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ABSTRACT

Intravascular optical coherence tomography (IVOCT) is an intravascular imaging modality which enables the visualization arterial structures at the micro-structural level. The interpretations of these structures is mainly on the basis of relative image intensities. However, even for homogeneous tissue light intensities can differ. In this study the incident light intensity is modeled to be related to the catheter position. Two factors, the distance between catheter and inner lumen wall as well as the incident angle of the light upon the lumen wall, are considered. A three-level hierarchical model is constructed to statistically validate this model to include the potential effect of different pullbacks and/or frame numbers. The model is solved using 169 images out of 9 pull-backs recorded with a St.Jude Medical IVOCT system. F-tests results indicate that both the distance and the incident angle contribute to the model statistically significantly with $p < 0.001$. Based on the results from the statistical analysis, a potential compensation method is introduced to normalize the IVOCT intensities for the catheter position effects and small shadows.

Keywords: intravascular optical coherence tomography, catheter position, distance, angle of incidence, compensation, hierarchical linear regression, statistical analysis, IVOCT.

1. INTRODUCTION

Intravascular optical coherence tomography (IVOCT) is a new imaging modality for coronary artery atherosclerosis analysis.[1] Because it uses near-infrared (NIR) light as the imaging source, it requires blood to be flushed during imaging. As long as the blood is well cleared, IVOCT can be used to visualize the arterial tissues at micro-structure level with a high axial resolution ($\sim 10 - 20\mu m$).

Recently, increasing research is being doing by looking at the correlation between tissue composition and OCT intensity values.[2,3] However, IVOCT image intensities are not always comparable even within the same 2-D image. Also, for uniform tissue, the light intensities may differ. This difference might be related to two aspects of the position of the catheter within the artery. The first aspect is the the distance between the catheter and the arterial wall, and the second aspect is the incident angle of the light upon the arterial wall.

This study aims at exploring these two aforementioned factors which may cause light intensity loss other than tissue attenuation. To the best of our knowledge, no specific research has been carried out to analyze the effect of these factors on the light intensity.

The structure of this paper is as follows. In Sec. 2.1, inspired by aforementioned work, the incident light intensity is modeled to be linearly related to both distance and incident angle. In Sec. 2.2 a multilevel linear model is constructed to validate this model taking into consideration patients and image slices as random factors. Linear regression is used to estimate the parameters of the linear model. Then, in Sec. 2.3, a potential compensation method is suggested. Results of the estimation and compensation method are presented in Sec. 3 and discussed in Sec. 4. Finally, in Sec. 5 conclusions are drawn.

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2. MATERIALS AND METHODOLOGY

The first order approximation of IVOCT signal intensity at a depth x can be modeled as:[4]

$$I_b(x) \cong \frac{1}{2} I_{in} \mu_b T(x) \cdot e^{-2\mu_t x}. \tag{1}$$

where I_{in} denotes the initial incident light intensity which enters the medium, $I_b(x)$ denotes the light intensity which is backscattered from distance x, μ_b represents the backscattering coefficient, μ_t is the total attenuation (summation of scattering and absorbing) coefficient and $2x$ indicates that the light goes back and forth from the start point to the point at distance x in a certain homogeneous medium. $T(x)$ is the confocal function which is defined as:

$$T(x) = \left[\left(\frac{x - z_0}{z_R} \right)^2 + 1 \right]^{-1/2}. \tag{2}$$

where z_0, z_R are the beam waist and the Rayleigh length, respectively.

2.1 Light transmission model

In Eq. (1), the light incident upon the arterial wall, I_{in}, can be related to the distance between the catheter and the arterial wall, and the incident angle.

If the blood is flushed correctly, the flush medium can be considered to be homogeneous, which means the light decay obeys the Lambert-Beer's law. The incident angle effect can be represented as the Fresnel transmission equation,[4,5] which describes the transmitted intensity decay when light propagates from one medium to another.

The Fresnel transmission equation is denoted as $Tr(\theta, n_i, n_t)$, which is calculated with the incident angle θ, index of refraction of the incident medium n_i and transmission medium n_t. With these two factors, the incident light upon the arterial wall can be modeled as:

$$I_{in} \sim I_0 \cdot Tr(\theta, n_i, n_t)^{\beta} \cdot e^{-R_l x}, \tag{3}$$

where I_0 represents the source light intensity which is a constant for one pullback; x is the distance that light travels before reaching the artery wall; R_l is the loss intensity per unit distance. With Eqs. (2) and (3) substituted, the logarithm can be taken for both sides of Eq. (1):

$$\ln I_b(x) \cong -R_l x + \beta \ln Tr(\theta, n_i, n_t) + \ln T(d + x_t) - 2\mu_t x_t + C(I_0, \mu_b), \tag{4}$$

where $C(I_0, \mu_b) \cong \ln I_0 \cdot \mu_b + E$, and E is an additive noise term with a zero mean. In the right side of Eq. (4), there are four items related to the catheter position including the distance from the catheter to the artery wall x, and the depth inside the lumen wall x_t. In order to analyze the relationship of the image intensities and these two factors, the model was implemented as follows.

