Near-Infrared Diffuse Wave Imaging with a Priori Ultrasound Anatomic Information Accounts for Chest Wall Effects in Breast Imaging

Yasaman Ardeshirpour and Quing Zhu
University of Connecticut, Storrs, CT, USA

Introduction

Optical tomography using diffused near infrared (NIR) light with ultrasound localization has shown potential value both in distinguishing benign from malignant tumors, and also as a monitoring tool for chemotherapy progress in cancer patients. In our system, patients were scanned in a supine position and multiple sets of ultrasound images and optical measurements with reflection geometry were simultaneously made at the lesion location and at a contra-lateral region of the normal breast (reference location). [1] In general, the breast tissue thickness has been reduced to less than 3 to 4 cm when patients are scanned in a supine position. Therefore, lesions close to the chest wall can be imaged with good sensitivity. However, when the chest wall thickness is less than 1.5 cm from the skin surface, the semi-infinite geometry is not a valid assumption for optical measurements, and the chest wall underneath the breast tissue affects the measurements obtained from distant source-detector pairs. With the assistance of co-registered ultrasound, the depth and the tilting angle of the chest wall can be determined and then used to model the breast as a two-layer medium.

In optical tomography, a typical perturbation approach requires two sets of measurements, obtained at the lesion breast (lesion site) and the contra-lateral normal breast (reference site) to compute the perturbation. Therefore, the chest wall underneath the breast tissue at both sites affects the imaging results. To reduce the effect of the difference between breast tissue and chest wall background geometry at the lesion and reference sites, we applied a simple correction to the perturbation formula. In this approach, we model the breast tissue-chest wall geometry with a 3D finite element model based on the information of two orthogonal ultrasound images that were captured at the same location on the lesion and reference sites. The background optical properties are estimated with a 3D FEM based two-layer model, measurement data of the reference site and a nonlinear regression algorithm based on Nelder Mead approximation. [2,3]

To reduce the effect of chest wall mismatch between the lesion and reference sites we used a normalized perturbation

$$\text{Normalized perturbation} = \frac{U0\_tar_{meas} - U0\_ref_{meas}}{U0\_ref_{meas}} = \frac{U0\_tar_{meas}}{U0\_ref_{meas}} - 1 \qquad (1)$$

In this case, the perturbation caused by the mismatch geometry of the reference and lesion sites can be reduced by multiplying and dividing the first term in equ. 1 with the calculated data based on the optical properties of the breast and chest wall layers and the 3D geometry model of reference site and lesion site, respectively (equ. 2).

$$\text{Modified perturbation} = \frac{U0\_tar_{meas}}{U0\_ref_{meas}} \times \frac{U0\_ref_{sim\_refgoem}}{U0\_ref_{sim\_targeom}} - 1 \qquad (2)$$

Since the numerator and denominator of the first term of equ. 2 are both from the measurements of the same DOT system, differences between gain and initial phase of each source-detector pairs will be canceled out. This is the same for the ratio of the calculated data.

Experimental results

In this paper we have addressed the effect of mismatch geometry for the three most common breast-tissue and chest-wall interface mismatches between the lesion and reference sites that happen in clinical studies. [4] Three different phantom experiments were performed to illustrate the cases when the depth of the two-layer interface at the reference site is deeper, shallower or tilted compared to the two-layer interface at the lesion site. [3] For each case an 8% intralipid solution and a plastisol phantom with calibrated optical properties of μ_a=0.08, μ'_s=6.5 cm^{-1} at 780nm were used as the first and the second layer, respectively. A 1cm diameter solid spherical phantom with calibrated optical properties of μ_a=0.07 and μ'_s=5.1 cm^{-1} was used as the target.

Fig. 1 demonstrates the effect of the depth mismatch for a case when the two-layer interface from the probe is shallower at the reference site compared to the target site. Top pictures at fig. 1(a) show two orthogonal ultrasound images of the two layer media at the lesion site (two left images) and two orthogonal images at the reference site (two right images). In both cases the second layer of both target and reference sites were located 2cm from the probe – no mismatch. The target is located at x=0, y=0 and z=1.4cm. The fitted optical properties of the background values at the reference site are μ_{a1}=0.03 , μ_{a2}=0.089 , μ'_{s1}=8.99 and μ'_{s2}=5.6 cm^{-1}. The reconstructed image of the absorption coefficient of the target is shown in Fig. 1(a). To demonstrate the mismatch effect, we moved the second layer at the reference site to 1.4cm from the probe (Fig 1(b)). The fitted optical properties of the two layers are obtained as μ_{a1}=0.03, μ_{a2}=0.097 , μ'_{s1}=10.49 and μ'_{s2}=7.25cm^{-1}. Fig. 1(b) shows the reconstructed image of the target. Since the second layer is shallower at the reference site, more photons are absorbed by the second layer and the reflectance measurement at the reference site is lower than that at the target site. Therefore, perturbation, which is the normalized difference between target and reference data, is reduced. To reduce the effect of mismatch between the two layer-interface at the target and reference sites, we calculated the fluence rate at the two sites with both the two-layer background geometry of the reference and target sites and with the optical values that were obtained with the nonlinear regression algorithm. Fig. 1(c) shows the reconstructed image after applying these data to the perturbation (equ. 2). The result shows a significant reduction of the effect of depth mismatch between the target and reference layers on the reconstructed image.

