Photoacoustic imaging in both soft and hard biological tissue

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Abstract. To date, most Photoacoustic (PA) imaging results have been from soft biotissues. In this study, a PA imaging system with a near-infrared pulsed laser source has been applied to obtain 2-D and 3-D images from both soft tissue and post-mortem dental samples. Imaging results showed that the PA technique has the potential to image human oral disease, such as early-stage teeth decay. For non-invasive photoacoustic imaging, the induced temperature and pressure rises within biotissues should not cause physical damage to the tissue. Several simulations based on the thermoelastic effect have been applied to predict initial temperature and pressure fields within a tooth sample. Predicted initial temperature and pressure rises are below corresponding safety limits.

1. Introduction

Recently photoacoustic (PA) imaging has been extended as a new modality for non-invasive medical diagnosis and visualization [1-4]. PA imaging overcomes the low-contrast limitation of pure ultrasound imaging for tissue sections with similar acoustic properties, and the shallow imaging depth of high-resolution optical imaging. It therefore provides a unique visualization method with a depth resolution in the range of tens of micrometers, for exploring the internal properties of biotissues at a desired penetration depth within several centimeters [5].

Effective techniques for diagnosis and visualization of early-stage human oral diseases such as the gingivitis and tooth decay are under development. Microscopic X-ray imaging has been used for imaging opaque objects and it can provide structural information with resolution less than 1µm [6]. However, X-ray is an ionization radiation source. Furthermore, most new dental decay occurs in the pits and fissures of the occlusal (biting) surfaces of posterior teeth. These caries lesions cannot be detected by x-rays during the early stages of decay due to the overlapping topography of the crown of the tooth [7]. For imaging soft-tissues, conventional X-ray imaging relies almost entirely on differences in the absorption of X-rays between tissues to produce contrast; these differences are small between different soft tissue types resulting in poor visualization of soft tissues [8]. Other pure optical or ultrasonic imaging methods have trade-offs between the imaging depth and resolution. For example, OCT has high image contrast and resolution for imaging both soft and hard tissues [9]; however, it can not reveal the internal tooth structures below 2.5mm.

In this study, PA imaging technique by using near-infrared laser pulses with nanosecond durations has been applied to obtain 2-D and 3-D images from both soft tissue and hard tissue to show the potential of this technique for visualizing oral diseases.
2. Initial temperature and pressure field simulations in a tooth

The temperature rise within a sample depends on the nature of the material and the magnitude of the absorption process. To maximize thermal and pressure effects in the thermoelastic process, the heat and pressure increment generated are usually 'confined' within the radiated volume during the time of laser pulse emission. Both 'thermal' and 'pressure' confinements usually are satisfied for nanosecond laser pulse irradiation. Under both confinements, a reasonable 1-D assumption that the temperature distribution $T$ as a function of depth $z$ is given by integrating the intensity of absorbed radiation with respect to time (thus time independence) is shown as follows:

$$T(z) = \frac{\mu_a \cdot \psi_0 \cdot e^{-\mu_a z}}{\rho \cdot C_p} = \frac{\mu_a \cdot \psi(z)}{\rho \cdot C_p},$$  

(1)

where $\psi_0$ is the average laser energy density at the sample surface ($J/m^3$). Its relationship with the laser energy density $\psi(z)$ at a depth $z$ follows the Beer-Lambert law. $\mu_a$ is the absorption coefficient of the material. $\rho$ and $C_p$ are respectively the mass density and the specific heat capacity at constant pressure of the material.

![Figure 1](image1.png)

Figure 1. (a) Result of the 2-D finite element thermal conduction model simulated using Comsol Femlab. (b) 1-D Diffusion approximation modelling result. (c) 2-D Monte Carlo modelling result.

