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Arterial Cross-Section Measurements from Dual Energy Transvenous Coronary Angiography Images

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Abstract

The synchrotron based coronary angiography project at the National Synchrotron Light Source[1] obtains images of coronary arteries using the digital subtraction technique after a distal venous injection of an iodine contrast agent[2]. It allows two areal mass densities to be calculated from these images; one of the iodine and one of the water. Analysis procedures have been developed to arrive at these areal mass densities with corrections to the values being made for detector cross-talk and beam harmonics. From the iodine mass density distribution the relative arterial cross-section area is determined by a line integration across the arterial feature[3,4]. Results will be given for an iodine tube phantom showing that the relative area of a feature whose lateral dimensions are smaller than the detector pixel resolution can be determined to a few percent. Also, results will be shown from a human image, showing the relative area of the right coronary artery mapped through a region of a previous stenosis subsequently treated by balloon angioplasty. Finally, limitation of the technique and plans to validate and improve the analysis will be discussed.

I. INTRODUCTION

Synchrotron radiation based coronary angiography takes advantage of the very intense, continuous x-ray spectrum produced by these sources allowing the monochromatization of the white beam[5].

Two monochromatic x-ray beams whose energies are just below (- index) and above (+ index) the iodine K absorption edge are used to image the coronary system resulting in two simultaneously acquired transmission images of the same volume. The number of photons in each image impinging on the detector is given by

$$N(E) = N_0(E) \cdot \exp\left(-\sum_i \left(\frac{\mu(E)}{\rho}\right)_i \cdot \rho_i \cdot t_i\right) \quad (1)$$

where $N_0(E)$ is the incident energy dependent photon flux, $(\mu(E)/\rho)_i$ is the mass absorption coefficient, ρ_i the density of the materials and t_i the transmitted path length. The subscript i denotes the different tissues and materials: iodine (I), bone (b) and soft tissue (t).

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The logarithmic subtraction of the two numbers results in an almost complete elimination of all materials other than iodine and the obtained signal is given by

$$S = \ln\left(\frac{N_+}{N_-}\right) = \sum_i \left(\frac{\Delta\mu}{\rho}\right)_i \cdot \rho_i \cdot t_i \quad (2)$$

In the present line-scan system, the monochromatic fan beams are produced by means of a bent Laue monochromator[6]. The two beams are separated by ≈ 300 eV in energy and their bandwidth amounts to 140 eV. After crossing at the patient position they are detected separately by a double array Si(Li)-detector[7], working in current mode. The current flow from each 0.25 mm pixel is digitized. By scanning the patient vertically through the 0.5 mm high beams (at the patient) with a speed of 12 cm/s, a full image is built up from 250 horizontal lines, each line being acquired in 4 ms integration time.

II. ABSOLUTE IODINE MASS DENSITY

A. Removal of the residual tissue background

The raw data images are measurements of the transmission through the object at energies just above and below the iodine absorption edge. Equation (2) shows, that besides the iodine signal there is a spurious background signal arising from the other body tissues.

In addition to the desired imaging energies about the I K-edge, there are multiples of this energy due to higher order allowed reflections from the monochromator and the incident spectrum on the monochromator.

Since there are measurements at two energies, one can solve for two quantities at each pixel location. The equations will be solved for iodine and water.

The detector response as a function of the incident photon energy E can be modeled as

$$R(E) = e \cdot \frac{E}{\epsilon_{e-h}} \cdot \frac{\mu_{ea}}{\mu_T} \left[1 - e^{-\mu_T^D \cdot t_{Det}} \right] \quad (3)$$

where e is the electron charge, ϵ_{e-h} the energy required to create an electron hole pair, μ_T the total and μ_{ea} the energy absorption coefficient. The transmission through the object is modeled by

$$I(E) = I_0(E) \cdot e^{-(\mu(E)/\rho)_I \cdot \rho_I \cdot t_I} \cdot e^{-(\mu(E)/\rho)_T \cdot \rho_T \cdot t_T} \quad (4)$$

where $I_0(E)$ is the energy dependent incident photon flux and $\rho_I \cdot t_I$ and $\rho_T \cdot t_T$ are the iodine and tissue mass densities, σ_I and σ_T . If I_0 is the number of photons that would strike a detector pixel with an unobstructed beam, then the charge liberated with the object in the beam is:

$$C(E) = I(E)R(E) \\ = I_0(E)R(E)e^{-(\mu(E)/\rho)_I \cdot \rho_I \cdot t_I} e^{-(\mu(E)/\rho)_T \cdot \rho_T \cdot t_T} \quad (5)$$

Since imaging occurs at discrete energy bands above and below the I K-edge and at multiplies of these energies the total charge released in a pixel can be written as:

$$C_{\pm} = \sum_{i=1}^H I_0(i \cdot E_{\pm}) R(i \cdot E_{\pm}) e^{-(\mu(i \cdot E_{\pm})/\rho)_I \cdot \sigma_I} \cdot e^{-(\mu(i \cdot E_{\pm})/\rho)_T \cdot \sigma_T} \quad (6)$$

The relative harmonic content is known and the relationship between the detector output and calibrated water absorbers is known:

$$C_{\pm}^c = \sum_{i=1}^H I_0(i \cdot E_{\pm}) \cdot R(i \cdot E_{\pm}) \cdot e^{-(\mu(i \cdot E_{\pm})/\rho)_w \cdot \sigma_w} \quad (7)$$

Since $C_{\pm}^0$ is measured and σ_w^c is known then the scale factors $I_0(i \cdot E_{\pm}) R(i \cdot E_{\pm})$ can be determined. These factors will be called: $\bar{C}_{\pm}^0$.