Second experiment shows the effect of the two-layer interface depth mismatch when the second layer's distance from the probe is deeper at the reference site compared to the target site. Top pictures in Fig 2(a) show two orthogonal ultrasound images from the lesion site (two left images) and two orthogonal images from the reference site (two right images). In this case the second layer of the target and reference sites are both located 1.4cm from the probe (no mismatch) and the target is located at x=0, y=0 and z=0.9 cm. The reconstructed image is shown in Fig. 2(a). Fig. 2(b) show the orthogonal ultrasound images of the target and reference sites when the second layer of the reference site moved to 2cm from the probe. When the interface is deeper at the reference site, the photons absorbed by the second layer decrease, thus increasing the strength of the detected signal. The artifacts that appear in the reconstructed image (Fig. 2(b)) are due to the extra perturbation produced by the background mismatch between the target and reference sites. To reduce the effect of the second layer depth mismatch between the target and reference sites, we used equ. 2 with the calculated fluence rate of a two-layer media with the fitted optical values, and the two-layer background geometry of the reference site and target site. Fig. 2(c) shows the reconstructed image after applying the correction to the perturbation formula (equ. 2). The result shows a significant reduction of the artifacts on the reconstructed images.

Fig. 3 demonstrates the effect of a tilting angle mismatch between the two-layer interface at the reference and target sites. Top pictures in Fig 3a show two orthogonal ultrasound images of the target site and two orthogonal ultrasound images of the reference site. In this case the second layer of the target and reference sites are both flat and located at 1.4cm (no mismatch) from the probe. The target is located at x=0, y=0 and z=0.9 cm. The reconstructed image is shown in Fig. 3(a). Fig. 3(b) shows the ultrasound images of the target and reference sites when the second layer of the reference site was tilted by -8 degrees. The artifacts that appear in the reconstructed image (Fig. 3(b)-bottom) are due to the extra perturbation produced by the part of the reference site with a lower absorption coefficient than the lesion site. The other part of the reference site with a higher absorption coefficient than the lesion site decreases the perturbation and shows lower absorption in the reconstructed image. Fig. 3(c) shows the reconstructed image after applying equ. 2 to the perturbation formula. The result shows that this method can help to reduce the artifacts produced by the angle mismatch.

Discussion

In this paper, we investigated the effects of depth and tilting angle mismatches between the two-layer interface of reference and lesion sites, on the image of reconstructed absorption coefficient. Depth or tilting mismatchs of the two-layer interface between reference and lesion sites can produce image artifacts, or reduce target contrast in reconstructed images. We also introduced a simple method to reduce the effect of mismatch by compensating for the extra perturbation that is produced by the background geometry mismatch of the breast-tissue and chest wall at the lesion and reference sites.

References

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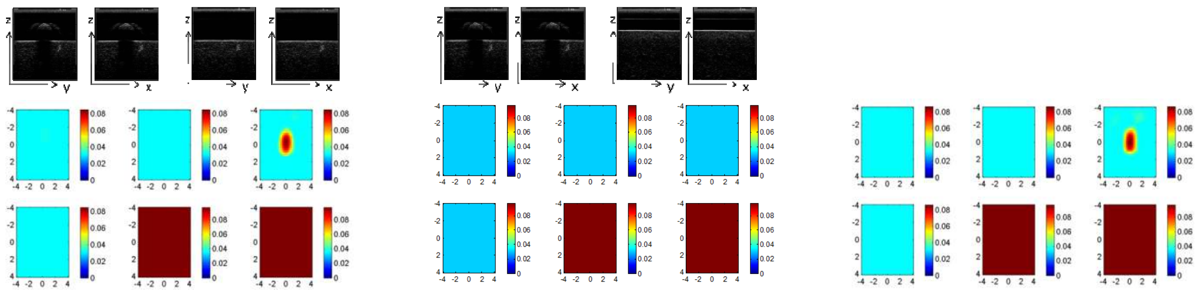
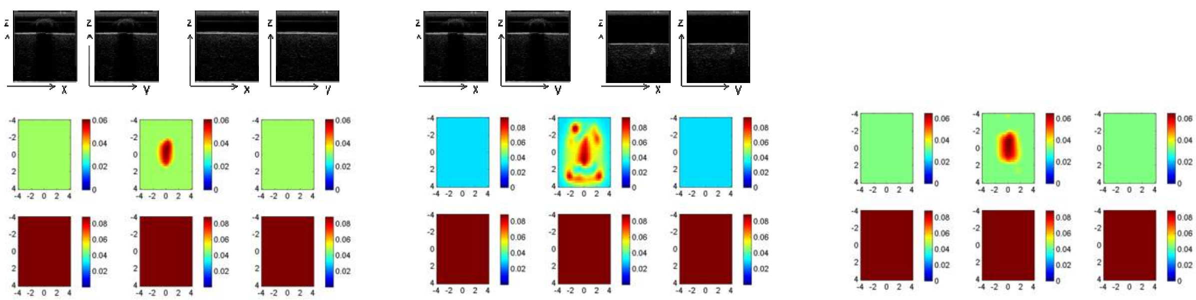