- Intensities of only a thin layer on top of artery wall are used for the statistical analysis, so that $x_t \approx 0$.
- The confocal function term, $\ln[T(d) + x_t]$, was linearly approximated. As x_t approximates to zero, the confocal function term is $\ln[T(x)]$. It can be linearly approximated as:

  $$L(d) \models x \cdot k(z_R, z_0), x \in (0, a). \tag{5}$$

  An example of the linear approximation can be seen in Fig. (1), where the parameters estimated in Ughi's works were used, where $z_0 \approx 1.5\ mm$ (focal point) and $z_R \approx 2mm$ (Rayleigh length) for the St. Jude $C7$ Dragonfly catheter.
- For the calculation of the Fresnel transmission ratio,[5] the index of refraction of the flush solution is read from the data as 1.449 mm. The transmission medium was considered to be intima tissue with an index of refraction $n_t \approx 1.338\ mm$.[6] Therefore, the incident angle is the only variable during the calculation of the transmission ratio for each point. The incident angle of each A-line was estimated using least-square-fitting of the arterial wall contour around each incident point in the Cartesian image.

Ultimately, the applied model becomes:

$$\ln I_b(x) = -(R_l - k) \cdot x + \ln Tr(\theta) + C(I_0, \mu_b). \tag{6}$$

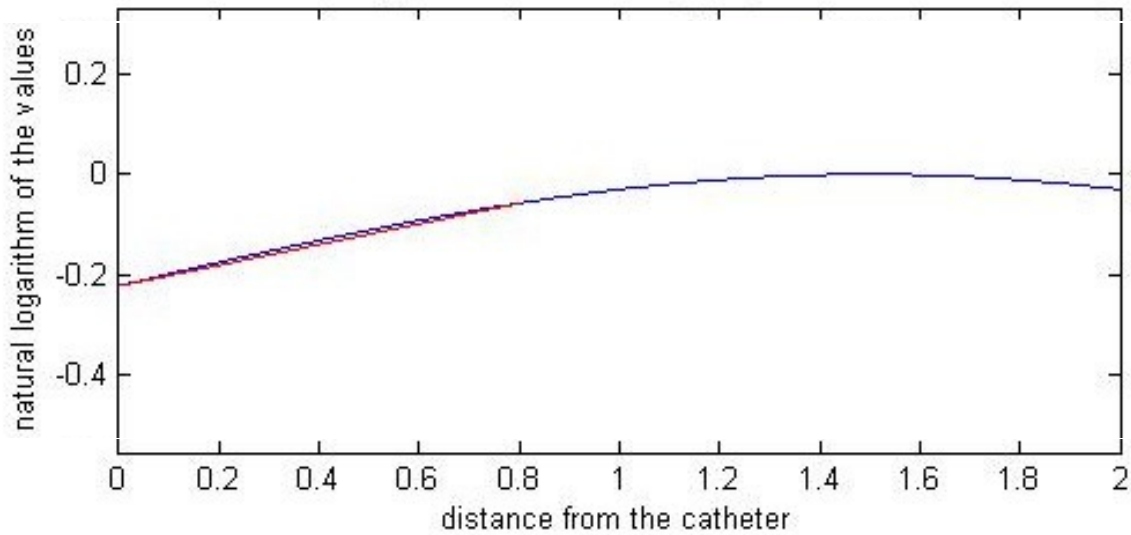

Figure 1: Linear approximation of confocal function

2.2 Hierarchical Linear Model

The model proposed in Sec. 2.1 can be solved with a hierarchical linear model.[7] The A-lines can be hierarchized into different frames, which in turn can be hierarchized into different pull-backs. Based on this observation, a three-level linear model is considered for this study (see Fig. (2)). Level one is the potential effect caused by different pull-back runs, and level two is modeled to consider the effect of different frame numbers. Linear regression for the given model should be done with both of these layers. So the hierarchical linear model used in this study is:

$$\ln I_b(x) = \beta_0 + \beta_1 \cdot x + \beta_2 \cdot \ln Tr(\theta). \tag{7}$$