These are the charges released when the unobstructed beam strikes the detector. When an iodine containing object is placed in the beams then

$$C_{\pm} = \sum_{i=1}^H \bar{C}_{\pm}^0 \cdot e^{-(\mu(i \cdot E_{\pm})/\rho)_I \cdot \sigma_I} \cdot e^{-(\mu(i \cdot E_{\pm})/\rho)_T \cdot \sigma_T} \quad (8)$$

From this measurement of $C_{\pm}$ and $C_{\pm}^0$ we would like to determine σ_T and σ_I . Ignoring the harmonic contributions and using only the $i = 1$ term from equation (8) we find

$$\ln(C_{+}/\bar{C}_{+}^0) = -(\mu(E_{+})/\rho)_I - (\mu(E_{+})/\rho)_T \quad (9a,b)$$

and

$$\ln(C_{-}/\bar{C}_{-}^0) = -(\mu(E_{-})/\rho)_I - (\mu(E_{-})/\rho)_T$$

and for the iodine and tissue mass densities:

$$\sigma_I = \frac{\frac{\mu(E_{+})_T}{\rho} \cdot \ln(\frac{C_{-}}{\bar{C}_{-}^0}) - \frac{\mu(E_{-})_T}{\rho} \cdot \ln(\frac{C_{+}}{\bar{C}_{+}^0})}{\frac{\mu(E_{+})_I}{\rho} \cdot \frac{\mu(E_{-})_T}{\rho} - \frac{\mu(E_{-})_I}{\rho} \cdot \frac{\mu(E_{+})_T}{\rho}} \quad (10a,b)$$

and

$$\sigma_T = \frac{\frac{\mu(E_{-})_I}{\rho} \cdot \ln(\frac{C_{+}}{\bar{C}_{+}^0}) - \frac{\mu(E_{+})_I}{\rho} \cdot \ln(\frac{C_{-}}{\bar{C}_{-}^0})}{\frac{\mu(E_{+})_I}{\rho} \cdot \frac{\mu(E_{-})_T}{\rho} - \frac{\mu(E_{-})_I}{\rho} \cdot \frac{\mu(E_{+})_T}{\rho}}$$

B. Harmonic Removal

In formula (9) the contributions owing to the harmonics are ignored. After passing through the patient, the 99 keV component of the incident beam can account for several percent of the detected signal in the image. Correction for the contamination is based on the known harmonic content of the incident beam and detector response.

One can summarize the terms arising from the harmonic contamination to

$$C_{\pm}^H = \sum_{i=2}^H \bar{C}_{\pm}^i \cdot e^{-(\mu(i \cdot E_{\pm})/\rho)_I \cdot \sigma_I} \cdot e^{-(\mu(i \cdot E_{\pm})/\rho)_T \cdot \sigma_T} \quad (11)$$

By substituting $r_{\pm}^i = C_{\pm}^i / \bar{C}_{\pm}^i$ the formula equivalent to (8) becomes

$$r_{\pm} = e^{-(\mu(E_{\pm})/\rho)_I} \cdot e^{-(\mu(E_{\pm})/\rho)_T} + r_{\pm}^H \quad (12)$$

This equation can be solved iteratively for the mass densities σ_T and σ_I using the following iteration loop:

- 1) Solve equation (12) for σ_T and σ_I ignoring $r_{\pm}^H$;
- 2) Use σ_T and σ_I to calculate a harmonic response $r_{\pm}^H$;
- 3) Subtract $r_{\pm}^H$ from $r_{\pm}$ to give a $r_{\pm}^I$;
- 4) Use this $r_{\pm}^I$ to solve for harmonic corrected σ_T and σ_I ;
- 5) If σ_T and σ_I have converged then exit - else go back to #2

III. TRACING OF ARTERIES

A. Artery detection

Artery detection is carried out on the original subtraction images searching for the arterial centerline and its borders separately. We start by removing low frequency background using the unsharp masking technique. For the border detection we additionally apply the Sobel operator to the unsharp masked image. By choosing an appropriate threshold value we obtain a binary image that is subsequently thinned to deliver a centerline and borderlines of one pixel thickness.

The centerline image is automatically analyzed to reject all residual non-arterial features. The remaining main artery centerline is used as a guide line for the correction of presumably erroneous borderline segments. These two images are passed to the following tracing algorithm.

