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Refractive effects on optical measurement of alveolar volume: A 2-D ray-tracing approach

Fatemeh N. Golabchi, Dana H. Brooks, Andrew Gouldstone, and Charles A. DiMarzio

Abstract— **Lung imaging and assessment of alveoli geometry in the lung tissue is of great importance. Optical coherence tomography (OCT) is a real-time imaging technique used for this purpose, based on near-infrared interferometry, that can image several layers of distal alveoli in the lung tissue. The OCT measurements use low coherence interferometry, where light reflections from surfaces in the tissue are used to construct 2D images of the tissue. OCT images provide better depth compared to other optical microscopy techniques such as confocal reflectance and two-photon microscopy. Therefore, it is important to detect and verify optical distortions that happens with OCT, including refractive effect at the tissue-air alveoli wall interface which is not taken into account in the OCT imaging model. In this paper, the refractive effect at the tissue-air interface of the alveoli wall is modeled using exact ray tracing and direct implementation of Snell's law, and differences between alveoli area computed from OCT imaging and those measured by exact ray tracing of the OCT signal are analyzed.**

I. INTRODUCTION

Measurement of alveolar volume across several length scales is clinically and physiologically important. At the scale of the whole lung, analysis of image contrast is important to determine efficacy of ventilation, and lung health or damage [1]. Locally, measurement of volume in single or several alveoli can provide information about stability, interdependence [2], and mechanisms of alveolar collapse or re-opening [3]. Accurate measurements of alveolar volume, specifically under mechanical ventilation or conditions of atelectatic recovery, have perhaps increased in importance due to recent studies suggesting heterogeneous recruitment [4], [5]. Specifically, during re-opening of distal alveoli, it is possible that several over-distend, leading to increased wall stresses and possible damage. As this would be important to ventilator-induced lung injury and acute respiratory distress syndrome, tools for alveolar measurement are necessary.

One tool that has been proposed for making such measurements, is optical coherence tomography (OCT) [6] which is a real-time imaging technique based on near-infrared interferometry that can non-invasively image several layers of distal alveoli in the lung tissue [3]. OCT measurements are based on optical scattering of near-infrared light entering layers of alveoli in the lung tissue. Although OCT has sufficient resolution to show alveoli structure, it suffers from distortions that impeded accurate quantitation.

One important type of distortion comes from the refraction effects that happen in tissue-air interfaces of the alveoli wall, which are not taken into account in the OCT imaging model. Other distortions are due to optical heterogeneity in the tissue that result in limited depth of imaging and high scattering from structures in the tissue such as elastin [7]. In this work we propose a model-based, computational, geometric optics method to assess the inaccuracy caused by refraction and which can lead to a systematic mechanism to correct for that distortion.

The refractive distortion is evident when one makes a comparison of OCT images of lung tissue with computational models, i.e. synthetic images constructed from simulations of other imaging modalities. For example, constructing synthetic images of the lung tissue using a finite difference time domain (FDTD) computational model of the OCT to approximate Maxwell's equations, has been shown to capture refraction, reflection, scattering, and diffraction effects that happens in the lung tissue [8]. An example of this computational model along with real OCT experimental measurements [5] are shown in Fig.1.

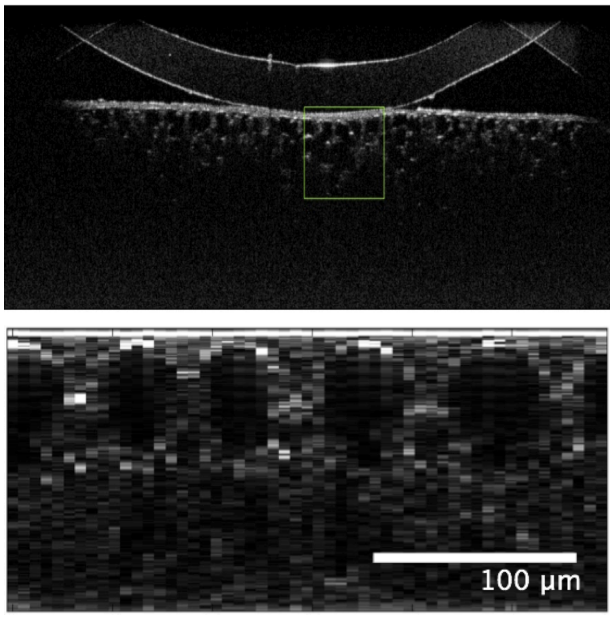

Fig. 1. Top image: real OCT image of the lung tissue showing a few layers alveoli, Bottom image: FDTD computational model of OCT for area shown in green box in the top image

