SELECTION OF APPROPRIATE LASER PARAMETERS FOR LAUNCHING SURFACE ACOUSTIC WAVES ON TOOTH ENAMEL FOR NON-DESTRUCTIVE HARDNESS MEASURMENT

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Abstract

In this paper the selection of optimal laser parameters for generating surface acoustic waves on dental enamel is discussed. Spectrophotometric analysis of enamel sections shows two near-IR wavelengths that are strongly absorbed. Simulations of Duhamel's theorem were used to determine the laser damage threshold and the appropriate laser pulse repetition rate.

1. Introduction

Dental caries is the most predominant disease to afflict mankind. It involves bacteria-controlled biofilms developing on teeth, which ferment carbohydrates and induce acidic dissolution and loss of the mineral component of teeth. Such demineralization can be treated surgically with restorative materials to replace the affected tooth structure, but this is not inexpensive. An alternate strategy is to remineralize carious lesions. Various chemicals, such as fluoride and calcium phosphate casein [1], have been advocated for remineralization, however, assessment of the efficacy of such treatment is still subjective due to lack of instrumentation. All currently available techniques for quantifying the level of mineralization involve measurement of the hardness and elasticity of exercised tooth enamel (e.g. nano-indentation), these techniques are destructive and require extraction of the tooth. For this reason we want to develop a non-destructive and *in-vivo* technique to monitor and quantify the elastic properties of dental enamel.

Surface acoustic waves (SAWs), also known as Rayleigh waves, are elastic ultrasonic waves which are able to propagate along the surface of the material to be investigated with most of the wave energy confined to a depth of about one wavelength below the surface (the penetration depth). This makes the SAWs very sensitive to surface properties [2]. The penetration depths of SAWs depend on their frequencies, different depths of the sample can be probed with different wavelengths and a range of depths can be probed simultaneously if the SAW is broadband. In a layered system, the propagation velocity of the SAW also depends on its frequency. This dispersive behaviour is governed by the elastic (or Young's) modulus and density of both the film and substrate materials as well as the film thickness [3, 4]. When a SAW is generated and allowed to traverse over a distance on a sample surface, the SAW waveform is changed by the dispersion effect. Interferometric methods can then be use to detect and measure this SAW signal [5, 6] and from this the dispersion spectrum can be obtained via Fourier transformation. By fitting the measured dispersion curve with a theoretically calculated curve one can derive the elastic modulus of the film, provided that other properties of the film and substrate are known [7]. This technique is sometimes called ultrasonic spectroscopy.

Before the discovery of lasers in the 1960s, the only way to generate SAWs on solid surfaces was by means of contact transducers such as piezoelectric material. These transducers are bulky in size and require couplants, e.g. water or oil, to transmit the ultrasound to the sample. Such limitations can significantly reduce the quality of measurements, for example, only narrow bandwidth ultrasound of relatively low frequency can be generated (few hundreds of kilohertz) and the presence of the transducer and couplant may damp the SAW signal. The problems outlined above can be overcome by laser-generated ultrasound which does not require physical couplant. In addition, focusing a laser beam allows remote generation of broadband SAWs on small samples with higher spatial resolution. If optical fibre is used to guide or to generate the laser beam, areas that are difficult to access, e.g. inside a patient's mouth, can be examined.
Combined with a non-contact optical interferometer, laser-generated SAW spectroscopy provides a completely remote system that can operate under

harsh environments and it has found many application in the field of non-destructive testing (NDT) for various surfaces and thin films [7, 8, 9].

Launching SAWs efficiently and non-destructively is vital in order for this testing technique to be applicable in dentistry. To obtain SAWs of sufficient amplitude while avoiding laser damage on the enamel surface, one needs to understand the optical and thermal properties of such dental material and determine how it will interact with laser beams of different parameters. In this paper we focus on the discussion of laser generation of SAW and present the technique and the results to date in determining optimal laser parameters to generate SAWs on human enamel.

2. Laser generation of SAWs

When a laser beam is directed onto a solid, the electromagnetic radiation interacts with electrons in the material close to the surface. In general some amount of the optical energy is absorbed by various mechanisms, depending on the nature of the sample and the wavelength of the radiation, while some of the remainder is reflected and scattered. Some energy will also be transmitted if the sample is thin and semi-transparent. The absorbed optical energy is converted to heat, leading to rapid localized temperature increase, this results in thermal expansion and thermoelastic stress fields which in turn generate the elastic waves in solids.