The modelling result using Eq. (1) has been compared with the result of the time-dependent thermal conduction model at the time that the laser pulse was just switched off. The similarity between them validated Eq. (1). However, the Beer-Lambert law applied in Eq. (1) didn’t consider the 'back-scattering' effect of photons. For biotissues, other than the light reflection and scattering from the sample surface, some of the incident light scattered from a collimated beam undergoes multiple reflections and propagates in the backward direction inside the sample. This scattered light can cause the light fluence rate $I$ ($W/m^2$) just below the surface of tissue to be larger than the incident irradiance $I_0$ [10]. Thus, Diffusion approximation [11] and Monte Carlo [12] methods have been used to simulate the light distribution $\psi(r)$ within a three-layered tooth. Fig. 1a shows the 2-D finite element modelling result of a tooth using thermal conduction method. Figs. 1b and 1c are respectively the modelling results using Diffusion approximation and Monte Carlo methods. Laser pulses were assumed with a wavelength of 1064nm for all simulations. Optical and thermal parameters of enamel, dentine and pulp were listed in the previous publication [11]. Laser pulses were approximated in both time and spatial domains with Gaussian shapes. In the thermal conduction model, a time duration of 8 ns (FWHM) was used for the laser pulse, and the maximum power density of the incident laser pulse was assumed to be $3.7 \times 10^{11} W/m^2$ when $t = 10$ns. The beam radius was taken to be 1.0 mm (1/e fall). The corresponding average laser energy density was 300mJ/cm$^2$ (1/e fall). Compare the modelling results of Figs. 1b and 1c with Fig. 1a, maximum temperature rises are higher in Figs. 1b and 1c than the rise in Fig. 1a. However, they are still below 1ºC, which will not cause pulpal necrosis ($\geq$5ºC).

Furthermore, the initial pressure rise distribution in a tissue can also be calculated. Eq. (2) shows the 1-D initial pressure rise under both the 'thermal' and 'pressure' confinements:

$$p = \frac{1}{\rho} \frac{du}{dz},$$

(2)
\[
P(z) = \Gamma \cdot \mu_c \cdot \psi(z),
\]
where \( \Gamma \) is called Gruneisen coefficient, reflecting the conversion ability of the material from absorbed thermal energy to induced mechanical energy based on thermoelastic effect. \( \Gamma \) also connects the thermal and pressure field predictions. Fig.2 shows two modelling results for initial pressure rises within a tooth using Diffusion approximation and Monte Carlo simulations. \( \Gamma \) was taken to be 1.4, 0.3 and 0.15 respectively for enamel, dentine and pulp layers. The maximum pressure rises in the dentine region are below 0.7MPa, which is well below the fatigue limit of the dentine (20MPa).

![Figure 2](image)

**Figure 2.** (a) Result of the 1-D Diffusion approximation simulation. (b) 2-D Monte Carlo modelling result.

3. **Experimental and imaging results**

![Figure 3](image)

**Figure 3.** (a) B-scan image of the chicken breast with embedded nylon hairs. (b) Photography of a healthy tooth. (c) C-scan image after recording max. values from 0 to 6µs. (d) Photography of a diseased tooth. (e) C-scan image recording max. values from 0.4 to 2.4µs. (f). C-scan image recording max. values from 2.4 to 4.4µs.
In the experiments, laser energy was about 6mJ/pulse (energy density at 1/e fall~190mJ/cm\(^2\) if the fibre tip to the sample surface was 7mm). For irregular tooth surface (depth variation was about 2-4mm), distances of around 4mm (energy density~280mJ/cm\(^2\)) to the highest point for both healthy and diseased teeth was chosen. Laser repetition rate for the raster scan was 2Hz. Fig. 3a shows the B-scan image of a raw chicken breast with embedded nylon hairs (diameter was about 50µm). Another imaging result using PVA tissue phantoms showed that the nylon hairs can be imaged to a depth of about 1.0 cm in turbid material (details will be shown in the future publication). Fig. 3c shows the C-scan image from a healthy tooth (Fig. 3b) after recording maximum values in the time range from 0 to 6µs. Figs. 3e and 3f shows C-scan images from a diseased tooth (Fig. 3b) after recording maximum values respectively in the time range from 0.4 to 2.4µs (the lowest point of the tooth enamel crown) and from 2.4 to 4.4µs. Fig. 4a shows the B-scan image in the position of line A in Fig. 3b. Fig. 4b shows the B-scan image in the position of line B in Fig. 3d. We can see signals from stains and diseased parts have much higher amplitudes compared with healthy parts. After setting a threshold for all acquired data, a 3-D caries distribution is shown in Fig. 4c. Furthermore, X-ray microscope imaging results of two main internal cavities in the diseased tooth show similarities with PA imaging results. These results will be shown in the future publication.

Above imaging results show that the PA technique can image both soft and hard tissues with high sensitivities for the diseased tissues. It can be used to explore various aspects of oral disease.

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