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Material Decomposition in Spectral CT using deep learning: A Sim2Real transfer approach

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Abstract

The state-of-the-art for solving the nonlinear material decomposition problem in spectral computed tomography is based on variational methods, but these are computationally slow and critically depend on the particular choice of the regularization functional. Convolutional neural networks have been proposed for addressing these issues. However, learning algorithms require large amounts of experimental data sets. We propose a deep learning strategy for solving the material decomposition problem based on a U-Net architecture and a Sim2Real transfer learning approach where the knowledge that we learn from synthetic data is transferred to a real-world scenario. In order for this approach to work, synthetic data must be realistic and representative of the experimental data. For this purpose, numerical phantoms are generated from human CT volumes of the KiTS19 Challenge dataset, segmented into specific materials (soft tissue and bone). These volumes are projected into sinogram space in order to simulate photon counting data, taking into account the energy response of the scanner. The network is trained to decompose the materials in the projection domain after which we apply any conventional tomographic method to reconstruct the different material volumes. The proposed decomposition method is compared to a regularized Gauss-Newton (RGN) method on synthetic data, experimental phantom data and human thorax data.

1 Introduction

The new generation of spectral computed tomography (SPCCT) scanners include photon-counting detectors (PCDs), which count single photons and resolve their energy [1]. With this extra dimension, SPCCT provides higher contrast with respect to conventional CT and allows for material decomposition, which opens up new diagnosis possibilities [2], [3]. PCDs can be exploited for material decomposition, including K-edge imaging, which uses the discontinuity at diagnostic energies of the linear attenuation coefficient of high-Z element-based contrast agents such as gadolinium, gold or bismuth. Material decomposition could be used to quantify calcium content to assess bone and teeth health, determine kidney stone composition, determine liver iron concentrations, determine bone marrow composition, characterize plaques [3], [4], abdominal imaging [5], evaluating the risk of breast cancer [6], and k-edge contrast agent quantification, among others.

A variety of approaches for material reconstruction have been investigated, which can be divided into projection-based [7], [8], image-based [9], [10], and one-step inversion approaches [11], [12]. Image-based methods reconstruct first each energy bin and then perform material decomposition on the image domain assuming that images are monochromatic, which may lead to beam hardening artefacts. Projection-based and one-step inversion methods are a natural choice as they take into account the inherent physics. One-step methods are mathematically the most elegant, as they solve the problem in one step, but they can be computationally expensive. Projection-based methods firstly decompose the energy-resolved sinograms into material-specific sinograms and then perform tomographic reconstruction of each of the decomposed sinograms [13]. An advantage of this approach is that it allows independent decomposition of each projection view, which can be parallelized, making the computational process more convenient [13].

Material decomposition in spectral CT is a nonlinear, nonconvex inverse problem [13], [14]. State-of-the art for solving this problem are variational approaches [13, 15–17] or iterative regularization schemes [14]. In [13], the authors proposed a regularized material-specific Gauss-Newton (RGN) method, which outperformed widely used unregularized maximum likelihood method. In [14], authors assessed an iterative regularization scheme based on the Bregman iteration. While variational methods are robust and have shown improved results with respect to unregularized approaches, they present several disadvantages. First, they can be computationally slow, given their iterative nature. Second, their performances rely on the prior knowledge of the scanner energy response and the choice of the regularization functional. With regard to the adopted model, for instance, the detector response can be degraded by charge sharing, pulse pileup, and energy loss due to K-escape [18]. The effect of assuming a perturbed detector response can have detrimental effects on image quality, as shown in [19]. With regard to the choice of the regularization functional, specific regularization for the decomposed materials have shown to improve image quality with respect to unregularized approaches [13], but the choice is far from being optimal.

Recently, deep convolutional neural networks (CNN) have shown outstanding results in several image processing tasks [20], [21]. CNNs have been also proposed for solving several inverse problems, including image restoration and image reconstruction [22–27]. Among the diverse CNN configurations, U-Net has shown outstanding results for image segmentation [28] and has been applied to postprocessing in CT [22] and to artifact removal of nonlinear CT images [29]. Several works have also proposed deep learning for solving the material decomposition, mainly in the image domain, using U-Net in [30] and VGG-16 in [31]. The previously proposed neural networks approaches in the projection domain have been based on multilayer perceptrons (using fully connected layers) for decomposing materials in a pixel-by-pixel basis [32], [33]. In [32], the authors used a neural network with two hidden layers followed by a denoising method to mitigate noise.

The recent success of CNNs for image processing tasks relies on large-scale annotated data sets such as ImageNet with millions of images [34], [20]. However, equivalent medical imaging data sets are not common. Transfer learning (TL) has emerged as a promising candidate technique to compensate the lack of large-scale medical imaging data sets. TL is the ability of a system or model to recognize and apply knowledge and skills learned in previous domains/tasks to novel domains/tasks. TL assumes that the model can generalize to new tasks or new domains by learning features or factors that are common between source and target tasks [35]. TL becomes relevant when labelled training data, time or computational power are limited, so it is particularly applicable to medical imaging. Different approaches for TL can be found in [36], [37]; most common is to pre-train a model on a large-annotated data set and then to fine tune the model on a more specific and smaller data set [38], [39]. Recent works proposed related TL approaches based on fine tuning for material decomposition [19, 33]. However, in this work we follow a Sim2Real transfer approach where the knowledge that one learns from synthetic data is transferred to a real-world scenario [40], [41]. The Sim2Real strategy is based on the fact that synthetic data is almost infinitely accessible and diverse, and it becomes pertinent when experimental annotated data are very limited or non existent, which is the case of SPCCT. While learning-based approaches perform reasonably well given experimental training data, they may not necessarily work well when trained only on synthetic data. In cases in which synthetic data generalize poorly to real data, data augmentation strategies are generally followed.