B. The tracing algorithm

The smoothed (7 point) centerline path is segmented into equally spaced sub-pixel size steps. For each step the arterial dimension transverse to the arterial path is calculated and a transverse cut of the iodine mass density distribution is carried out by bilinear interpolation of the data. This procedure results in a straightened artery that is displayed.

The transverse arterial dimension is used to define a background band around the artery. Finally, we integrate the

iodine mass density within the transverse dimensions of the artery at each step and subtract the background. If we would know the contrast agent density in the artery ρ_I we could directly calculate its cross-section from this integral by:

$$A = \frac{1}{\rho_I} \int \rho_I t_I dp. \quad (13)$$

Since ρ_I is generally not known then the integral is $\rho_I A$. If we assume the artery cross-section to be round at some point along the arterial path, then the area can be estimated from the its diameter, d_a , which will determine ρ_I , and A is known along the path:

$$\rho_I = \frac{4 \int \rho_I t_I dp}{\pi d_a^2}. \quad (14)$$

Even without assuming a round arterial profile the integral $\rho_I A$ will always yield the relative projected lumen of the vessel since significant changes in ρ_I along the arterial path can be excluded due to the venous injection of the contrast material.

IV. RESULTS

A. Iodine Tube Phantom

Figure 1a shows an absolute iodine image if a phantom with diluted iodine contrast agent filling various diameter holes. The iodine concentration is 18.5 mg/ml. The phantom was imaged with 4 mm of Al and 6.5 cm of water in the beams. The smallest bore hole region in the box is shown magnified in Figure 1b. The smallest tube has a diameter of 0.34 mm which is smaller than the 0.5 mm detector pixel size. Figure 1c shows the transverse integral across the tube. The tube has diameters of 3.175 mm to 0.343 mm to 0.787 mm from left to right. Using the iodine integral of the 3.175 mm region as a reference, the iodine integral of the 0.343 mm diameter region is 0.0122 compared to a known geometrical area ratio of 0.0117, an estimated error of 4.3 %. The iodine integral ratio of the 0.787 mm region is 0.0674 compared to the geometrical ratio of 0.615, an estimated error of 9.6 %. The other stenotic regions of the phantoms give similar results when integral ratios are compared to known ratios.

Figure 1: a) Iodine mass density distribution of an iodine tube phantom

b) Image detail of lateral size (.343 mm) smaller than detector pixel resolution (.5 mm). c) Transverse iodine line integral along the path

B. Human Image

Figure 2a shows the iodine mass density distribution as reconstructed from a human subtraction image in a LAO 30° projection. It shows the right coronary artery (RCA) down to its distal region as well as the left main coronary artery, dividing into the left anterior descending (LAD) and circumflex arteries. The bilinear interpolated and straightened RCA is represented in Figure 2b. It shows a stenosis previously treated by balloon angioplasty. The transverse iodine line integral along the path of the artery is plotted in figure 2c. The stenotic region is clearly distinguishable in the plot.

Figure 2: a) Iodine mass density distribution from a LAO 30° projection with outlined right coronary artery (RCA).

V. LIMITATIONS

A. Water model

Since we have only two measurements we can solve only for two parameters. This does not account for the complex structure of the human body. The bone mass absorption coefficients show significant deviations from water at the different energies. A third measure taken at a different energy would allow to take this into account and to correct for it.

B. Stereoscopic Systems

Even though the demonstrated technique allows the determination of transverse integrals of the absolute iodine mass density along the path of an arterial feature, the data does not necessarily reflect changes in the arterial lumen correctly. The measured integral depends on the projection angle of the vessel on the detector plane. It is thus not possible to distinguish between a change in the density - cross-section product and a change in the projection angle.

To overcome this deficiency it is proposed to install a system for stereoscopic coronary angiography at the NSLS[8]. Such a system would consist of two bent Laue monochromators of different crystal reflections. The Si(111) and Si(311) reflections would form an angle of 6.27 ° between the two beams crossing at the patient position and would be registered by two dual array detectors.

The two simultaneously acquired images allow the spatial reconstruction of the vessels enabling the correction of the transverse iodine integral (eq. 13) for projection shortening. Furthermore, in conjunction with the knowledge of the arterial dimension along its path, the integral will give a measure of the vessel asymmetry in regions of stenosis.

Figure 2 b) Bilinear interpolated and straightened RCA.

c) Transverse iodine line integral along the path

Finally, a stereoscopic display of the images will ease the interpretation of the angiograms by physicians because of the elimination of projection dependent ambiguities.

VI. CONCLUSIONS

We demonstrated the quantitative analysis of the luminal cross-section of coronary arteries in synchrotron radiation based transvenous angiography images.

An iodine tube phantom image shows that the relative area of a feature whose lateral dimensions are smaller than the detector pixel resolution can be determined to a few percent under conditions where the beam attenuation and the contrast agent concentration are comparable to those in human imaging scans.

The right coronary artery from a human image can be evaluated in a projected length of 120 mm. Its transverse iodine line integral reveals a stenosis in the region previously treated by balloon angioplasty.

A stereoscopic extension of the existing medical research facility would allow the spatial reconstruction of arteries and subsequently the determination of arterial cross-sections on an absolute scale.

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