A WAY TO REDUCE THE ARTIFACTS CAUSED BY INTENSELY ABSORBING AREAS IN COMPUTED TOMOGRAPHY

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ABSTRACT

Artifacts caused by intensely absorbing areas are encountered in computed tomography and may obscure or simulate pathology in medical applications, hide or mimic the cracks and cavities in the devices at industrial applications. We simulated sinograms with different levels of absorption to demonstrate the artifacts dynamics. If the analysis of the measured data shows the presence of strongly absorbing areas in the object under study we propose to use quadratic programming technique for solving the inverse problem. Although the technique is time-consuming it allows us to avoid the typical artifacts. We compare the images reconstructed with different techniques including the proposed one.

OVERVIEW

One known problem in computed tomography (CT) is the appearance of metal artifacts in reconstructed CT images (Barrett and Keat 2004; Boas and Fleischmann, 2012). This effect is due to the presence of intensively absorbing areas inside the objects or in the human body, such as dental fillings or hip prostheses, and leads to inconsistencies in the Radon or projection space. These inconsistencies always observed in the integral attenuation values are due to the polychromatic X-ray spectrum produced by the X-ray tube and may occur in the monochromatic mode. This often leads to artifacts in the reconstructed images in the form of dark stripes between metal objects with light, pin-striped lines covering the surrounding tissue. Besides beam hardening, another source of metal artifacts is high ratio between scattered and primary radiation, that causes a low SNR in the metal shadow. Additionally, the partial volume effect is a source of metal artifacts in transmission CT images. Especially in those cases in which the radiation is completely absorbed due to the thickness of the materials, very bright stripes are found radially around the object; thus the whole image loses its diagnostic value (Buzug, 2008).

Some manufacturers use different tricks to avoid these artifacts, like hardware corrections (automatic x-ray tube current modulation), reconstructed results filtering and adaptive filtration to reduce the streaking in photon-starved images during sinogram pre-processing. This software correction smoothes the attenuation profile in areas of high attenuation before the image is reconstructed (Barrett and Keat, 2004). Others provide tomography measurements using multi-energy scanning (Bamberg et al., 2011).

On the other hand, statistical reconstruction techniques are especially capable of dealing with the metal artifact reduction problem. The core of some of these methods consists of an a priori knowledge of the statistical distribution of photon counts and a system matrix that models the physical X-ray absorption process (Muller, 2006; Buzug, 2008). Each row of the system matrix represents a single X-ray beam running through the measurement volume. Contrary to filtered back-projection, the influence of each single beam on the image reconstruction can be weighted separately. In this way, beams through the metal objects can be treated appropriately. This maximum likelihood (MLEM) algorithm (Buzug, 2008) and modified MLEM algorithm is called λ-MLEM (Oehler and Buzug, 2007) improves image quality compared with pure interpolation or missing data concept (Amirkhanov et al., 2012). An expert system based on fuzzy logic (Martorelli, 2011) is used to process the CT images and to clean them automatically leaving the decisional phase to the computer. We propose to use quadratic programming technique for solving the inverse problem. Although the technique is time-consuming it allows us to avoid the typical artifacts. The quadratic programming technique was used to remove the ring artifacts in transmission tomography (Titarenko et al., 2010). We first propose to implement it to reduce artifacts caused by intensely absorbing areas. We compare the images reconstructed with different techniques including the proposed one.

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SIMULATION

This part of the article aims to track the artifacts caused by intensively absorbing areas through mathematical analysis. In our model only photoelectric effect of "true absorption" is considered. The scanning scheme is parallel (Figure 1) and the monochromatic mode used for probing.