Comparison of OCT measurements with the FDTD model, as well as other imaging techniques such as confocal reflectance and two-photon microscopy, has lead to increased awareness of optical distortion effects that are not considered in OCT imaging model and that result in inaccurate assessment of the alveoli volume from the OCT images [8]. This has also been seen in the direct comparison of OCT images

with conventional histology. Therefore, correction models should be applied to OCT measurements if characterization of the alveoli volume is to be utilized in clinical applications.

In this paper an exact implementation of Snell's law to model the refractive effect of the OCT signal at tissue-air interface of the alveoli wall is suggested and analyzed. The simplified alveoli model includes parallel near-infrared rays entering a single layer array of simple spherical or ellipsoidal alveoli of different sizes, eccentricities and orientations, and uses a ray tracing method to determine the effect of refraction on the path traversed by each ray. The alveoli model is considered in 2D for simplicity in this initial work, which is also compatible with real 2D OCT experimental measurements. The differences between 2D alveoli size computed directly from real OCT measurements and the sizes as assessed by ray-tracing model are analyzed and a correction model is proposed. In section II, the OCT imaging model, exact ray tracing of the OCT signal using Snell's law, and the proposed alveoli models are explained. In section III, the ray tracing results for different alveoli shapes and sizes, the comparison of these results with real OCT experimental measurements, and the resulting correction models are shown. Finally, section IV reports on the conclusions drawn from our proposed model and analysis to date.

II. METHODOLOGY

1) Optical coherence tomography (OCT): The OCT measurements are based on low coherence interferometry, where light reflections from surfaces in the tissue are used to construct 2D images of the tissue up to a depth limited by the depth of penetration of the light in the tissue. In lung tissue, OCT can achieve sub-alveolar resolution for depth up to a few layers of alveoli, i.e. distance of about 120 μm according to our measurements. The standard interpretation of OCT measurements assumes there is no refraction in the tissue-air alveoli walls and that the transverse location of the back-scattred rays are directly below the detector.

However, in reality refraction of the light waves as they enter the alveoli volume changes the propagation direction in a manner that is not accounted in the imaging model. Specifically, in the presence of objects with significant difference in the indices of refraction such as alveoli wall, the angle of rays entering and exiting the alveoli walls changes due to refraction and the OCT signal that is reflected from the alveoli wall is seen at a narrower angle and thus results in a distorted narrower alveoli. In the alveoli wall, the index of refraction of the tissue is 1.4 which is considerably higher than 1 which is the index of the refraction of the air in alveoli. Here we propose a modeling method using exact ray tracing at the alveoli wall, based on Snell's law to model this refraction that is not considered in the OCT imaging model.

2) Optical ray tracing using Snell's law: Using Snell's law, and the index of refraction for the tissue being larger than air in the alveoli, parallel rays entering a spherical alveoli wall with an angle less than the critical angle are diverged.

Assuming the rays entering the alveoli wall with angle θ, index of refraction of 1 for air and n for tissue, according to Snell's law the normalized refracted rays exit the alveoli wall with angle θ':

$$\vec{v}_n = \frac{\vec{v}}{n} + (\cos\theta' - \frac{\cos\theta}{n})\vec{n}$$

where $\vec{v}$ and $\vec{v}_n$ are normalized input and output rays at the alveoli surface, entering the alveoli. Fig. 2 is a 2D demonstration of rays entering a spherical alveoli volume and the corresponding refraction using direct implementation of Snell's law for curved surfaces, compared to what the OCT imaging model measures, where the red circles represent the location of rays as assumed by an OCT imaging model and the red x's show the true location of rays considering the refraction that occurs at the alveoli wall.