In this work, we propose a deep learning strategy for solving the projection-based material decomposition problem based on a U-Net architecture and a Sim2Real transfer approach. In order for the Sim2Real approach to work, synthetic data must be realistic and representative of the experimental data. For this purpose, numerical phantoms are generated from human CT volumes of the KiTS19 Challenge dataset [42]. Numerical data are then computed by segmenting CT volumes into specific materials (soft tissue and bone), projecting the material volumes and simulating photon counting data, taking into account the energy response of the scanner. The U-Net is trained to decompose the materials in the projection domain and is compared to a regularized Gauss-Newton (RGN) method [13] using both numerical and experimental data. After material decomposition of the photon counting data, we apply a conventional tomographic method to reconstruct the different material volumes. The proposed approach combines a CNN model with training on human data, which allows to implicitly learn the real prior distribution of the decomposed materials [43]. This is particularly interesting from a Bayesian perspective to inverse problems as it motivates the choice of the regularization functional as a particular selection for the prior distribution of the unknown variable. Thus, our method differs from previous CNN methods based on the image domain and approaches that are trained on experimental phantom data only.

2 Methods

2.1 Forward model

We assume a 2D sensor with P pixels and I energy bins and an object with V voxels made of M materials. We image the object under Θ projections. Let $\mathbf{s} = (s_{1,1}^1, \dots, s_{i,p}^\theta, \dots, s_{I,P}^\Theta)^\top$ be the measurement vector, where $s_{i,p}^\theta$ represents the photon counts measured in the i -th energy bin at the p -th pixel under the θ -th projection, and $\boldsymbol{\rho} = (\rho_{1,1}, \dots, \rho_{m,v}, \dots, \rho_{M,V})^\top$ be the (unknown) mass densities vector, where $\rho_{m,v}$ is the mass density for the m -th material at the v -th voxel. Solving the spectral CT problem in one step requires the inversion of the nonlinear system

$$\mathbf{s} = \mathcal{G}(\boldsymbol{\rho}) \quad (1)$$

where \mathcal{G} represents the forward model that maps $\boldsymbol{\rho}$ onto \mathbf{s} .

An alternative to solving the problem (1) in one step is the two-step approach [13, 15]. In this case, the forward model \mathcal{G} can be seen as the composition of the linear X-ray transform and a non-linear spectral mixing operator. The X-ray transform \mathcal{X} applies to each material independently, i.e.,

$$\mathbf{a}_m = \mathcal{X}(\boldsymbol{\rho}_m), \quad 1 \leq m \leq M \quad (2)$$

where $\boldsymbol{\rho}_m = (\rho_{m,1}, \dots, \rho_{m,v}, \dots, \rho_{m,V})^\top$ and $\mathbf{a}_m = (a_{m,1}^1, \dots, a_{m,p}^\theta, \dots, a_{m,P}^\Theta)^\top$ represent mass density and projected mass density for the m -th material, respectively. Spectral mixing applies to each view and detector pixel independently, i.e.,

$$\mathbf{s}^\theta = \mathcal{F}(\mathbf{a}^\theta), \quad 1 \leq \theta \leq \Theta \quad (3)$$

where $\mathbf{a}^\theta = (a_{1,1}^\theta, \dots, a_{m,p}^\theta, \dots, a_{M,P}^\theta)^\top$ and $\mathbf{s}^\theta = (s_{1,1}^\theta, \dots, s_{i,p}^\theta, \dots, s_{I,P}^\theta)^\top$. In particular, we consider the following non linear mixing [15]

$$s_{i,p}^\theta = \int_{E \in \mathcal{E}} n_{0,p}(E) d_{i,p}(E) \exp \left[- \sum_{m=1}^M a_{m,p}^\theta \tau_m(E) \right] dE \quad (4)$$

where \mathcal{E} is the range of energy delivered by the x-ray tube, $n_{0,p}(E)$ is the x-ray source energy spectrum at the p -th pixel, $d_{i,p}(E)$ is the energy-dependent detector response function at the p -th pixel for the i -th bin, and $\tau_m(E)$ is the energy-dependent mass attenuation coefficient of the m -th material.

2.2 Spectral CT image reconstruction

Solving the inverse spectral CT image reconstruction problem in one step involves inverting the $\mathbb{R}^{MV} \rightarrow \mathbb{R}^{I P \Theta}$ nonlinear mapping (1). In this work, we instead adopt the two-step approach, which splits the problem into Θ inversions of $\mathbb{R}^{MP} \rightarrow \mathbb{R}^{IP}$ mappings (3) followed by M inversions of $\mathbb{R}^V \rightarrow \mathbb{R}^{P \Theta}$ (2). The two-step approach reduces the size of the subproblems, offering a natural parallelization scheme (e.g. decomposing all angles in parallel, reconstructing all materials in parallel), and allows to separate nonlinear (i.e. spectral) from linear (i.e. tomographic) mixing. Thus, the two-step approach allows to reduce the computational complexity with respect to the one-step method by first solving the material decomposition problem (3), projection by projection, and then solving the tomographic reconstruction problem for each material (2).