The characteristic features of the generated SAWs are governed by the laser-sample interaction and the laser parameters. These include the duration of the laser pulse, the size of the laser beam spot on the surface, and the amount of optical power absorbed.

The most effective non-contact method to generate SAWs is likely to be a high power low energy pulsed laser. Pulsed lasers can deliver very high instantaneous optical power for the duration of each pulse to induce sharp temperature gradients on the sample surface and thus useful ultrasonic amplitude. When the repetition rate of the laser pulse is not very high, the total energy incident on the sample remains low and bulk heating is negligible.

As stated earlier, the gradient of the temperature increase is responsible for SAW generation. This becomes most important at the peak of the laser pulse because the optical intensity varies more rapidly there. So if high frequency SAWs are desired, a very short optical pulse is required. The useful SAW frequencies for most NDT applications are above 20*MHz*, this typically requires a laser pulse of sub-nanosecond width [10, 11]. The shorter the laser pulse, the more energy is launched in higher frequency SAWs, and the wave will be more closely confined to the surface.

When ultrashort nano- or pico-second laser pulses are used, the frequency spectrum of the SAW is primarily determined not by the pulse duration but by the spot size of the laser [2, 11]. Qualitatively speaking, the elementary waves launched from each surface element irradiated will interfere with each other either constructively or destructively. At large laser spot diameters, only a small area is effective in SAW generation. At small diameters in comparison to the SAW wavelengths obtained, the efficiency is also low because it is a point. From this it is obvious that there is a condition where mainly constructive interference should occur over the size of the laser spot, and hence the SAW generation efficiency should be maximal.

This condition has been found and reported to be $a = \upsilon_R \tau$ [12], where $2a$ is the laser spot diameter (full width at the 1/e point) of a TEM_{00} Gaussian power distribution, υ_R is the Rayleigh velocity, and 2τ is the laser pulse full width at 1/e point.

The Rayleigh velocity in dental enamel has been reported to be $\sim 3 \times 10^3 ms^{-1}$ [13]. From a simple calculation we can find that if a ~1*ns* duration pulse is used, the beam radius to use for maximal efficiency will be a ~1.5 μm. This relation gives us an idea of how to choose pulse width and beam size for this application.

In the paper, we utilize the calculated optimal laser pulse and beam size parameters to determine the amount of optical power that can be absorbed before damage starts to occur on irradiated enamel. We need to first understand how well the enamel absorbs laser light.

3. Appropriate wavelength selection

To generate SAWs efficiently on the surface of tooth enamel, the source laser needs to be strongly absorbed by the material so that more incident radiant energy can be converted into elastic-wave energy within the region very close to the surface. To determine the optimal wavelength for SAW generation, an accurate knowledge of the sample's optical properties, especially the absorption efficiency as a function of wavelength, is required.

Dental enamel is the outer layer of a tooth, with uneven thickness typically about 1.5*mm* and supported by the underlying dentin. It is the hardest and most mineralized substance of the human body, consisting of 8~12% water, 85~95% hydroxyapatite mineral, and 2~3% proteins and lipid. These components are distributed unevenly throughout the enamel. Light absorption in water and in hydroxyapatite changes tremendously depending on wavelength. UV light is well absorbed, and the enamel has strong absorption peaks in the ~7 μm and 9~11 μm regions due to the carbonate and the phosphate group in its mineral structure, respectively [14, 15, 16]. There is another absorption peak near 2.9μm which almost coincides with the water absorption peak near 3 μm . However, with present technology these wavelength regions are generally difficult to generate and even more difficult to guide by optical fibres. Thus we investigate whether there is significant, consistent absorption in the near-IR region, specifically 1 μm ~2 μm , for different types of teeth.

One molar, three incisors and one wisdom tooth were provided by the Sydney Dental Hospital. They were in healthy condition and have been sterilized by anti-bacterial solution and stored in de-ionized water to preserve the mineral composition. Sections on the teeth were performed by an experienced hospital technician with a diamond blade to make thin layers of enamel. The samples are slightly curved on the outer sides and their thicknesses are reasonably consistent. Different types of tooth means different enamel layer shape and the flattest samples were obtained from the incisors. Two of them are shown in Fig. 1.