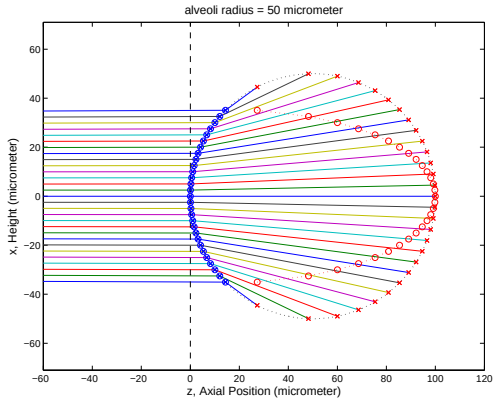

Fig. 2. Comparison of rays from OCT measurements with rays from the refraction model in a spherical alveoli. Light is incident to into the tissue surface (vertical dotted lines) from the left. The light rays encounter the alveolar surface (blue x inside blue circle). When the rays refract they encounter the boundary of the true alveolus at the red x's. However because the OCT imaging model does not take refraction into account, the apparent alveolar wall in the OCT image is seen at the locations of the red circles, thus distorting the shape from circular to elliptical.

The above model assumes a spherical geometry for the alveoli volume, i.e. a circular shape in 2D. Assuming similar refraction model, the ray tracing model is applied to other alveoli geometries including spheres of different sizes by varying the alveoli radius and ellipsoids of different sizes, eccentricities, and different orientations, i.e. ellipses in 2D.

Fig. 3 is a 2D demonstration of rays entering an ellipsoidal alveoli volume and the corresponding refraction, compared to what the OCT imaging model measures. For different alveoli geometries and sizes, the true alveoli areas are compared with alveoli areas from OCT imaging model which assumes the rays enter the alveoli wall without any refraction. For spherical geometry, the comparisons are repeated for alveoli with different diameter, and for the ellipsoidal geometry comparisons are repeated for different alveoli, first by varying eccentricity while the area and orientation are kept

constant, and then by varying the orientation angle while eccentricity and area are kept constant.

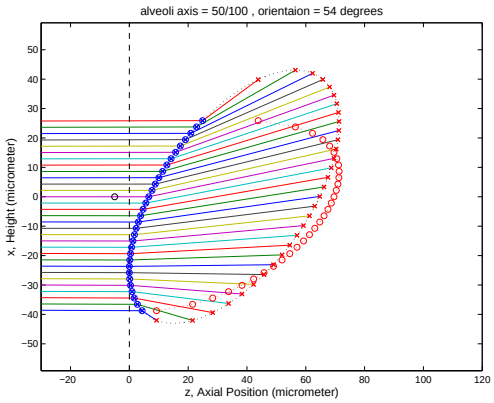

Fig. 3. Comparison of rays from OCT measurements with rays from the refraction model in an ellipsoidal alveoli model. Format is the same as in the previous figure.

III. RESULTS

Comparison of true alveoli areas computed from exact ray tracing with alveoli areas computed from OCT imaging model for different alveoli geometries are shown in Figs. 4-6. Fig. 4 shows the comparison results for circular alveoli for different alveoli radiuses. Fig. 5 shows the comparison results for ellipse alveoli for different alveoli eccentricities, while the area and the orientation angle are kept constant. Fig. 6 shows the comparison results for ellipse alveoli for different alveoli eccentricities, while the area and the orientation angle are kept constant. In all figures normalized error between the two results compared are also shown.

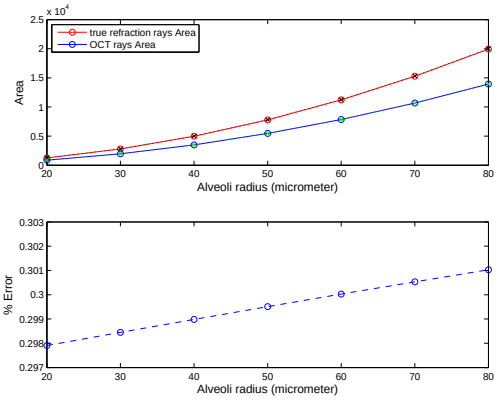

Fig. 4. Alveoli area as measured by OCT and by exact ray tracing vs. alveoli radius for a circular model. Top panel shows curves of area (in μm^2) against radius for the true alveolus and for the apparent one imaged by OCT, as a function of the alveolar radius. Bottom panel shows the % error in the area as computed from the simulated OCT image.