In this work, we consider the material decomposition problem with two different approaches: a regularized variational approach and a deep learning strategy, as described in the next two subsections. Material decomposition is done projection by projection, which allows to include an explicit regularization functional or to implicitly learn the prior distribution. A previous work found that a regularization approach led to significant improvement with respect to a unregularized pixel-by-pixel decomposition, which was very sensitive to noise [13]. An alternative to solving the material decomposition problem projection by projection would be to decompose all projection views at once. While this could allow to introduce further regularization on the angle dimension, we chose single projection decomposition as it allows for parallelization across angles, reducing computational complexity.

2.2.1 Material decomposition with a variational method

Using a variational framework, material decomposition has been previously formulated as the minimization of the following cost function [13]:

$$\Phi(\mathbf{a}^\theta) = \frac{1}{2} \|\mathbf{s}^\theta - \mathcal{F}(\mathbf{a}^\theta)\|_{W^\theta}^2 + \alpha \sum_m \mathcal{R}_m(\mathbf{a}^\theta), \quad 1 \leq \theta \leq \Theta \quad (5)$$

where the data fidelity term is a weighted least square functional to approximate a Poisson distribution, $W^\theta = \text{diag}(1/\sqrt{\mathbf{s}^\theta})$ is a weighting matrix, α is a regularization parameter, and \mathcal{R}_m accounts for specific material regularization. We chose second-order and first-order Tikhonov regularization for soft tissue and bone, respectively, as suggested in [13]. While two different values of the regularization parameter could be used for different materials, it was previously found to not lead to significant improvement [13]. Cost function (5) is efficiently minimized by using a Gauss Newton algorithm (RGN); MATLAB code for RGN is available from the Spectral X-ray image reconstruction SPRAY toolbox [44].

2.2.2 Material decomposition with the proposed deep learning method

Deep learning has been recently proposed for inverse problems [22, 45]. In this work, we want to learn the mapping

$$\mathcal{H}_\beta : \tilde{\mathbf{s}}^\theta \mapsto \tilde{\mathbf{a}}^\theta, \quad (6)$$

where β indicates the parameters of the neural network, $\tilde{\mathbf{s}} = \ln(\hat{\mathbf{s}}/\mathbf{s}^\theta)$ corresponds to the normalized data with $\hat{\mathbf{s}} = \mathcal{F}(0)$ representing the measurement in the absence of the object, and $\tilde{\mathbf{a}}$ corresponds to the normalized outputs. We normalize the output material per material across the entire training set, i.e., $\tilde{a}_{m,p}^{\theta,n} = a_{m,p}^{\theta,n} / \max_{p,\theta,n} a_{m,p}^{\theta,n}$.

Normalization of input and output variables prior to training a neural network model is common practice to avoid a slow or unstable learning process. In this case, normalization of the outputs is motivated by the fact that different materials may have different range of values. For the inputs, photon-counting data have very large values; in addition; normalizing the data by the acquisition in the absence of the object, $\hat{\mathbf{s}}$, may decrease the sensitivity to modeling errors.

Learning means finding the parameters β of the network that minimizes the following loss function

$$\Xi(\beta) = \sum_{n=1}^N \|\mathcal{H}_\beta(\tilde{\mathbf{s}}^n) - \tilde{\mathbf{a}}^n\|^2 = \sum_{n=1}^N \sum_{m=1}^M \|\mathcal{H}_{\beta,m}(\tilde{\mathbf{s}}^n) - \tilde{a}_m^n\|^2, \quad (7)$$

where $(\tilde{\mathbf{s}}^n, \tilde{\mathbf{a}}^n)$ are N input-output vector pairs (3) that can be reshaped into input-output array pairs of size $(P_x \times P_y \times I, P_x \times P_y \times M)$, N is the number of projection images in the training set, and each projection image is of size $P_x \times P_y$. Minimization of (7) was done with Adam method under TensorFlow, with learning rate 10^{-5} and batch size of 16. Training and test losses were computed during training and early stopping was adopted to avoid overfitting.

We use a U-Net architecture [28], which consists in a contracting multi-scale decomposition path and a symmetric expanding path with skip connections at each scale. The contracting path comprises 3×3 convolutions ('same' padding convolutions) each followed by a rectified linear unit (ReLU) and alternated with a 2×2 max pooling operation with stride 2 for downsampling by half every two convolution operations. At each downsampling the number of feature channels is doubled, with 32 channels in the first scale and 128 channels in the last scale. The expansive path comprises the same convolution operations as in the contracting path but alternated with upsampling by two every two convolution operations. The final layer is a 1×1 convolution used to map the 32 channel layer to the output with the desired M number of decomposed materials. This leads to a total of 14 hidden layers (figure 1).

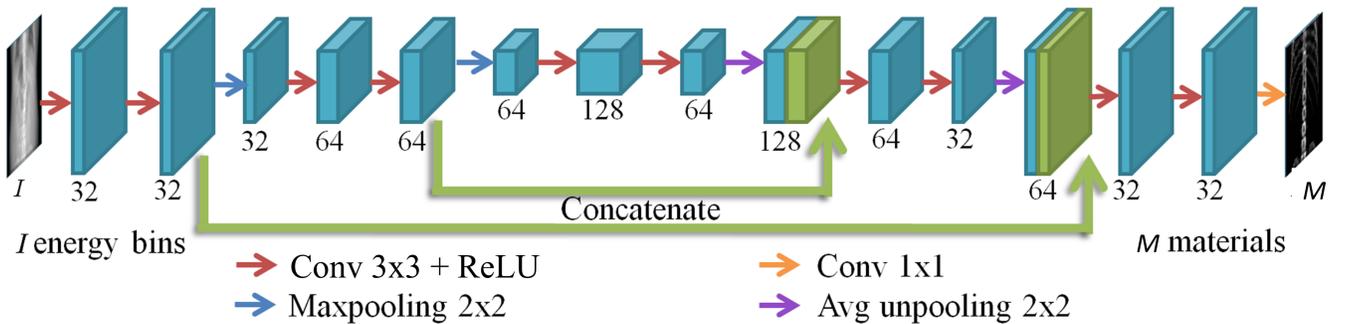


Figure 1: U-Net architecture for material decomposition. Input is a stack of photon counting projection images for I energy bins $(P_x \times P_y \times I)$, and output is a stack of the decomposed material projection images for M materials $(P_x \times P_y \times M)$ for a projection angle θ .

