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Attenuation of near-IR light through dentin at wavelengths from 1300–1650-nm

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Abstract

Light scattering in dental enamel decreases markedly from the UV to the near-IR and recent studies employing near-IR transillumination and reflectance imaging including optical coherence tomography indicate that this wavelength region is ideally suited for imaging dental caries due to the high transparency of enamel. The opacity of dentin is an important factor in optimizing the contrast of demineralization in reflectance measurements. It also influences the contrast of occlusal lesions in transillumination. Light scattering in dentin is an order of magnitude larger than in enamel, it is highly anisotropic and has a different spectral light scattering dependence than enamel. The objective of this study was to measure the optical attenuation of near-IR light through dentin at near-IR wavelengths from 1300–1650-nm. In this study the collimated transmission of near-IR light through polished thin sections of dentin of 0.05 to 0.6 mm thickness was measured. Beer-Lambert plots show that the attenuation coefficients range in magnitude from 20 to 40 cm^{-1}. Attenuation increased significantly with increasing wavelength and the increases were not entirely consistent with increased water absorption.

Keywords

light scattering; dentin; near-IR imaging

1. INTRODUCTION

Several *in vitro* and *in vivo* studies have shown that the near-IR region is highly promising for imaging dental caries due to the high transparency of sound enamel [1–8]. Upon demineralization, the scattering coefficient increases by almost two orders of magnitude which provides high contrast for imaging in both transillumination and reflectance [9]. Optical coherence tomography has also been most successful at 1310-nm for imaging teeth due to the high transparency of enamel [10–12]. Light scattering in enamel appears to be dominated by Rayleigh scattering and the magnitude of Rayleigh scattering is inversely proportional to the fourth power of the wavelength[13]. Therefore, the magitude of light scattering in enamel decreases markedly from the UV to the near-IR. Light scattering in dentin is primarily due to the dentinal tubules which behave as cylindrical Mie scatterers. The wavelength

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dependence is more complex and measurements from 500–1050-nm do not show the marked reduction is scattering that has been observed for enamel [13–16].

The amount of light reflected (backscattered) from the underlying dentin depends on the ratio of scattering to absorption. Assuming absorption is solely due to water in the near-IR, we postulate that longer near-IR wavelengths where scattering decreases and absorption increases will yield the highest contrast in reflectance. Hyperspectral reflectance measurements of Zakian [8] show that the dentin gets darker with increasing wavelength even when water absorption is not increasing. Previous angularly resolved optical measurements of sound dentin carried out at 543, 632 and 1050-nm indicated that the measured scattering and absorption coefficients of dentin are both almost an order of magnitude larger than for enamel[13]. In contrast with enamel, the scattering and absorption coefficients did not change significantly with wavelength, which is consistent with measurements using an integrating sphere, values did vary significantly with the method of measurement, μ_s = 250–280 cm^{-1} for angularly resolved goniometer measurements and 130–180 cm^{-1} for integrating sphere measurements. Values of the scattering anisotropy (g) also vary considerably, Fried et al. [13] report a g value of 0.93 that was determined using angularly resolved goniometer measurements while Zijp and ten Bosch report values of 0.40 calculated by integrating the measured scattering distributions of 20-μm-thick dentin samples [15, 16].

In order to accurately measure the internally optical properties it is extremely important to reduce the influence of surface scattering. In previous studies of dental enamel, samples were placed in an index matching bath of the same refractive index, namely carbon disulfide or a cargille liquid [1, 13]. However, dentin poses an additional challenge since it is a highly porous structure composed of tubule structures which are typically filled with fluid and it is the scattering of the tubules that dominate the optical properties. Therefore, if dentin was immersed in an index matching fluid the internal scattering would also be greatly reduced and would not represent the optical behavior expected clinically. We therefore chose to either place the dentin sections in water or maintain a layer of water on the surface to best represent the *in situ* optical environment. This approach was taken in a prior study involving highly porous caries lesions and demineralized enamel where index matching agents would likely imbibe into the lesion greatly reducing the internal scattering in the lesion[9]. We were unable to measure all the samples immersed in water due to the excessive attenuation at some of the near-IR wavelengths where there is strong water absorption, e.g. 1450-nm, therefore we measured these samples immediately after immersion in water, i.e., with a water layer. The study design emphasizes measurement of relative differences in optical attenuation at near-IR wavelengths from 1300–1650-nm.