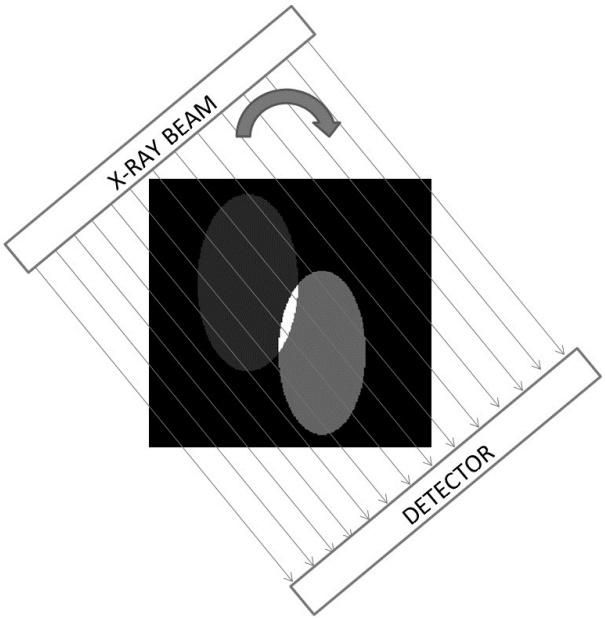

Figure 1: Schematic diagram of CT

We discuss the problem directly in discrete form as later we will use algebraic technique RegART (Prun et al., 2013) to reconstruct the images. We superimposed a square grid on the image $\mu(x,y)$. We assume that in each pixel the function $\mu(x,y)$ is constant. Let μ_i denotes this constant value in the ith pixel. The values of the pixels are considered as variables collected in a vector. Like the image the projections will also be given in one-index representation. Let I^j denotes the value measured by a detector cell at particular rotation angle. Then the size of the vector I is equal to the number of the detector cells multiplied by the rotation angles. Then the absorption process is described by the expression:

$$I^j = I_0 \exp\left(-\sum_i \mu_i \, \omega_{ij}\right) \qquad (1)$$

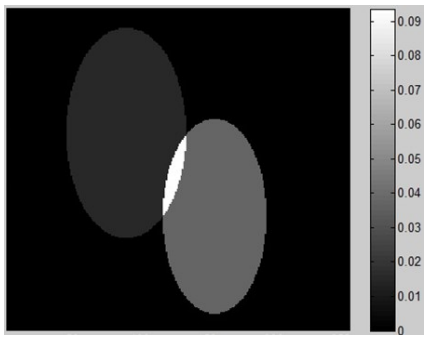

Figure 2: Phantom used for simulations

Here μ is the linear absorption coefficient, ω_{ij} represents the contribution of the ith pixel to the jth component of the vector I.

The phantom presented on Figure 2 was used for simulation of the sinograms. The values of the absorption coefficients were chosen for 8keV. The description of the phantom: first ellipse is filled by Si, the second one is filled by Ge, the values in the intersection area are consistent with Gd. 2D parallel projection data contain 180 projections, and each projection contains 256 simulated detector measurements. The images below were reconstructed on a 600 × 600 grid with different pixel spacing. We excluded Poisson error (photon counting noise). As registered signal is a discrete signal we use the approximation (APP): to write int(I^j) in the jth component of the vector instead of I^j.

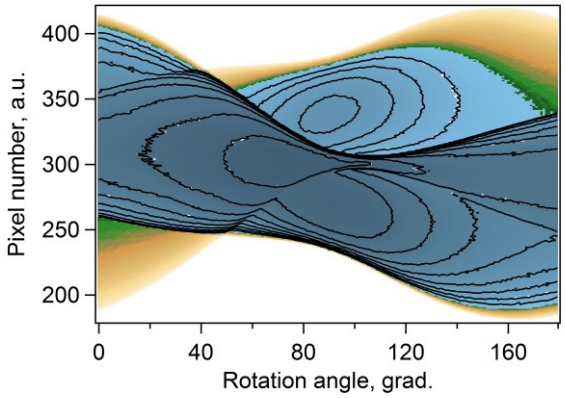

Figure 3: Sinogram for 0.5 μm pixel size

The simulated sinogram for 0.5 μm pixel size is presented in Figure 3. Fragments of the reconstructed with RegART images are presented on Figure 4.