2.2.3 Tomographic reconstruction

Tomographic reconstruction solves the problem (2). For numerical data, we assume parallel geometry and perform tomographic reconstruction using filtered back-projection algorithm. For experimental data, tomographic reconstruction is done using a regularized conjugate gradient method within the Reconstruction Toolkit (RTK) [46], which minimizes the following cost function:

$$\Psi(\boldsymbol{\rho}_m) = \|\mathcal{X}\boldsymbol{\rho}_m - \mathbf{a}_m\|_2^2 + \gamma\|\nabla\boldsymbol{\rho}_m\|_2^2, \quad (8)$$

where γ is a regularization parameter. The forward projection uses ray casting with trilinear interpolation and the backprojection is voxel-based with bilinear interpolation. Parameters were selected empirically as $\gamma = 100$ and $k = 25$, where k is the number iterations of conjugate gradient.

2.3 Data and image evaluation

2.3.1 Numerical human phantoms

Numerical human phantoms are made of soft tissue and bone and are built from CT volumes obtained from the KiTS19 challenge data set (2019 Kidney Tumor Segmentation Challenge, <https://kits19.grand-challenge.org/data/>) [42] (figure 2). We used 50 phantoms for training, 3 for evaluation, and 9 for test. CT volumes have voxel size $0.96 \text{ mm} \times 0.96 \text{ mm} \times 1.8 \text{ mm}$. In order to facilitate numerical simulations, we processed CT volumes to get the same dimension: volumes were cropped to have 640 voxels in x - and y -dimensions and 100 voxels along z -axis. CT volumes were semi-automatically segmented into bone and soft-tissue using a combination of thresholding and morphological operations. After segmentation, we created mass density volumes by normalizing the segmented volumes to have median value in the kidneys equal to $1.1 \text{ g}\cdot\text{cm}^{-3}$. Then, mass densities for both materials were projected using the Radon transform (assuming parallel geometry) to create the projected mass densities.



Figure 2: Two examples from the CT Kits19 data base.

Photon counting data were simulated taking into account the manufacturer source spectrum and detector response function and the linear attenuation coefficients (LACs) of the two materials. Both phantom creation and simulation were realized with the SPRAY toolbox [13, 44]. Noisy data was considered by assuming a Poisson distribution for a tube current of 200 mA, which corresponds to a total number of photons in a central pixel of the detector equal to $N_0 = 6 \cdot 10^5$. Projected data consisted of 360 projections over a 180 angle span and 924×8 detector pixels.

Training data for U-Net then consisted of 18000 projections (360 projections for 50 phantoms). In order to perform further data augmentation, projected volumes were randomly cropped along the z -dimension and data simulation and noise corruption were performed at runtime during training.

2.3.2 Experimental data

A SPCCT prototype system (Philips Medical Systems) installed at CERMEP, Lyon, was used for this study [47]. It is a 500 mm FOV system equipped with a conventional x-ray tube that can be set with a tube voltage at 80, 100, or 120 kVp and tube current values between 10 and 500 mA. Tube filtration absorbs low-energy x-rays so the subject is exposed to a spectrum that ranges from 35 to 120 keV. The system is based on photon counting detectors of 2-mm-thick cadmium zinc telluride with a pixel pitch of $270 \mu\text{m} \times 270 \mu\text{m}$ at isocenter, ChromAIX2 application-specific integrated circuits combined with cadmium zinc telluride as sensor material, and operates in single photon-counting mode with energy discrimination. Photo-counting detectors allow up to 5 consecutive energy

thresholds between 30 and 120 keV. The scan field of view is 500 mm in-plane, with a z-coverage of 17.5 mm in the scanner isocenter. Axial and helical scans over 360° are performed with 2400 projections per rotation.

We consider two data sets: an experimental phantom and human thorax data. For the experimental phantom, we used the CIRS (Computerized Imaging Reference Systems, Inc, USA) Electron Density Phantom [48], which is made of a cylinder of size $330 \text{ mm} \times 270 \text{ mm} \times 50 \text{ mm}$ and a density of $1.029 \text{ g}\cdot\text{cm}^{-3}$ and inserts as surrogates of different biological tissues. We used a mixture of inserts from the CIRS phantom and the GAMMEX Tissue Characterization Phantom Model 467 (Gammex Middleton, USA). Data were acquired using a tube voltage of 120 kVp and tube current of 198 mA using axial geometry, with a standard set of energy thresholds (30, 51, 62, 72, 81 keV). Data were binned to a final size of $462 \times 16 \times 2400 \times 5$. Final reconstructed voxel size was $1 \text{ mm} \times 1 \text{ mm} \times 2 \text{ mm}$. In order to compare the data SNR with previous work [19], we compute the total number of photons provided by the source in a central pixel of the detector. For this data set, the total number of photons was approximately $N_0 = 6 \cdot 10^5$.