2. MATERIALS AND METHODS

Plano-parallel sections ranging from 50-μm to 600-μm thick were cut from twenty-five extracted molars/premolars and serially polished to a 0.3 μm diamond grit finish. Samples were stored in a 1% thymol solution to prevent decay. Light from an Ocean Optics (Dunedin, Fl) fiber-coupled tungsten-halogen lamp (Model HL-2000-FHSA) with a filter wheel with bandpass filters (FBxxxx-12-FWHM 12-nm) Thorlabs (Newton, NJ) centered at 50-nm intervals from 1300–1650-nm was used to illuminate specific areas of the sample

with a spot size of approximately 100 μm. The spot size was confirmed through knife-edge beam profiling techniques. Phase sensitive detection was employed using a lock-in amplifier (Model SR 850) and an optical chopper from Stanford Research Systems (Stanford, CA). A large-area Ge photoreceiver Model 2033 from New Focus (Santa Clara, CA) equipped with a variable aperture was used for detection. The setup for the attenuation measurements is shown in Fig. 1.

Samples were measured either fully immersed in a cuvette (1-cm sq.) filled with water (n=1.3) or wet. In order to measure the samples with a wet surface the sample was wetted before each measurement by raising the cuvette filled with water to wet the surface. This procedure ensured that the position of the incident collimated light beam was fixed while cycling through the filters. Five measurements of collimated transmission were recorded at various sections within each sample. The collimated signal (I) at the detector was compared to the initial intensity of the beam (I_o). Using this ratio and the thickness of the samples (T), the attenuation coefficient (μ_t) was calculated using Beer's law and Beer-Lambert plots:

$$\mu_t = -\ln(I/I_o)/T \quad (1)$$

Beer's law states that the intensity of light passing through a sample decreases exponentially as thickness increases. This overall decrease is known as attenuation or extinction, which is caused by surface scattering or specular and diffuse reflection at front and rear surfaces, and internal scattering and absorption. The aperture was sufficiently small in front of the detector to ensure that light scattered at small angles did not contribute significantly to the collimated transmission.

3. RESULTS

Figures 2 and 3 contain Beer-Lambert plots at 1300-nm for samples with a water layer and fully immersed in water. The values for the attenuation coefficients were 23 ± 4.8 and 35 ± 10 respectively. Correlation coefficients (R^2) were 0.49 and 0.34 and both slopes were significantly different from zero. The slopes for all the Beer-Lambert plots for the samples with the applied water layer were significantly different from zero. This indicates that surface scattering does not prevent determination of the internal scattering and absorption even though it is approximately an order of magnitude higher. Surface scattering is relatively independent of sample thickness, therefore if attenuation is completely dominated by surface scattering, than there would be only a minimal change in attenuation with increasing sample thickness. This has been observed for previous measurements for enamel in the near-IR where the internal scattering was extremely low and attenuation was dominated by surface scattering if the samples were not placed in an index matching fluid [1]. Only the slope for the 1300-nm measurements was significantly different from zero for the dentin samples fully immersed in water. This was due to the extremely low signal intensity caused by the absorption of water in the cuvettes. Figure 4 contains a plot of the attenuation coefficients for each wavelength along with a plot of water absorption taken from the paper of Hale and Querry [17]. The attenuation values lie between 20 and 40 cm^{-1} and increase significantly with increasing wavelength. Comparison of the mean attenuation values with

ANOVA indicates that there is a significant linear increase in attenuation with increasing wavelength ($P<0.0001$).

