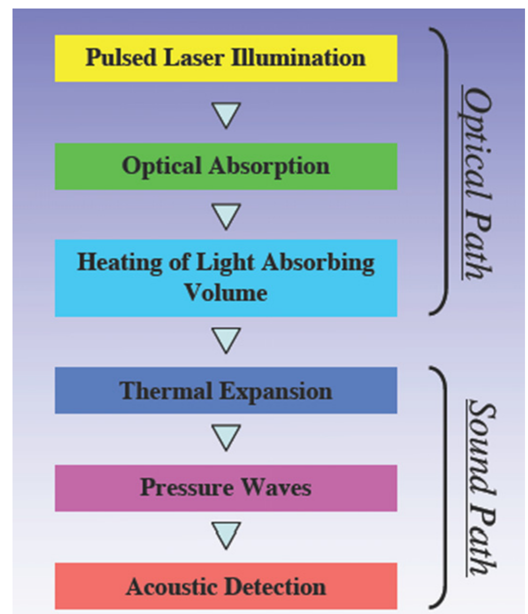

Fig. 2. Mechanism of photoacoustic signal generation and data retrieving.

Photomechanical processes will dominate the laser–tissue interaction. The resulting temperature distribution creates internal stresses, which propagate as acoustic waves. The material responds to these stresses through thermoelastic deformations, which can be described by the thermoelstic wave equation.

A numerical time dependent solution of the thermoelastic wave equation has been obtained for laser-induced heating of a polymer film in a three-dimensional cylindrical symmetric geometry [7]. This solution shows that the sample will undergo thermoelastic expansion [8] as the surface reaches a new equilibrium position. The time dependence of this expansion is governed by the ratio of the $1/e$ depth of laser light absorption, to the speed of the sound in the sample, $1/\mu_{eff}C_l$.

As shown in Fig. 2 the mechanism of photoacoustic signal generation starts with pulsed laser illumination. The light distribution in a tissue can be approximated by an exponentially decaying profile with tissue depth ($z>0$), and the laser radial profile $L(r)$. The distribution of light intensity is:

$$I(r,z) \cong I_0 L(r) e^{(-\mu_{eff} z)} \quad (1)$$

where I_0 is the surface intensity.

In a turbid medium such as dental tissue, light is subjected to both absorption and scattering. The three-dimensional temperature