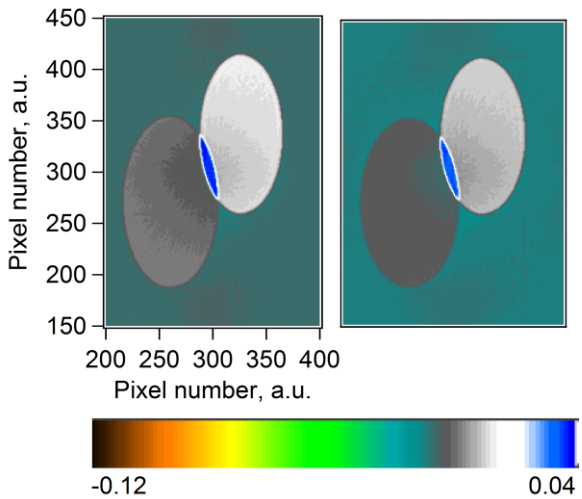

Figure 4: Fragments of the reconstructed image.

Difference between two images (left and right) is that for the first one the sinogram was calculated according

to Expression (1), and for the second one the approximation (APP) was used. The absence of strong differences in the images gives us the right to continue to use this approximation.

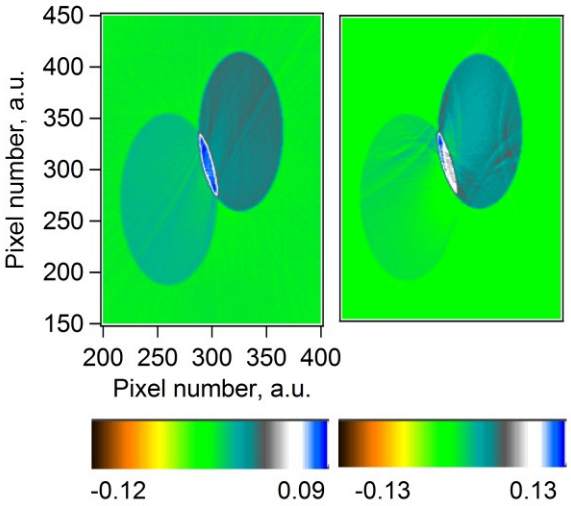

Figure 5: Fragments of the reconstructed image

Once we have increased the size of the pixel (Figure 5. Left - pixel size is 1.0 μm; right - pixel size is 1.5 μm.), thereby increasing the absorption, we are witnessing the emergence of characteristic artifacts (Figure 6. Pixel size is 1.5 μm.) in the reconstructed image. Images are reconstructed with RegART.

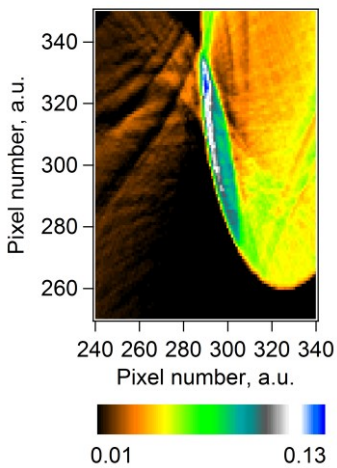

Figure 6: Fragment of the reconstructed image

Figure 7 shows a sinogram calculated for the case where the pixel size is 1.5 microns. The middle part of the sinogram has intensity decrease to almost zero. We have introduced a minimum threshold to approximate the real measurements. To reduce the artifacts that may obscure or simulate pathology in medical applications, hide or mimic the cracks (Figure 6) and cavities in the devices at industrial applications we proposed to solve the inverse problem by the quadratic programming technique instead of RegART.