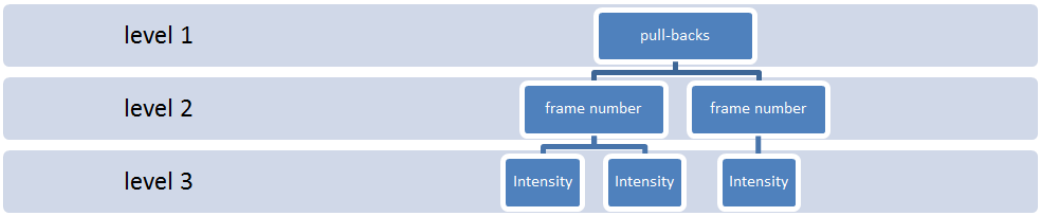

Figure 2: Multi-level linear model

2.3 Image compensation

With the linear regression, parameters β_0, β_1 and β_2 can be estimated. Based on the linear regression model, a compensation function can be derived to normalize the catheter-position-determined effect:

$$I_{compensated} = \frac{I_{original} \cdot e^{\hat{\beta}_0}}{I_b(x)}. \tag{8}$$

where the constant β_0 is the estimation result.

3. RESULT

3.1 Hierarchical linear regression

In this study, 169 IVOCT slices from 9 patients were used. The slices were selected to be non-pathological and containing different catheter positions. In the hierarchical linear regression, three fixed effects and two random effects were considered.

Three effects, light-source-dependent, distance-dependent and the incident-angle-dependent, statistically significantly contribute to the model with $p < 0.001$. The significance test results were reported as follows. When the logarithm of measured OCT intensity was predicted, the constant term, β_0 was estimated as $\beta_0 = 6.5121$, with the standard error of $SE = 0.1615$, and $p < 0.001$; The distance-related term was estimated as $\beta_1 = -0.0023$ with $SE = 8,1427E-006$ and $p < 0.001$); the angle-related term was estimated as $\beta = 2.8178$ with $SE = 0.0725$ and $p < 0.001$) $-$ all these three fixed effects were significant predictors.

The covariance estimation of the two random factors, frame number and pull-back number, indicated that both of them contributed almost two thirds of the total variance (about 0.368). The covariance parameter of the frame number was estimated as $Nu = 0.0060$ with the $SE = 0.0007$ and $p < 0.001$; The covariance parameter of the pull-back number was estimated as $Nu = 0.2342$ with $SE = 0.1173$, and the $p < 0.05$. Furthermore, it was worth noting that the covariance contribution of the frame number is relatively very small ($\sim 1.6\%$) comparing to the other contributors. Based on this observation, this random effect can be ignored during modeling.

Another random factor is the residual ($Epsilon = 0.1276, SE = 0.0006, p < 0.0001$), the histogram can be seen in Fig. 3. It was distributed quite symmetrically around zero with a mean value of -2.01$E-11$ and a standard deviation of 0.3568, thus it is indicated that the data fits the model quite well.

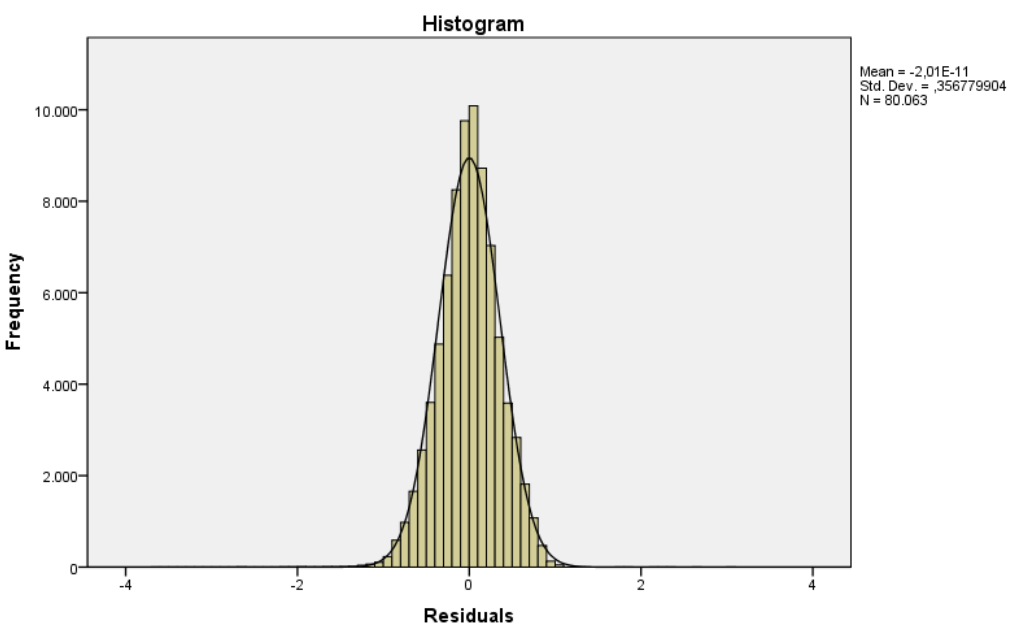

Figure 3: Histogram of residuals from regression.