The human thorax data set corresponds to an adult male volunteer. Data were acquired using a tube voltage of 120 kVp and exposure of 62 mAs using helical geometry (85 mA, 0.75 s rotation time and 1.03 pitch). Data were binned to a final size of $924 \times 8 \times 24912 \times 5$. This corresponded to a total number of photons of $N_0 = 1.8 \cdot 10^5$.

We remark that data calibrations and corrections included in the scanner manufacturer pipeline have not been included in this work.

2.3.3 Image evaluation

U-Net and RGN were evaluated on synthetic data and on the two experimental data sets. The performance of RGN depends on the selection of the regularization parameter. Selecting the optimal regularization parameter in terms of MSE may lead to images with low noise but with loss of image details [14]. Thus, we provide results for RGN for two values of the regularization parameter: $\alpha = 0.1$ (low regularization) and $\alpha = 0.6$ (intermediate regularization). For the numerical test set, methods were assessed in terms of normalized Mean Squared Error (MSE) and Structural Similarity metric (SSIM) on both decomposed projections and reconstructed images. Normalized MSE was computed for each material as follows:

$$\text{MSE} = \frac{\|\mathbf{a}_m - \mathbf{a}_m^{\text{Target}}\|_2}{\|\mathbf{a}_m^{\text{Target}}\|_2}. \quad (9)$$

In addition, methods were assessed by visual inspection. For the experimental phantom data, methods were assessed in terms of quantification of soft tissue and bone inserts by computing the mean in a circular region with radius of 1 cm for a central slice. Both experimental phantom data and human thorax data were evaluated visually. For experimental data, we also computed a 'virtual' monochromatic image (for 60 keV) from the decomposed images in order to compare visually to the standard CT image provided by a conventional scanner. We chose 60 keV as a previous study found that monochromatic images between 60 and 70 keV could be considered as optimal surrogates of conventional CT images [49]. Monochromatic images were computed as the linear attenuation coefficient at energy E and point \mathbf{x} in the reconstructed volume:

$$\mu(E, \mathbf{x}) = \sum_{m=1}^M \rho_m(\mathbf{x})\tau_m(E). \quad (10)$$

3 Results

3.1 Numerical thorax phantom data

Figure 3 shows decomposed images by RGN for two different values of α and by U-Net. For $\alpha = 0.1$, which corresponds to little regularization, RGN provides accurate decomposition, recovering image details, but images are noisy. For $\alpha = 0.6$, RGN reduces noise but at the expense of excessive blurring, particularly in bone regions. This is well appreciated in the error images in figure 4. On the contrary, U-Net is able to reduce noise while maintaining image details.

Tomographic reconstructions of decomposed data with RGN and U-Net are shown in figure 5. These results are consistent with results obtained for the decomposed projections. For $\alpha = 0.1$, RGN recovers image details well but images are noisy. For $\alpha = 0.6$, RGN is able to remove noise but image details are lost. U-Net leads to similar reduction of noise than RGN for $\alpha = 0.6$ while maintaining image quality, as it is well appreciated in the error images in figure 6.

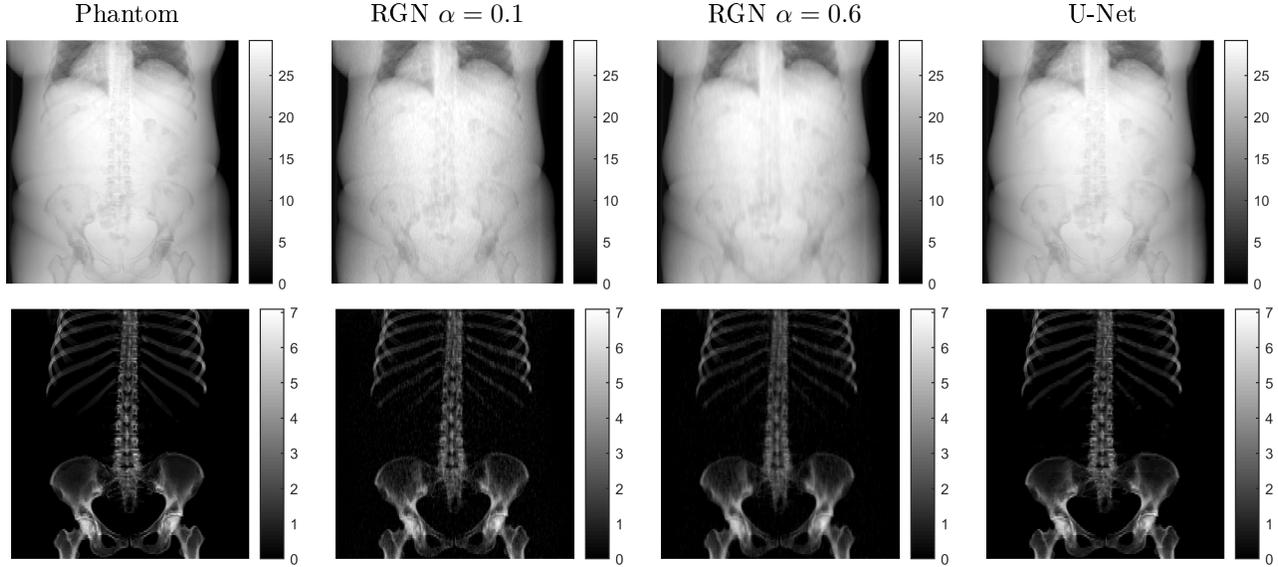


Figure 3: Phantom and decomposed projections (for 0° view) for RGN and U-Net for a synthetic phantom from the test set. Top: soft tissue. Bottom: Bone.