4. DISCUSSION

The magnitude of the attenuation coefficients measured for dentin between 1300 and 1650-nm in the near-IR are about an order of magnitude higher than for enamel. The values are about an order of magnitude lower than the values measured at 1050-nm using angular resolved goniometer measurements[18]. Measurements on human cranial bone show a small decrease in the reduced scattering coefficient from 1400–2000-nm[19]. However, the opacity in occlusal transillumination measurements tends to follow the trend in water absorption with dentin appearing with the highest opacity near 1450-nm. These studies suggest that the attenuation coefficient for dentin increases significantly with increasing wavelength, however it is only in the range of 100–200%[20]. For transillumination a lower attenuation coefficient is desirable and the best transillumination images have been acquired at 1300–1400-nm[20]. Reflectance is more complicated, strong absorption in dentin is expected to increase contrast since the reflectance/backscatter decreases in areas with underlying dentin. However, a higher scattering coefficient in dentin may actually increase the amount of reflected/backscattered light from the underlying dentin and decrease contrast. Therefore the scattering albedo or ratio of scattered to absorbed light is likely to be most important. Reflectance studies of natural and artificial lesions found that 1600-nm also yielded high contrast, of a similar magnitude to 1450-nm. It may be advantageous to use 1600-nm as opposed to 1450-nm for reflectance measurements [20, 21]. The water absorption at 1450-nm is more than 3 times higher than at 1600-nm. Even though water absorption increases contrast, it may prove problematic due to the increasing influence of absorption by water on the tooth surface and measurements may be overly sensitive to the state of hydration. Samples are typically dried to increase contrast for fluorescence and visual examination of early lesions, this process may also prove to be more challenging at 1450-nm than for other wavelengths. Another problem is that all optical components have some water absorption so it is more difficult to provide the required illumination intensity near 1450-nm.

In conclusion, these measurements demonstrate that attenuation coefficient of dentin increased significantly between 1300 and 1650-nm. Lower values of optical attenuation are desirable for near-IR transillumination and optical coherence tomography.

Acknowledgments

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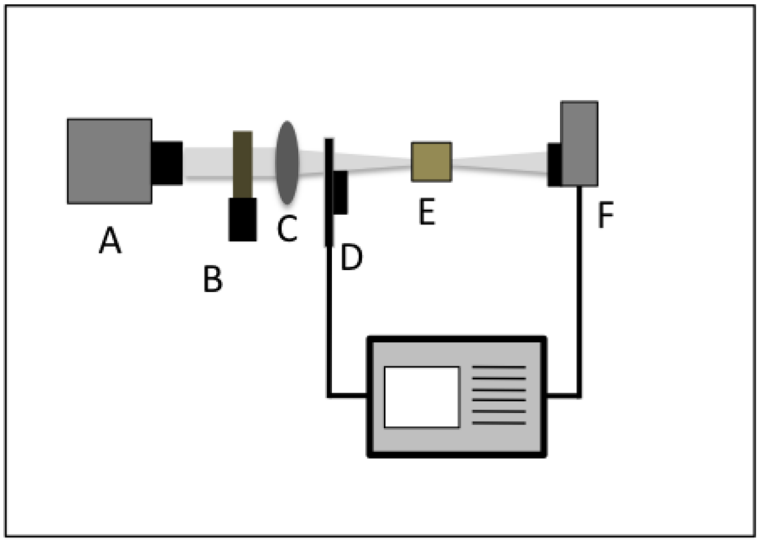

Fig. 1.

The experimental setup for measurement of the optical attenuation through thin sections of dentin placed in a cuvette. (A) Tungsten–halogen lamp with (B) filter wheel, (C) collimating and focusing lenses, optical chopper (D), sample cuvette (E) and (F) detector are shown.

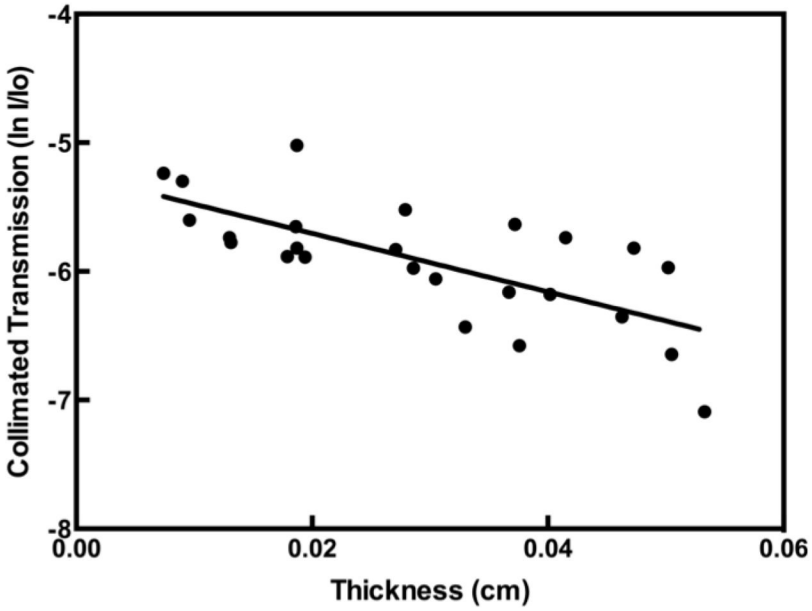

Fig. 2.