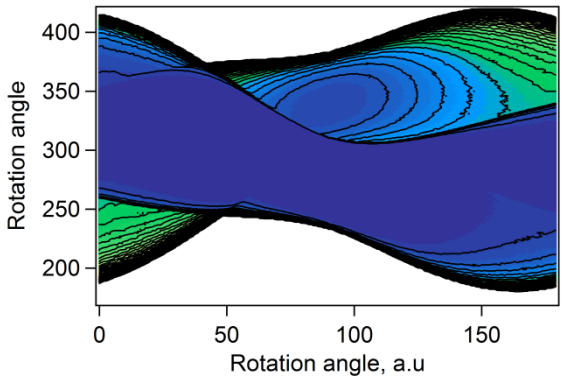

Figure 7. Sinogram for 1.5 μm pixel size

RESULTS

The quadratic programming technique was applied to reduce the artifacts caused by intensely absorbing areas. A phantom of size 10 by 10 pixels was used (Figure 8a). It consists of 2 parts. The first from left looks like a bracket and has high level of absorption. The second is a small cross which has lower level of absorption. The image (Figure 8a) is forward projected to get simulated projection data.

Number of the rotation angles is 35. The angles are equally distributed on the range from 0 to 180 degrees. MATLAB function *radon* was used to perform forward projection. To model the effect of intense absorption we introduced the following condition: if the simulated projection value is greater than the threshold, the threshold value is attributed:

$$P^j = \begin{cases} \sum_i \mu_i \omega_{ij}, if \ \sum_i \mu_i \omega_{ij} < Threshold; \\ Threshold. \end{cases} \quad (2)$$

Here j is an index of the ray-sum, i is an index of pixel of the image that is being reconstructed. The threshold value is 800. Pixel size is 1.

Image reconstructed with MATLAB *iradon* function with default parameters is presented on Figure 8b. *iradon* uses the filtered back-projection algorithm to perform the inverse Radon transform.

Image reconstructed with quadratic minimization technique (3) is presented in Figure 8c. We solved the optimization problem, which can be defined by the constraints:

$$\sum_i \mu_i \, \omega_{ij} \ = P^j, j = 1, \dots, 35 * 10 \quad (3)$$

with an objective function to minimize:

$$\|P^{measure} - \ P^k\|^2 \rightarrow min_\mu.$$

Here $P^{measure}$ is calculated by (2), k is a number of current iteration.

To advance the quadratic programming technique we replace the linear system of the constraints equations (3) by a system of the equations and the inequalities in agreement with

$$\begin{cases} if \sum_i \mu_i \ \omega_{ij} < Threshold, then \sum_i \mu_i^k \omega_{ij} = P^j \\ \qquad else \sum_i \mu_i^k \omega_{ij} > Treshold \end{cases}$$

We obtain new optimization problem:

$$\begin{cases} \|P^{measure} - P^k\|^2 \to min_\mu, with\ constraints \\ if \sum_i \mu_i\, \omega_{ij} < Threshold, then\ \sum_i \mu_i^k\, \omega_{ij} = P^j \\ \qquad else\ \sum_i \mu_i^k\, \omega_{ij} > Treshold, \end{cases} \quad (4)$$

This optimization problem was solved by the quadratic programming technique. Figure 8d represents the result. Comparison of the results: (a) The phantom used for sonogram simulation; (b) Image reconstructed by MATLAB *iradon* with default parameters; (c) Image reconstructed by quadratic minimization without inequality restrictions; (d) Image reconstructed by quadratic programming with inequality restrictions.

