Deep optical imaging of tissue using the second and third near-infrared spectral windows

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Abstract. Light at wavelengths in the near-infrared (NIR) region allows for deep penetration and minimal absorption through high scattering tissue media. NIR light has been conventionally used through the first NIR optical window with wavelengths from 650 to 950 nm. Longer NIR wavelengths had been overlooked due to major water absorption peaks and a lack of NIR-CCD detectors. The second NIR spectral window from 1100 to 1350 nm and a new spectral window from 1600 to 1870 nm, known as the third NIR optical window, were investigated. Optical attenuation measurements from thin tissue slices of normal and malignant breast and prostate tissues, pig brain, and chicken tissue were obtained in the spectral range from 400 to 2500 nm. Optical images of chicken tissue overlying three black wires were also obtained using the second and third spectral windows. Due to a reduction in scattering and minimal absorption, longer attenuation lengths and cleaner optical images could be seen in the second and third NIR optical windows compared to the conventional first NIR optical window. A possible fourth optical window centered at 2200 nm was noted.

Keywords: near-infrared light; scattering; turbid media; near-infrared imaging; imaging through turbid media; near-infrared therapeutic window; second and third optical windows; breast cancer; attenuation length; optical mammography.

1 Introduction

It is well known that light at wavelengths in the visible to near-infrared (NIR) region from 650 to 1350 nm is a noninvasive optical tool to detect and image tissue abnormalities. Optical mammography, for example, has been studied as an alternative NIR technique, which utilizes NIR light to image cancerous breast lesions. NIR light allows greater depth penetration, minimal absorption, and scattering into tissue than does shorter wavelengths in the visible region. By choosing the appropriate wavelength of light and charge-coupled device (CCD) detector, one can increase the penetration depth into tissue media and produce clearer optical images. In 1929, Max Cutler reported using white light and optical transillumination to image the breast. He had hoped to replace the use of x-rays with the use of longer wavelengths of light in the visible and NIR regions; however, due to lack of appropriate detectors and laser sources, he was unsuccessful. Since that time, better detectors, laser sources, and computer technologies have allowed others, such as Chance and Alfano, to use frequency modulation and time-resolved imaging, respectively, to image tissue abnormalities. Nowadays, the NIR region with wavelengths from 650 to 950 nm, called the first therapeutic window, is conventionally used for most NIR tissue imaging studies and photodynamic therapy applications. Due to Rayleigh scattering, (which varies as the inverse fourth power of the wavelength) and due to Mie scattering (which varies as 1/\lambda^6 with \lambda \geq 1) at longer wavelengths, there is less scattering and minimal absorption when using this window (longer wavelengths) than there is in the visible range (shorter wavelengths). For these reasons, it is expected that the longer NIR wavelengths of light, above 950 nm, would show less scattering and higher contrast images than the first optical window.

The use of longer NIR wavelengths for tissue studies, at the second (1100 to 1350 nm) and third (1600 to 1870 nm) NIR optical windows, is introduced and investigated in this paper. Optical attenuation from normal and malignant breast and prostate tissues, pig brain, and chicken tissue in the spectral range of 400 to 2500 nm was measured. With these NIR optical windows and an InGaAs camera detector, optical images of chicken tissue overlying black wires were also obtained. The most effective measure in the reduction of mortality and morbidity from cancer and other disease conditions is detection at an early stage of disease by x-ray mammography. The use of these longer NIR wavelengths with new two dimensional (2-D) photodetectors and high-speed computers may allow for improved methods of imaging breast and other tissues.