3.2 Image compensation

Results of the image compensation method are shown in Fig. 4. Images on the left are original images and images on the right are compensated images. In the top original image, there is a distinctive intensity loss, while in the compensated image , the light intensities are more homogeneous and detailed information is enhanced in deeper parts. The bottom image of the figure shows that the shadow at 1 o'clock in the original image was compensated and the shadow removed.

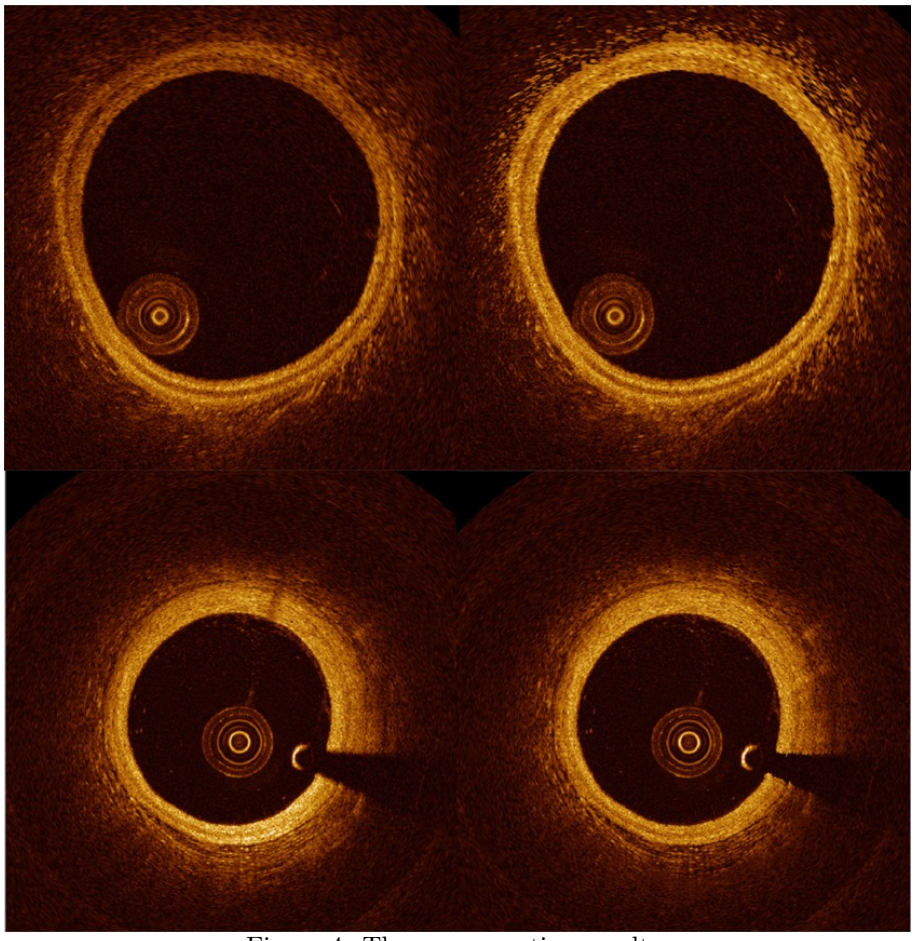

Figure 4: The compensation results.

4. DISCUSSION

The main goal of this study was to investigate the effects to the IVOCT intensities regarding to the distance between the catheter and the arterial wall, and the incident angle of the light beam upon the arterial wall. These two factors are both related to the catheter position. Results showed the relationship is statistically significant.

To the best of our knowledge, this is the first quantitative analysis regarding the effect of the catheter position to IVOCT intensities. Not only the significance has been revealed, the linear regression model also shows that as either distance or the angle increase, the IVOCT intensities will decrease.

Awareness of this relationship is very important for the acquisition and the interpretation of the IVOCT images. This relationship indicates that the best position for IVOCT images acquisition is when both the distance variance and incident angle are minimized, that is when the catheter is at the center of the artery lumen. When the images were acquired with an eccentric catheter position, the quantitative analysis within large regions of interest may contain bias, which might cause misinterpretation of the arterial tissues.

The proposed compensation indicates that these catheter position-dependent effects can be normalized with the regression results mathematically. However, the hierarchical linear regression results also indicates that the regression model for the prediction is significantly different from pullback to pullback. That is to say, the regression parameters could depend on the catheter or the image acquisition system. So the compensation requires a specific linear regression result for each pullback. In other words, for each pullback the parameters have to be determined again.

5. CONCLUSION

In this paper, the effects of the catheter position on IVOCT image intensities were analyzed regarding two aspects, the distance between the catheter and the arterial wall, and the incident angle of the light upon the arterial wall. Both of these factors are statistically significantly related to the IVOCT images. Awareness of these effects is very important for the acquisition, the reconstruction and the interpretation of IVOCT images. A potential compensation algorithm has been proposed based on the linear regression result.

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