Figure 7 shows image details of the reconstructed images for soft tissue. RGN for $\alpha = 0.6$ misses image details while U-Net presents better recovery than RGN for $\alpha = 0.1$. While the image quality of U-Net is generally superior, it presents some 'fake' structures in the soft tissue image that seem to originate from noisy data (shown by green arrows in figure 7). Artefacts also appear in the RGN image for the larger regularization parameter but they are slightly more apparent for U-Net because of the overall reduction of noise. In addition to these artefacts, RGN present large errors in some bone locations.

Figure 8 displays normalized MSE and SSIM for decomposed projections and tomographic reconstructions for U-Net and RGN. For soft tissue, U-Net presents a 29 % and 6 % decrease in normalized MSE on decomposed projections with respect to RGN for $\alpha = 0.1$ and $\alpha = 0.6$, respectively. Normalized MSE for bone is larger than for soft tissue due to noise, as most part of the image corresponds to background. U-Net led to similar decrease in MSE for bone as for soft tissue. U-Net also led to increase in SSIM with respect to RGN for bone and soft tissue. For bone tissue, U-Net presents a 2.5- and 3-fold increase in SSIM on decomposed projections with respect to RGN for $\alpha = 0.1$ and $\alpha = 0.6$, respectively. For tomographic reconstructions, U-Net leads to similar improvements in MSE and SSIM.

3.2 Experimental data

3.2.1 Experimental phantom data

Figure 9 shows decomposed sinograms and figure 10 shows tomographic reconstructions, for the experimental phantom data. Methods lead to similar quantitative results, especially for soft tissue inserts. Decomposition of bone materials are slightly different. In terms of image quality, U-Net presents less noisy images and slightly less artefacts than RGN. The 60 keV images provided by the two methods are very similar, with U-Net presenting less noise.

Figure 11 shows quantification of the phantom inserts using the ROIs shown in figure 10. Methods provide similar quantification overall but there is a drift between RGN and U-Net for the recovery of bone like inserts. Figure 11(left) shows the quantification of all inserts from the decomposed soft image. Both methods show accurate quantification (Pearson's correlation coefficient $r > 0.98$) for soft tissue-like inserts (displayed with symbol +). We remark that bone-like inserts (displayed with symbol \times) are not close to zero, as they are comprised of both soft tissue and bone.

Figure 11(center) shows the quantification of all inserts from the decomposed bone image. Both methods show accurate quantification ($r > 0.98$) for soft bone-like inserts (\times). In this case, bone material is around zero for soft tissue-like inserts (+).

Figure 11(right) shows the quantification of all inserts from the 60 keV image. RGN and U-Net show almost

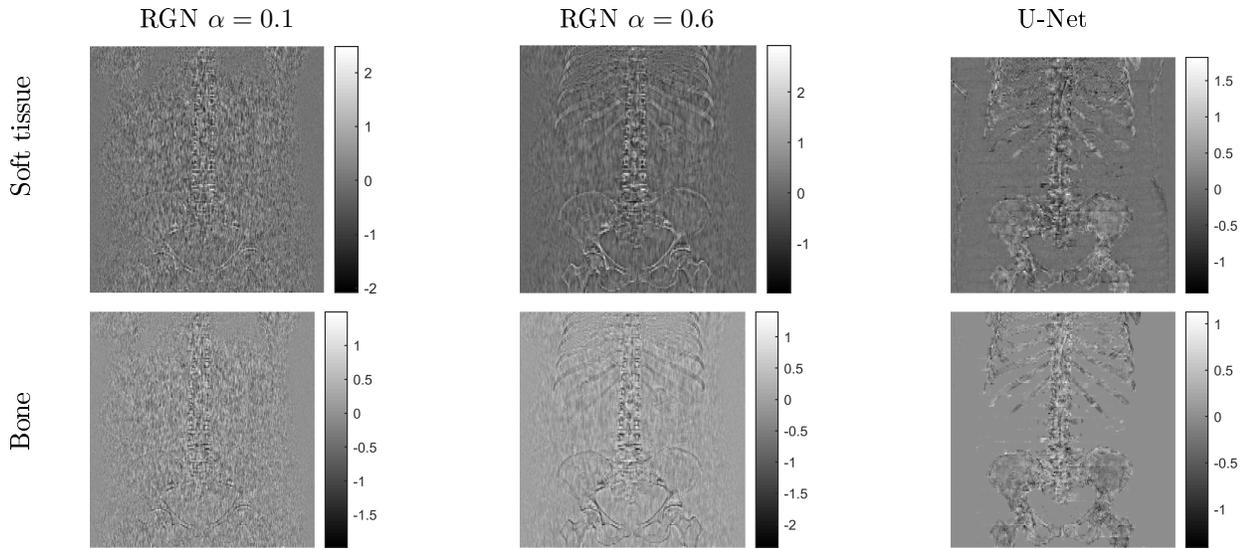


Figure 4: Error images of decomposed projections by RGN and U-net for the same view as in figure 3.