Optical transmittances (T) of the samples were measured with a CARY 5E UV-Vis-NIR spectrophotometer. Because light will be reflected diffusely from the curved surfaces of the teeth, a Labsphere DRA-CA-5500 integrating sphere was attached to the spectrophotometer for the reflectance (R) measurement. In the integrating sphere, the specimen was positioned at a port in the sphere wall, and illuminated directionally by the light from the spectrophotometer. The reading obtained from this detection system was compared to a similar reading obtained for a calibrated standard of known reflectance. The results of this comparison were then used to derive a measurement of the reflectance of the enamel specimen. The measured units were in percentage transmitted and reflected. The spectral range of the measurements

Fig. 1: *Teeth come in different shapes and dimensions, the incisor section (left) has a flatter and smoother surface and the molar section (right) is more curved and has irregular shape.*

was between 300*nm* and 2500*nm*. The absorbance of the sample (A) was then calculated using $A = (100\% - T - R)$. A complete spectrum from one of the incisor is shown below.

From Fig. 2 we can see the UV light is strongly absorbed by the enamel, as mentioned before. In the visible spectrum the light is mostly reflected, this explains why healthy teeth are near white in colour. The absorbance increases abruptly and

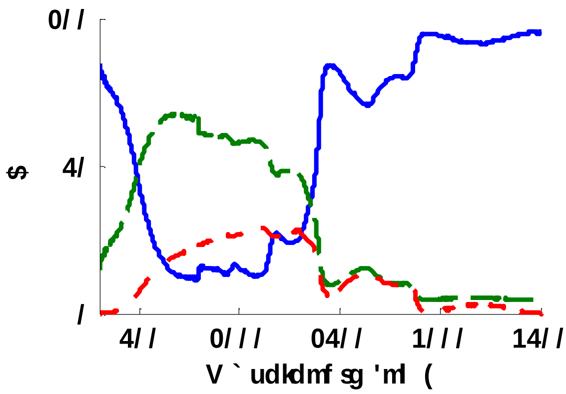

Fig. 2: *Spectrophotometric spectrum of one of the incisor enamel sections showing the Reflectance (dashed line), the Transmittance (dotted line) and the Absorbance (solid line).*

reaches a peak at about 1460*nm*. There is another peak near 1945*nm*. The transmittance at these two wavelengths is very low, which is a desired property. The two absorption peaks were confirmed to exist in all five of our enamel sections, as shown in Fig. 3. A dental Cerec2 Mark II ceramic, commercially used to replace lost dental hard tissue, was also measured with the intention to use it as a test material. The result shows that its optical absorption is completely different from the enamel in this spectral range and thus it would not be suitable for this application.

The variations in the absorbance levels could be explained by the difference in thickness and surface shapes, and their individual mineral composition, between all five enamel sections. Smaller variation

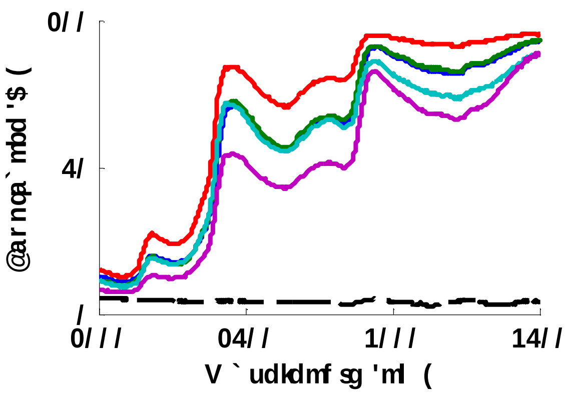

Fig. 3: *Absorbance spectra of all five enamel sections (solid lines) and the dental ceramic (dashed line), showing two absorption peaks in enamel near 1460nm and 1945nm.*

for the 1945*nm* peak suggests that the absorbance at this wavelength might be more consistent and reliable between different teeth.

The absorption coefficients of the two peaks can be calculated by the Beer-Lambert Law:

$$\gamma = -\frac{ln\left(I_Z / I_0\right)}{z} +$$ '0(

where γ is the absorption coefficient, z is the thickness of the sample, I_0 is the power incident into the sample (after reflection took place) and I_z is the transmitted power, and because reflectance needs to be accounted for in this spectrophotometer setup:

$$I_Z / I_0 = \frac{(100\% - R - A)}{(100\% - R)} \quad T -$$ '1(

The sample thicknesses were measured with a Mitutoyo absolute digimatic micrometer and found to average $1.35 \pm 0.2mm$. Equations (1) and (2) were then used to calculate the absorption coefficient spectra for the five samples and from which the average absorption coefficients at 1460*nm* and 1945*nm* were found to be $1133 \pm 100m^{-1}$ and $2473 \pm 200m^{-1}$ respectively.