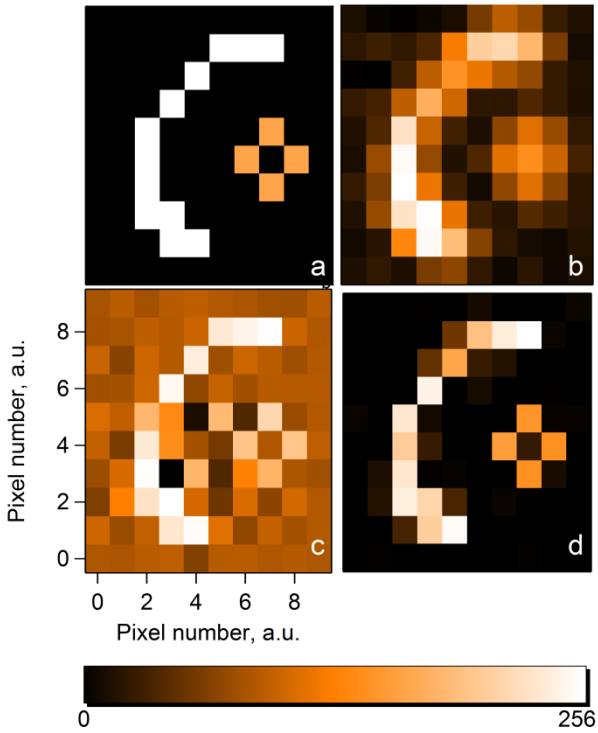

Figure 8: Comparison of reconstruction results

To compare the reconstruction quality for each image (Fig.8b, Fig. 8c, Fig. 8d) we calculate the mean value E and the variance Var for the bracket and for the cross. The boundary of a region within which a value has been calculated is exactly the same as on the phantom. The calculated values for the bracket:

$$E_{8a}=255,\ E_{8b}=146,\ E_{8c}=233,\ E_{8d}=225;$$
$$Var_{8a}=0,\ Var_{8b}=25,\ Var_{8c}=30,\ Var_{8d}=26.$$

The calculated values for the cross:

$$E_{8a}=164,\ E_{8b}=69,\ E_{8c}=170,\ E_{8d}=151;$$
$$Var_{8a}=0,\ Var_{8b}=3.6,\ Var_{8c}=19.5,\ Var_{8d}=5.2.$$

It is easy to see that the border areas on the last image less blurred than in other. If we sum all the facts given above, we get the following output. The use of quadratic programming technique allows to keep the sharp boundaries, without increasing the variation on homogeneous areas.

CONCLUSIONS

We have shown that the application of the quadratic programming technique can significantly reduce artifacts caused by presence of the areas intensively absorbing or totally blocking the X-ray beams. Presence of such regions is fairly common situation, for example, in implantology. The presence of artifacts in this case is critical. They change the true picture of implants dynamics.

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REFERENCES

Amirkhanov, A., M. Reiter, J. Kastner, M.E. Gröller, Ch. Heinzl. 2012. "Evaluation of Projection-Based Metal-Artifact Reduction for Multi-Material Components." *Conference on Industrial Computed Tomography*, Wels, Austria, 279-287.

Bamberg, F., A. Dierks, K. Nikolaou, M. Reiser, C. Becker, T. Johnson. 2011. "Metal artifact reduction by dual energy computed tomography using monoenergetic extrapolation." *Eur. Radiol*., No.21(7)(Jul), 1424-1429.

Barrett, J. and N. Keat. 2004. "Artifacts in CT: Recognition and Avoidance" *RadioGraphics*, vol. 24, No.6, 1679-1691.

Boas F.E. and D. Fleischmann. 2012. "CT artifacts: Causes and reduction techniques." Imaging Med. nu. 4(2), 229-240.

Buzug, Th. 2008. "From Photon Statistics to Modern Cone-Beam CT." *Springer*.

Martorelli, M. 2011. A Novel Method of Removing Artifacts in Dental CT Images. " Proceedings of *the IMProVe 2011 International Conference on Innovative Methods in Product Design,* Venice, Italy, 685-690.

Müller, J. 2006. "Metal artifact reduction for X-ray computed tomography in combination with iterative reconstruction algorithms". Thesis, Rhein AhrCampus Remagen.

Oehler M, Buzug T.M. 2007. "Statistical image reconstruction for inconsistent CT projection data." *J Methods Inform. Med.,* vol. 3, 261.

Prun, V.E., A.V Buzmakov, D.P. Nikolaev, M.V. Chukalina, V.E. Asadchikov. 2013. "A computationally efficient version of the algebraic method for computer tomography." *Automation and Remote Control*, No.74(10), 1670–1678.

Titarenko, S., V. Titarenko, A. Kyrieleis, Ph.J. Withers. 2010. "A priori information in a regularized sinogram-based method for removing ring artefacts in tomography." *J Synchrotron Radiat.,* No.17(4), 540-549.

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