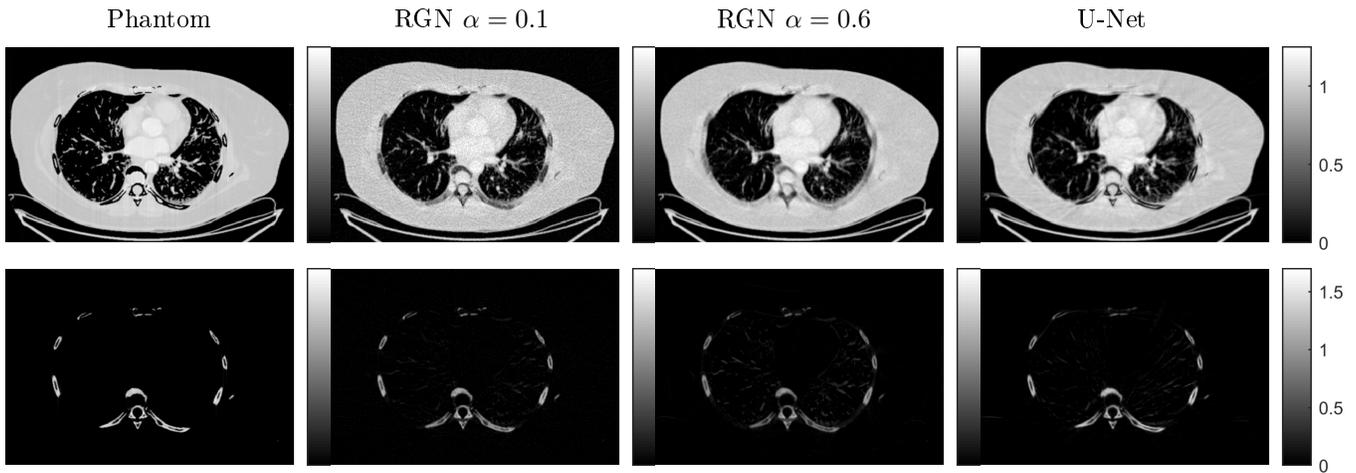


Figure 5: Tomographic reconstruction of a numerical human phantom for RGN and U-Net. Image reconstruction corresponding to decomposed projections shown in figure 3. Top: soft tissue. Bottom: Bone.

identical results.

3.2.2 Human thorax data

Figure 12 shows tomographic reconstructions and the 60 keV image for RGN and U-Net. RGN provides noisier decomposed images and appears more sensitive to ring artefacts than U-Net. In terms of decomposed materials, U-Net seems to have more bone material in the soft tissue image than RGN. Compared to the decomposed images, 60 keV images do not show ring artefacts. Zoomed details show that ring artefacts are more apparent for RGN than for U-Net. U-Net also appears to provide slightly finer recovery of bone details.

3.3 Computation time

We run our code on a Windows computer with a 64-bit operating system, Intel Xeon(R) E5-1650 v4 3.60 GHz CPU and 128 GB RAM. The U-Net is trained on a GeForce NVIDIA GTX 1080 Ti graphics card, which requires 6.5 days for 2000 epochs. Decomposition and tomographic reconstruction times for synthetic data and the human thorax data set are displayed in Table 1. Material decomposition with RGN is done on CPU by using straight-

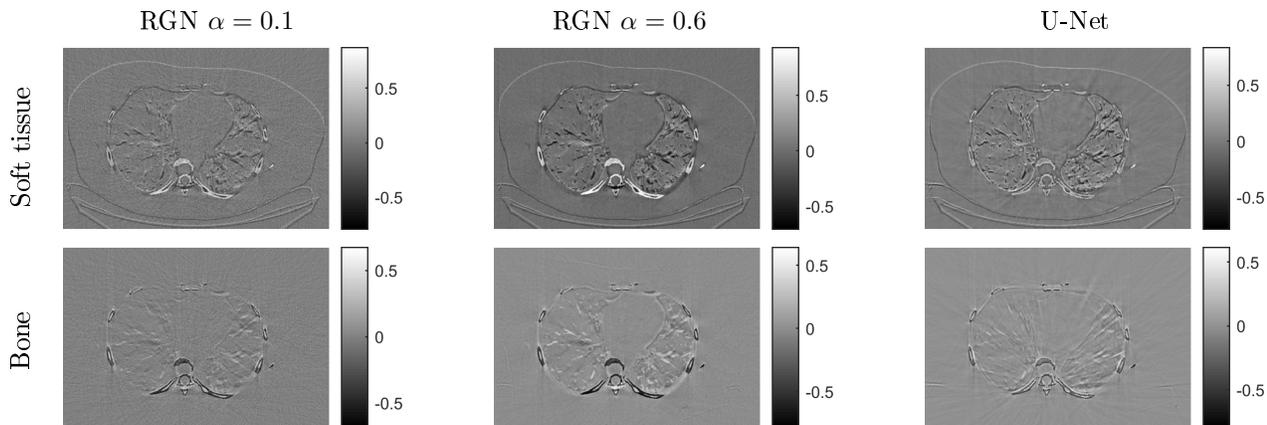


Figure 6: Error images of tomographic reconstructions of a numerical human phantom (corresponds to figure 5). Top: soft tissue. Bottom: Bone.