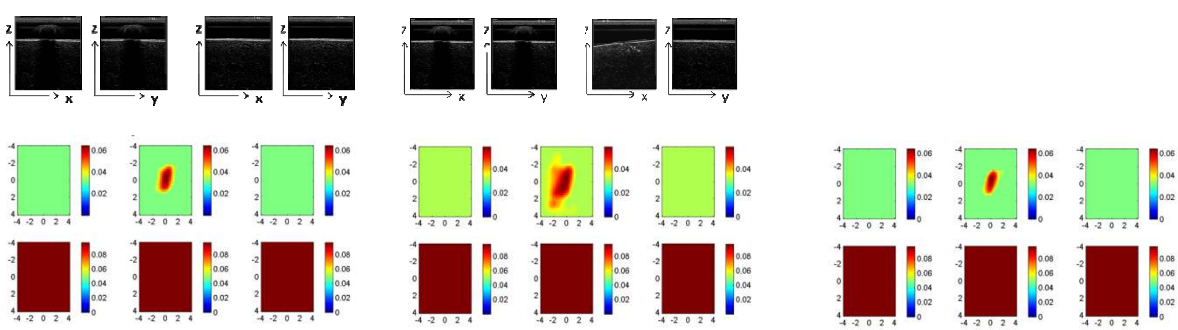
(a) (b) (c)

Fig. 1 Effect of depth mismatch when the two-layer interface depth from the probe is shallower at the reference site compared to the lesion site. Top pictures in part (a) show two orthogonal ultrasound images of the target site (two left images) and reference site (two right images).The second layer is located at 2cm depth, and target is at x=0,y=0 and z=1.4cm. (a) Reconstructed image of the target in case of no mismatch. Top pictures in part b show two orthogonal ultrasound images of target and reference sites when second layer at the target site and reference site are located at 2cm and 1.4cm, respectively. (b) Reconstructed image of the absorption coefficient of the target before applying the mismatch correction, and (c) after applying the mismatch correction formula. The scales are adjusted to show a better quantification of absorption coefficient. In the absorption map, each slice presents a spatial image of 8 cm x 8 cm obtained between 0.4 cm and 2.9 cm underneath the probe surface, with 0.5 cm spacing between slices. The format is the same in all figures.



(a) (b) (c)

Fig. 2 Effect of depth mismatch when the two layer interface distance from the probe is deeper at the reference site compared to the target site. Top pictures in part (a) show two orthogonal ultrasound images of the target site (two left images) and reference site (two right images). Both sites have second layer located at 1.4cm depth. Target is located at x=0,y=0 and z=0.9cm. (a) Reconstructed image of the target in case of no mismatch. Top pictures in Part (b) show two orthogonal ultrasound images of the target and reference sites when the second layer at the target and reference sites are located at 1.4cm and 2.0cm, respectively. (b) Reconstructed image of the absorption coefficient of the target before applying the mismatch correction, and (c) after applying the mismatch correction formula.



(a) (b) (c)

Fig. 3 Effect of tilting angle mismatch when the two layer interface has a different tilting angle at reference site compared to lesion site. Top pictures in part (a) show two orthogonal ultrasound images of the target (two left images) and reference site (two right images). Both sites have a flat second layer interface located 1.4cm from probe. Target is located at x=0,y=0 and z=0.9cm. (a) Reconstructed image of the target in case of no mismatch. Top pictures in part (b) show the ultrasound images when target site has a flat two layer interface and reference site has a two layer interface with -8 degree tilt in x direction, both at 1.4cm from the probe. (b) Reconstructed image of the absorption coefficient of the target before applying the mismatch correction, and (c) after applying the mismatch correction formula.