1.1 Imaging through Tissue Media

Light through turbid media can be described by the trajectories (diffusive, ballistic, and snake) of photons. With increasing propagation distance, these photons will be attenuated by the effects of scattering and absorption, resulting in a reduction in image quality. Absorption of light in tissue media can occur by selected biomolecules, such as collagen and elastin, lipids, hemoglobin or water in tissue media, while scattering can be caused by cells or intracellular matrix. Water molecules, in particular, greatly affect image quality and penetration depth.
due to strong absorption peaks from vibrational modes at \( \sim 900, \sim 1200, \sim 1400 \) and \( \sim 1900 \) nm. These effects can be minimized by imaging through thin tissue slices (less than 1 mm), thus allowing the ballistic photons (described by Lambert–Beer’s intensity law) to govern over the diffusive photons. These photons can be measured by the total attenuation coefficient \( (\mu_t) \), where \( \mu_t \) is the inverse of the total length traveled by the ballistic photons in the tissue media [also known as the total attenuation length \( (l_t) \)] and is determined by combining the absorption \( (\mu_a) \) and scattering \( (\mu_s) \) coefficients. \( \mu_t = \mu_a + \mu_s \). \(^{11-13}\)

### 1.2 First, Second, and Third Optical Windows

In the first region of minimal water absorption between water peak maxima (first NIR optical window from 650 to 950 nm), image quality is reduced due to strong absorption peaks from lipids and from hemoglobin and deoxyhemoglobin, and is blurred due to the molecular process of Rayleigh/Mie scattering. Figure 1 highlights the absorption properties of deoxyhemoglobin (Hb), hemoglobin (HbO\(_2\)), and water (H\(_2\)O) in the first optical window. \(^{14}\)

Recently, a new NIR wavelength transmission window from 1100 to 1350 nm, located between two additional water peaks, has been used for in vivo imaging. \(^{15}\) Limited studies on this second optical window have been done due to strong water absorption and lack of 2-D NIR photodetectors. Today, advances in the spectral response of NIR-CCD image sensors have made NIR camera specificity possible up to a wavelength of 2200 nm. As a result, longer wavelengths can be used. We report on the use of a new third NIR spectral region from 1600 to 1870 nm, between two strong water peaks (1444 and 1950 nm), to image deeply into tissue media. \(^{16,17}\) This region had been previously ignored due to water absorption. However, Yoo and Alfano have demonstrated that a small amount of absorption can help to minimize the detection of diffusive photons, which cause images to blur, and can highlight the ballistic and snake photons, which are responsible for producing clearer images. \(^{18}\)

### 2 Experimental

Optical attenuation measurements from tissue in the second and third NIR spectral windows (1100 to 1350 nm and 1600 to 1870 nm, respectively) were obtained but not explored due to lack of detector imaging sensitivity. Normal and malignant human breast and prostate tissues were supplied by the cooperative human tissue network under an institutional review board protocol. Breast cancer tissues were obtained from two patients. Patient (1) was a 48-year-old woman with lobular carcinoma, which was estrogen receptor positive, progesterone receptor positive, and Her-2-Neu negative. Patient (2) was a 51-year-old woman with ductal carcinoma, which was estrogen receptor positive, progesterone receptor positive, and Her-2-Neu negative. The tissue samples were not fixed or chemically treated. Samples were kept in a low-temperature freezer (minus 80°C) to preserve freshness. Fresh chicken muscle from the breast region for optical imaging studies was also acquired and placed in a conventional freezer \( \sim 18°C (0°F) \).

Once frozen, the chicken tissue was cut into thin slices of various thicknesses. Prior to the spectroscopic studies, all tissue samples were removed from the freezer and allowed to reach room temperature. The pig brain sample was not frozen prior to the study; measurements were performed within 24 h of resection.

#### 2.1 Optical Attenuation

The optical density spectra from tissue slices were obtained using the Varian Cary 500 Scan UV/VIS/NIR spectrophotometer in the spectral range of 400 to 2500 nm. Thin tissue slices were necessary for the ballistic light to dominate over the diffusive light. Optical attenuation measurements from normal and malignant human breast and prostate tissues, pig brain, and chicken tissue were obtained at each of the four optical windows. Breast and prostate tissue samples were cut into thicknesses of \( \sim 50, 100, \) and \( 200 \) \( \mu \)m and placed in thin quartz cuvettes. Pig brain tissue was cut into a thickness of \( \sim 100 \) \( \mu \)m. The chicken tissue samples were acquired from two regions: the muscle and the fatty outermost regions.