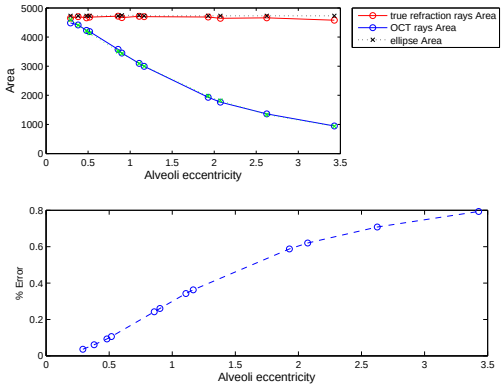

Fig. 5. Alveoli area as measured by OCT and by exact ray tracing vs. alveoli eccentricity for an ellipse model while area and orientation are kept constant. Similar format to previous figure except that the horizontal axis is eccentricity (ratio of horizontal to vertical axis) rather than area, since area is held constant.

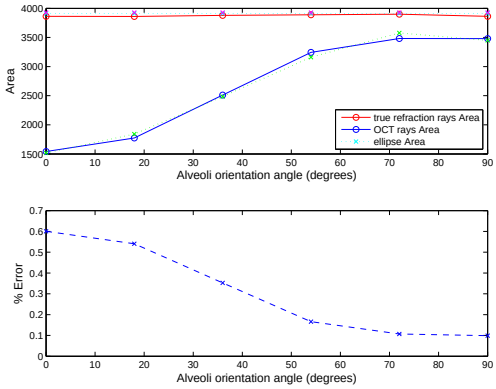

Fig. 6. Alveoli area as measured by OCT and by exact ray tracing vs. alveoli orientation angle for an ellipse model while area and eccentricity are kept constant. Same format as previous figures except for the change in independent variable.

IV. DISCUSSION

As expected, the effect of the mismatch between constant index of refraction, as assumed by the OCT image reconstruction model, and the change in index between tissue and alveoli, caused a notable decrease in apparent area of alveoli in the model compared to the true area. Figs. 2 and 3 demonstrated this decrease in detail.

In Fig. 4 the difference between true alveoli area and those measured by OCT increases with increase in the alveoli radius. This increase is quadratic and may offer a means to develop a method to convert alveoli area measured from OCT to true alveoli area, or more generally to correct the OCT images for this distortion effect. In Fig. 5, also, the difference between the two alveoli areas increases quadratically with the increase in the eccentricity, i.e. as the alveoli model changes from a vertical ellipse through a circle to a more horizontal ellipse. This again could potentially be used to correct for

the alveoli areas measured by OCT. Fig. 6 show the the difference between the two areas decrease, as a vertical ellipse is rotated from being orthogonal to the parallel rays to being parallel to them. This results suggests that for horizontal ellipses the OCT measurement are close to the true geometry. A combination of above models is expected to lead to an inversion model that can be applied to OCT measurements to correct for the refraction effect.

However, the next step before combining the above models is to also include spatially varying propagation velocity across the alveoli length. This correction model will compensate for both refraction effects and velocity effects that happen as the light travels through the tissue, which combined with a 3D version of the approach, may result in a more accurate estimate of the alveoli volume measured by OCT imaging model.

V. CONCLUSION

This paper demonstrates the refraction that happen in the tissue-air interface of the alveoli wall for rays entering the alveoli, and models the effect they have on alveoli area that is measured by OCT imaging model. The model gives a better understanding of the OCT measurements of the alveoli in the first layer by predicting the the alveoli walls to be thicker than they are, as seen in the FDTD model and OCT measurements. The model needs to be improved to allow for next layers of alveoli as well as other structures in the lung tissue. Another improvement can be achieved by taking into account the change in the speed of rays as they enter the alveoli.

VI. ACKNOWLEDGMENTS

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