Contrast Enhancement by Edge Effect Corrections in Frequency-domain Optical Mammography

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Abstract

The planar tandem scan of light source and optical detector in a transmission geometry produces 2-D projection images of the human breast. This approach to near-infrared optical mammography suffers from the strong influence of edge effects: breast thickness variability, lateral photon losses, and variability in the optical coupling with breast tissue. Taking advantage of the information content of frequency-domain data, we have developed an algorithm for the correction of edge effects which is based on data at a single wavelength. We compare the images yielded by this single-wavelength algorithm with those provided by the ratio of the ac amplitudes at two wavelengths. The edge correction is excellent in both cases. However, the tumor detectability in the optical mammogram of a human breast affected by cancer is significantly better in the single-wavelength edge corrected image. We attribute this result to the similar effect caused by the tumor on the ac signal at the two wavelengths employed in this study (685 and 825 nm).

Keywords

Medical optics instrumentation, Medical and biological imaging, Optical diagnostics for medicine, Imaging systems.

1. Introduction

Optical mammography is one of the most actively investigated applications of near-infrared imaging of tissue. A proposed approach to optical mammography consists of a simultaneous planar scan of light source and optical detector to obtain a 2-D projection image of the breast. In this approach, the breast is slightly compressed between two glass plates. Encouraging results have been obtained using this method with both steady-state [1] and frequency-domain [2] systems. In the frequency-domain, one measures the ac amplitude and the phase of the photon density wave generated by an intensity modulated light source. The main difficulty with interpreting data collected in this sampling geometry (a planar scan in transmission mode) arises from breast thickness variations, lateral photon losses through the sides of the breast, and variability in the optical coupling with breast tissue. These effects become particularly important as the scanner approaches the edge of the scanned region, and hence we collectively call them “edge effects.” Our objective is to produce the highest image contrast and enhance the detectability of optical inhomogeneities in the breast by correcting for edge effects.

2. Definition of edge effects

Transillumination optical scanning produces bidimensional projection images of the breast. In pixels close to the edge of this 2-D image, the measurement's fidelity presents several problems:

1) Breast thickness variability. The thickness of breast tissue, which coincides with the separation between the glass plates in the central part of the image, decreases toward the edge of the image;

2) Lateral photon losses. As the scanner approaches the edge of the breast, additional photons are lost through the sides of the breast;
3) Variability in the optical coupling with breast tissue. The source-tissue optical coupling can vary as a result of different reflectivities at the breast surface. The tissue-detector optical coupling varies when the distance between the breast surface and the optical detector changes. The edge effects are pictorially described in Fig. 1.

![Figure 1. Pictorial description of the edge effects in transillumination optical scanning of the female breast.](image)

Edge effects dominate the raw data images, where the strongest visible effects are an increase in the intensity and a decrease in the phase toward the image edge. As a result, the presence of optical inhomogeneities, such as a tumor, appear in the raw data images as a slight deformation of the general pattern determined by edge effects. Under these conditions, image contrast is strongly reduced, and the sensitivity to tumor detection is weakened.

3. Edge Effects Corrections

A rigorous mathematical description of the boundary conditions in optical mammography is a challenging problem. The major difficulty comes from the variability of the geometry of the compressed breast for different patients. This variability does not allow for a univocal characterization of the edge effects and of the boundary conditions. For this reason, we have approached the problem of edge effects following two alternative procedures.

**Edge correction from data at two wavelengths**

This method assumes that lateral photon losses and variability in optical coupling (edge effects (2) and (3) in Section 2) are wavelength independent. This is a reasonable assumption if the two wavelengths are not too far apart. If the optical properties of breast tissue (absorption coefficient $\mu_a$, reduced scattering coefficient $\mu_s'$, refractive index $n$) are similar at the two wavelengths, the photon path distributions will also be similar. Consequently, lateral photon losses should affect the data at the two wavelengths in a similar fashion. Also the variations in the optical coupling with tissue and in the skin reflectivity can be reasonably assumed to be approximately the same at the two wavelengths. The idea is then to combine the data at two wavelengths to correct for edge effects. We can expect the ac amplitude to be influenced by wavelength independent edge effects much more than the phase. In fact, additional photon leakage and changes in the reflectivity at the breast surface will strongly influence the ac signal while should have little effect on the phase. For this reason, a first step in the two-wavelength edge correction is simply to take the ratio between the ac amplitudes at the two wavelengths:

$$\text{ac ratio} = \frac{\text{ac}(\lambda_1)}{\text{ac}(\lambda_2)}. \quad (1)$$

It is clear that an optical inhomogeneity will be detected in the ac ratio image only if it produces different effects on the ac at the two wavelengths.