For SAW generation purpose, it is desired that most of the energy should be absorbed as close to the surface as possible. A good measure of this is the skin depth, defined to be the distance penetrated by the radiation when its energy falls to 1/e of the initial value. The skin depth can be calculated by re-arranging the Beer-Lambert Law:

$$z_{skin} = -\frac{ln\left(1/e\right)}{\gamma} = \frac{1}{\gamma} -$$ '2(

This gives skin depths of $0.88 \pm 0.08mm$ and $0.40 \pm 0.03mm$ at 1460*nm* and 1945*nm*, respectively. For comparison, the skin depth at 1550*nm* was calculated to be 1.1±0.1*mm*. Clearly 1945*nm* is the most suitable wavelength for generating SAWs because optical energy will be absorbed strongly close to the enamel surface while at 1550*nm* more energy is dissipated into the bulk of enamel. Any wavelength that has a skin depth thicker than the enamel layer thickness should not be considered for our project. Rare-earth doped optical fibre lasers can readily be designed to operate at 1460*nm* (erbium doping) and 1945*nm* (thulium doping).

4. Determining the laser damage threshold

During laser NDT operation, the incident power should be low enough such that no damage or permanent deformation should occur on the sample surface during the process. This is known as the thermoelastic regime and in this regime the amplitudes of the generated ultrasonic waves increase linearly with the applied optical power. It is desirable that the amplitude be as high as possible for easier detection. The limiting factor is not the power that is available from the laser, but rather the threshold for damage in the irradiated material. When the laser pulse used is very short, it is the energy per pulse that largely controls the ultrasonic amplitude. However, the threshold of damage depends on the laser power density on the sample surface, measured in W/cm^2. For example a 1*mJ* per 1*ns* pulse in a circular beam spot of diameter 2*mm* has approximate peak power of 1*MW* and thus a power density in excess of $30MW/cm^2$, if we focus the beam diameter to 8 mm (typical core diameter for single mode optical fibre) the maximum power density can be as high as $2 \times 10^{12} W/cm^2$. So it is very important, when a beam size is chosen, to note the amount of optical power density that the sample surface is experiencing during the SAW generation as well as the pulse energy.

In order for laser generation of SAW to be applicable for future *in-vivo* dental examination, the total temperature rise in the tooth due to the absorbed laser radiation must be small enough such that no melting or cracks are incurred on the enamel surface; another consideration is the maximum temperature of the pulp cavity. Melting of the individual enamel prisms varies from 800 ^oC to 1200 ^oC [16, 17] and is inversely related to the carbonate content. Studies have shown that the pulp temperature rise should never exceed 5 ^oC to prevent permanent damage [14].

The laser-induced temperature rise, T, as a function of depth, z, and time, t, can be treated as a one dimensional problem, if we assume that the laser pulse is so short that the thermal conductivity into the bulk of the enamel can be neglected, then from Duhamel's theorem [10]:

$$T(z,t) = A(\lambda)\frac{\kappa^{1/2}}{K\pi^{1/2}}\int_0^t \frac{I(t-t')exp(-z^2/4\kappa t')}{t'^{1/2}}dt' , \quad (3)$$

where $A(\lambda)$ is the wavelength-dependent absorption ratio obtained from the spectrophotometer measurement, K and κ are the thermal conductivity and diffusivity, respectively. We assume the laser pulse has a Gaussian temporal profile of $I(t) = I_0 exp(-t^2/\tau^2)$, where I_0 is the maximum power density in W/m^2, τ is the full pulse width at 1/e height.

Dental enamel is a ceramic-like material and has poor thermal properties compared to metals. Its thermal conductivity and diffusivity have been reported to be $9.3\times10^{-3} W/cmK$ and $0.47 mm^2/s$, respectively [17]. During the timescale of the laser pulse, the absorption takes place uniformly down to a depth called the thermal length, and for enamel and other insulators this length is much shorter than the optical skin depth.

Fig. 4 shows the temperature distribution in the enamel at different depth as a function of time when irradiated with a 1*ns* laser pulse of $3.4\times10^7 W/cm^2$ at 1945*nm*.