Beer-Lambert plot at 1300 nm for dentin with a thin layer of water applied to the surface. Collimated transmission for varying sample thickness ranging from 50 to 600-μm are plotted. Each point represents the mean attenuation of each of the five measurements on each sample.

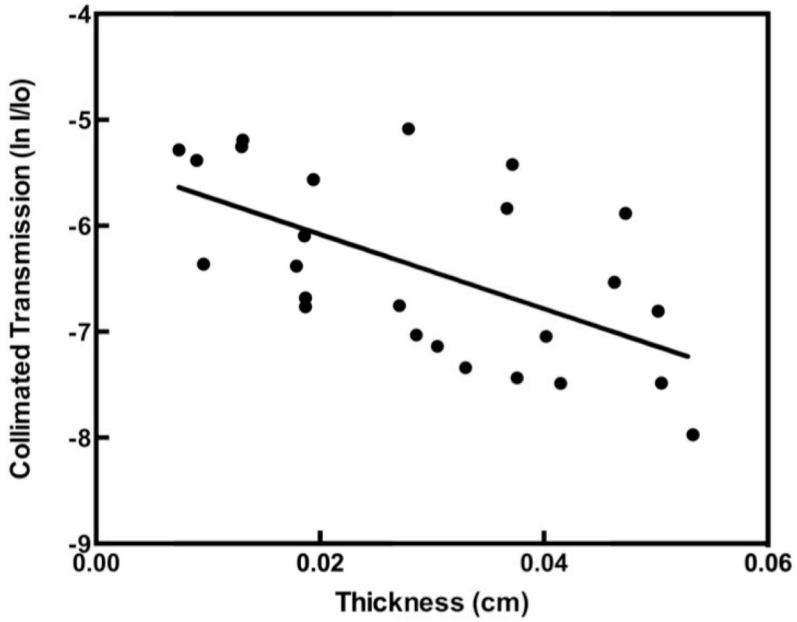

Fig. 3.

Beer-Lambert plot at 1300 nm for dentin immersed in water. Collimated transmission for varying sample thickness ranging from 50 to 600-μm are plotted. Each point represents the mean attenuation of each of the five measurements on each sample.

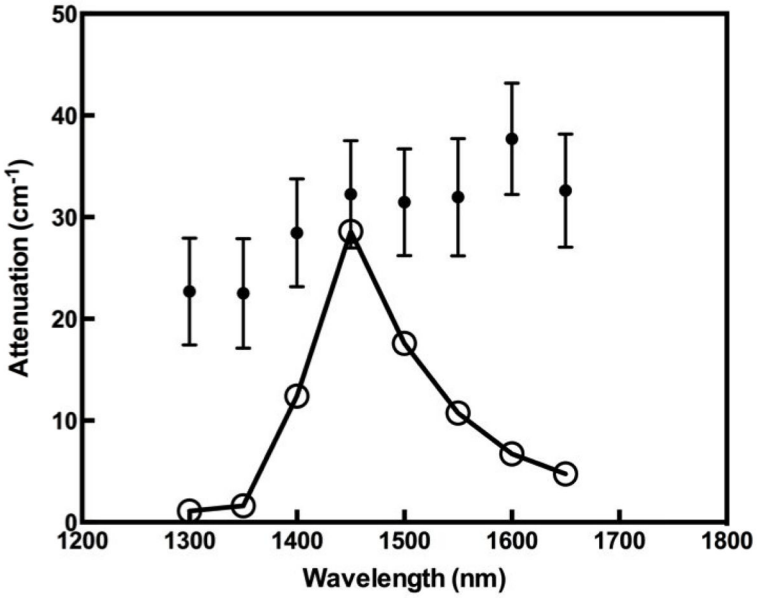

Fig. 4.

The mean (± s.d) attenuation coefficients calculated for each wavelength from Beer-Lambert plots are shown along with a plot (solid circles) of the attenuation coefficients for water (open circle) from ref [17].