**Edge correction from data at a single wavelength**

This method is based on the knowledge of the breast thickness in each pixel ($r(x,y)$) and of the dependence of the ac on $r$ in the homogeneous case. From diffusion theory, we can predict a linear dependence of the phase on $r$. Consequently, we can employ the phase information to obtain $r(x,y)$ [3]. The phase acts as a parameter weakly affected by all edge effects but thickness variability. On the contrary, the ac is strongly affected by all edge effects, so that we empirically determine its $r$ dependence in the homogeneous case, i.e. in the absence of inhomogeneities. On the basis of the ac dependence on $r$, we define a parameter $N$, which is independent of edge effects. If, for example, $\text{ac} \sim r^\alpha$, we define $N$ as follows:

$$N(x,y) = \frac{r_0^\alpha \text{ac}_0}{r^\alpha (x,y) \text{ac}(x,y)}, \quad (2)$$

where $r_0$ is the separation between the compression plates, and $\text{ac}_0$ is the ac amplitude at a pixel with breast thickness $r_0$. In the case of an exponential decay, $\text{ac} \sim \exp(-\beta r)$, we define $N$ as:

$$N(x,y) = \frac{\beta r(x,y)}{\ln\left[\frac{\text{ac}_0}{\text{ac}(x,y)}\right] + \beta r_0}. \quad (3)$$

The $N$ image will detect optical inhomogeneities whose optical properties differ from those of the background medium at the wavelength considered.
4. Results

Laboratory test
In our study, we have employed data collected with the light mammography apparatus (LIMA) developed at Carl Zeiss, Germany [2,3]. To perform an initial test of the edge corrections provided by the ac ratio and by $N$, we have employed a solid block of Delrin, a substance with optical properties similar to those of breast tissue in the near infrared. Specifically, the solid block has $\mu_a \approx 0.02$ cm$^{-1}$ and $\mu_s \approx 12$ cm$^{-1}$ at 800 nm. This Delrin block has a central region 6 cm thick, 11.7 cm long, and 11.4 cm wide. On two sides of the block the thickness decreases, in a smooth way on one side, and by steps on the other. The total length of the block is 23 cm. The 2-D projection images of the block obtained with the LIMA are shown in Fig. 2.

Figure 2(a) shows the ac image, which is dominated by edge effects. Figures 2(b) and 2(c) show the ac ratio image and the $N$ image, respectively. The edge correction provided by ac ratio and by $N$ is excellent and results in a much larger effectively imaged area. The ac ratio in Fig. 2(b) has been obtained from data at 685 and 825 nm. The parameter $N$ in Fig. 2(c) has been calculated from Eq. (3) with $\beta = 1.15$ cm$^{-1}$. The images of Figs. 2(a) and 2(c) refer to $\lambda = 685$ nm, but similar results are obtained at 825 nm.

Clinical test
In Fig. 3 we show the ac, ac ratio, and $N$ images of a breast affected by cancer. In Fig. 3(c), the parameter $N$ has been calculated from Eq. (2) with $\alpha = 1$. Again, the ac image is dominated by edge effects and shows poor contrast. Both the ac ratio and the $N$ images effectively correct for edge effects. However, only the $N$ image clearly shows the darker area corresponding to the tumor position. To detect an optical inhomogeneity with the ac ratio, the difference between the optical properties of the inhomogeneity and those of the surrounding medium must vary at the two wavelengths considered. In the case shown in Fig. 3, it appears that the difference between the optical properties of healthy and cancerous tissue at the two wavelengths employed (685 and 825 nm) are similar, so that the tumor is not seen in the ac ratio image.

5. Conclusion
Near infrared imaging of thick tissue intrinsically provides low spatial resolution because of the diffusive nature of light propagation in tissue. For this reason, we have focused our attention to improving image contrast and tumor detectability, which constitute the main promise of optical mammography. We have developed algorithms for contrast enhancement which are not based on computer manipulation, but rather on an analysis of light propagation in tissues. Tests on tissue-like phantoms and initial results on human breast data show the effectiveness of these algorithms in correcting for edge effects. However, the tumor detectability with the two-wavelength algorithm depends critically on the spectral properties of the tumor. For this reason, the choice of the wavelengths in the multi-wavelength edge correction algorithm is of paramount importance.
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7. References


Figure 3. Optical images of a breast affected by tumor: (a) ac image at 685 nm; (b) ac ratio image based on data at 685 and 825 nm; (c) N image at 685 nm from Eq. (2) with $\alpha = 1$. 