#### 2.2 Imaging Using the Second and Third Optical Windows

Transmission images (322 \times 240 pixels) of chicken breast tissue with black wires of various thicknesses were obtained using the second and third NIR optical windows and the optical setup in Fig. 2. Three wires with thicknesses of 0.75, 0.95, and 1.35 mm were inserted between a chicken tissue layer of 1.6 mm (located near the front of the light source) and thin chicken slices of various thicknesses (located near the front of the detector). The chicken tissue slices were measured at thicknesses of 1.6, 2.8, 3.9, and 7.4 mm.

The optical setup (seen in Fig. 2) includes a halogen lamp light source with spectral distribution from 200 to 2500 nm, selective filters at 1120 nm HW 40 and at 1500 nm longpass for the second and third optical windows, respectively, and an IR-CCD InGaAs camera (Goodrich Sensors Inc, high response camera SU320KTSW-1.7RT, Princeton, New Jersey) with spectral response between 0.9 and 1.7 \( \mu \)m (highlighting the second and third optical windows). The 1500 nm longpass filter was used to cutoff light of wavelengths below 1500 nm. This allowed for the transmission of light from the third window with wavelengths from 1600 to 1700 nm.

![Fig. 1 Absorption spectra of deoxyhemoglobin (Hb), hemoglobin (HbO\(_2\)), and water in the visible and NIR regions.](image-url)
3 Results and Discussion

3.1 Optical Attenuation Spectra of Human Prostate Tissue, Human Breast Tissue, and Pig Brain

Figure 3 shows the spectrum of total attenuation coefficient ($\mu_t$) in ($\text{mm}^{-1}$) from thin prostate tissue with a thickness of 200 $\mu$m in the spectral range of 400 to 2500 nm with four optical windows highlighted. The total attenuation coefficient ($\mu_t$) is defined by combining the absorption ($\mu_a$) and scattering ($\mu_s$) coefficients where ($\mu_a + \mu_s$) equals $\mu_t$. The ballistic light in the thin tissue media depends on $\mu_a$ plus $\mu_s$. The total attenuation coefficient ($\mu_t$) was calculated using Eq. (1)

$$\mu_t = \frac{(2.303 \times \text{OD})}{z}, \quad (1)$$

where OD corresponds to the optical density results from the tissue and $z$ is the thickness of the tissue sample. Equation (1) was derived from the light intensity $I$ over the incident light $I_0$ and is described by the Lambert–Beer’s Eq. (2)

$$\frac{I}{I_0} = \exp[-(\mu_t z)] = 10^{-\text{OD}}. \quad (2)$$

The normal prostate tissue sample used to acquire the spectrum in Fig. 3 was cut into additional thicknesses of 50 and 100 $\mu$m. Figure 4 shows the corresponding spectra of the total attenuation lengths ($l_t$) in micrometers, in the spectral range of 400 to 2500 nm, from normal prostate tissues at thicknesses of 50, 100, and 200 $\mu$m, where the total attenuation lengths ($l_t$) are the inverse of the total attenuation coefficient ($\mu_t$) (seen in Fig. 3).

The total attenuation lengths ($l_t$) are noticeably larger in the third optical window with a peak maximum at 1835 nm.

Figures 5 and 6 show the total attenuation lengths ($l_t$) in micrometers from normal and malignant prostate and breast tissues with thicknesses of 200 $\mu$m. A larger total attenuation length ($l_t$) occurs in the new third optical window compared to the first, second, and fourth optical windows. We also notice that $l_t$ from normal tissue is larger compared to the malignant tissue samples.