This analysis shows that the temperature at the surface rises almost instantly when the energy is being absorbed (within 1*ns*) to a maximum about 800 ^oC, which is near the lower melting point of enamel. Once the laser pulse finishes, the temperature falls as the heat is conducted away from the irradiated region. The maximum temperature rise at increasing depth below the surface is less and occurs later and that at a depth of 1 mm the temperature never exceeds 100 ^oC. So we can safely say that the peak temperature rises for this laser pulse at the dentin and pulp cavity,

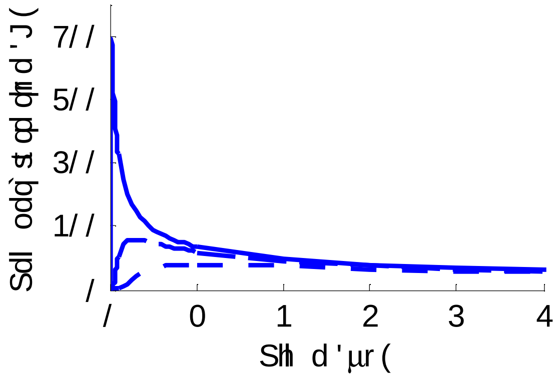

Fig. 4: *Rise in temperature as a function of time, for depth of 0 mm (solid line), 0.5 mm (dashed line) and 1 mm (dotted line), in response to a Gaussian laser pulse of 1ns width and power density of $3.4\times10^7 W/cm^2$.*

which are more than 1.5*mm* away from enamel surface, are negligible. We can then consider that the laser damage threshold for enamel at 1945*nm* is $3.4\times10^7 W/cm^2$. If we use a beam diameter of 10 mm, the corresponding pulse energy is ~10*nJ*.

Using the same method, the threshold values for 1460nm and 1550nm were also determined to be $4.3\times10^7 W/cm^2$ and $5.1\times10^7 W/cm^2$, respectively. This is a reasonable observation because the absorption strength at these two wavelengths is lower than that of 1945*nm*, thus the higher threshold.

The previous simulations involve only a single pulse, however, in practice a pulsed laser will emit a stream of pulses with constant time interval between them and we would also like to study this effect to determine the appropriate pulse repetition rate.

Fig. 5 shows the simulation for two pulses separated by 70*ns* and it can be seen that when the second pulse was irradiated onto the surface before the thermal effect from the first pulse was completely dissipated, the energy will accumulate and the peak temperature rise will increase.

The time for the single pulse induced temperature rise to dissipate completely was found to be over 10 μs and so we can conclude from the simulation that this time interval should be the ideal separation between pulses to prevent accumulating thermal damage, which corresponds to a repetition rate lower than 100*kHz*.

5. Conclusion and future work

In this study, spectrophotometer measurements revealed that tooth enamel has two consistent absorption peaks at 1460nm and 1945nm in the

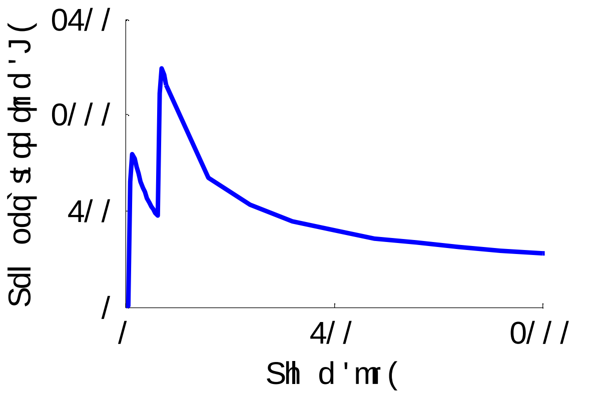

Fig. 5: *The temperature rise increases when the enamel surface is irradiated by two pulses separated by 70ns.*

near-IR region of the laser spectrum. The calculated absorption coefficients and skin depths suggest that these peaks have great potential for efficient non-destructive generation of SAWs using an optical fibre laser.

Simulations on laser induced temperature rise were performed using Duhamel's theorem to determine the damage threshold power density. The threshold was determined to be $3.4\times10^{7}\,W/cm^{2}$ at 1945*nm* and has lower values for 1460*nm* and 1550*nm*. Repetition rate of the laser pulses should be lower than 100*kHz* to prevent thermal accumulation.

With all the optimal laser parameters in hand, the next step in our investigation is to setup an erbium doped fibre laser, which operates near 1550*nm* and is much easier to obtain compared to a thulium fibre laser, to generate SAW on human enamel and perform various initial tests to understand the SAW characteristics. The detection of the wave signal will be done with a novel optical fibre interferometer.

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