Table 1 summarizes the results obtained from the total attenuation lengths ($l_t$) of normal and cancerous breast and prostate tissues, pig brain, and chicken tissue at selected wavelengths representing the four optical windows. The total attenuation lengths ($l_t$) were measured at wavelengths of 750, 1200, 1700, and 2200 nm. Wavelengths of 1200 and 1700 nm were chosen to correspond to wavelengths in the second and third optical windows, and were used in the optical imaging setup. As the wavelength is increased, $\mu_a$ is reduced and $\mu_s$ dominates. The maximum penetration depths ($l_t$) occurred in the third optical window around 1835 nm for all tissue samples. Normal prostate tissue with a thickness of 100 $\mu$m had a penetration depth of $\sim 800 \mu$m at a wavelength of 1700 nm using the third optical window.
window. Results from chicken tissue samples show a slight decrease from the second optical window to the third optical window. We suspect that this is due to the large amount of protein in the chicken samples compared to all the other tissue samples, which have a higher lipid content. At wavelengths greater than 1900 nm (in the fourth optical window), a reduction in $l_t$ can be seen due to a combination of vibrational modes from lipids, collagen, and water molecules in the tissues. We also note that $l_t$ is greater in the fourth optical window than the first optical window.

3.2 Images of Chicken Tissue and Wires at Different Depths

Transmission images, seen in Fig. 7 and acquired using the optical setup in Fig. 2, illustrate the appearance of the black wires [0.75 mm (left), 0.95 mm (middle) and 1.35 mm (right)] through tissue slices (ranging from no overlying tissue to a layer of 3.9 mm in thickness with a bottom layer of 1.8 mm). Light from wavelengths in the second and third windows was able to penetrate the thick tissue layer and give clear images of the hidden wires. The images of the three wires, using the second and third optical windows, and an overlying 7.4 mm thickness (not shown), were apparent on the screen during the experimental procedure and before the imaging process. Images of the three wires and chicken tissue were recorded with a transmission maximum penetration depth of 3.9 mm.

Penetration depth analysis was done on the images of the three (1, 2, and 3) wires through chicken breast tissue at the second and third optical windows. The corresponding digitized spatial intensity distributions of the images were obtained by integrating the image intensity over the horizontal rectangular region [as marked by the black boxes in Figs. 7(a)–7(d)]. Plots of intensity (arbitrary units) versus distance (pixels) of the wires and chicken breast tissue in the second and third windows were also obtained and are shown in Figs. 7(a)–7(d). The image intensity can be described by the light intensity transmitted through the chicken breast tissue onto the three wires.

Table 2 summarizes the contrast results from images of chicken tissue of different thicknesses and three wires using the second and third optical windows. The degree of contrast can be calculated as the intensity of signal minus intensity of background divided by intensity of signal plus intensity of background times 100. The second and third optical windows have similar signal to background ratios. From Figs. 7(a)–7(d) and Table 2, images from the second and third optical windows show minimal noise and demonstrate the three black wires through tissue media. Images of the three wires with an overlying tissue thickness of 1.6 mm have the highest contrast.
percentage while the images of the wires with an overlaying layer of 3.9 mm are slightly blurred but still visible. Due to greater muscle content in the chicken tissue at thicknesses of 2.8 and 3.9 mm, the second optical window has a greater percent contrast than the third optical window.

4 Conclusion

Due to a reduction in scattering in tissue media at longer NIR wavelengths, longer attenuation and clearer images can be seen in the second and third NIR optical windows and may provide additional information to that observed using the conventional first NIR window. Deeper NIR images can be achieved due to a reduction in the scattering coefficient, allowing the absorption coefficient to be the main determinant of image quality. Optimizing tissue image contrast from the NIR second and third optical windows is needed. Better NIR light sources, such as intense tunable lasers Forestrite (1150 to 1300 nm), Cunyite (1200 to 1500 nm) and LSO (1110 to 1600 nm), the supercontinuum laser source (400 to 2400 nm), or semiconductor laser diodes, will eliminate photon starvation and improve sensitivity and signal to noise ratios. Using a more intense NIR light source in optical mammography could provide deeper penetration and better optical images of abnormalities, which are hidden behind normal tissue. We also note that this technique may be of benefit when imaging objects through fog or cloudy water.

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References


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