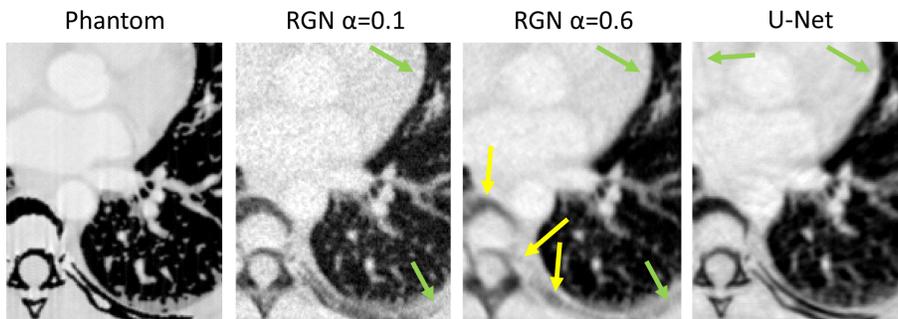


Figure 7: Detail of tomographic reconstruction of soft tissue for the images shown in figure 5. Yellow arrow points where bone details are not well recovered. Green arrow points to the appearance of 'fake' details.

forward parallelization on 4 cores. For U-Net, we provide decomposition times on CPU and GPU. Tomographic reconstruction is done on CPU for synthetic data and on GPU with RTK for the human thorax data set.

4 Discussion

We have proposed a deep learning approach based on a U-Net architecture for solving the material decomposition problem in spectral CT and compared it to a RGN method. Given that medical imaging data and, specifically, spectral CT data are scarce, we have followed a Sim2Real approach by training on simulated human phantoms and assessing on experimental phantom data and human data. We found that U-Net leads to improved image quality with respect to RGN, which presented a strong trade off between noise and regularization. Increasing the regularization parameter for RGN decreased noise but it led to loss of image details. On the contrary, U-Net was able to remove noise while maintaining image quality. In terms of quantification of decomposed materials, both methods led to similar results on the assessed experimental phantom data. Furthermore, U-Net led to a significant reduction in prediction time, specially for large data sets.

The superiority in terms of image quality of U-Net with respect to RGN must be discussed in detail. This can be explained by the fact that U-Net can learn implicitly the probability distribution of the decomposed materials. In fact, Bayesian deep learning interprets supervised learning as recovering the posterior distribution and it approximates the conditional mean when using the MSE loss [43]. This is particularly relevant for material decomposition in the projection domain, as in this case the choice of prior distributions is less clear. The RGN method included material specific regularization where regularization functionals for each material were selected as suggested in [13]. We used first- and second-order Tikhonov for bone and soft tissue, respectively. While this choice seems reasonable, it is not optimal. A second disadvantage of variational methods is the high dependence on the regularization parameter, which requires fine tuning. Here, we showed results for RGN for different values

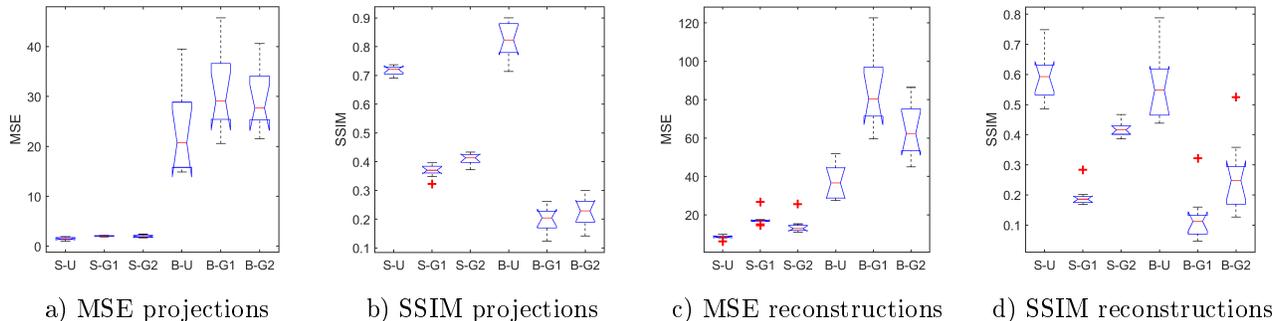


Figure 8: Quantitative results on the numerical thorax test set. Boxplots for normalized MSE and SSIM of decomposed projections and tomographic reconstructed images, given by U-Net (U), RGN for $\alpha = 0.1$ (G1) and RGN for $\alpha = 0.6$ (G2), for soft tissue (S) and bone (B). Boxplots show the median, 25-th and 75-th percentiles, and minimum and maximum values excluding outliers, which are displayed with red crosses. When notches do not overlap, one can conclude, with 95% confidence, that the true medians do differ.

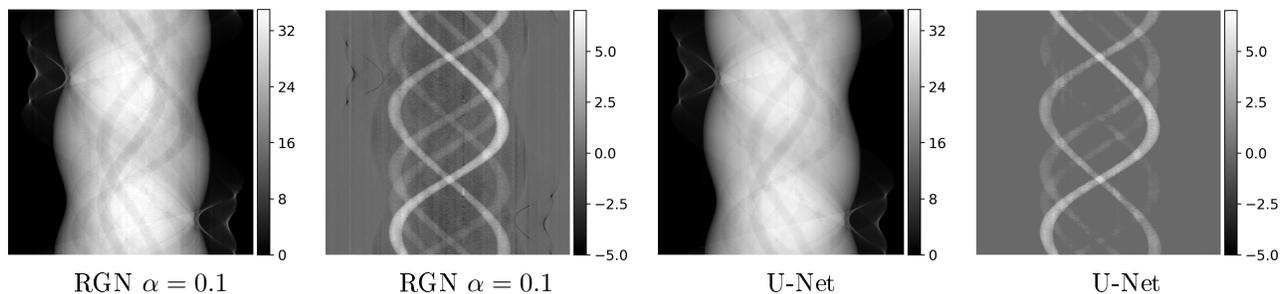


Figure 9: Material decomposed sinograms of soft tissue and bone for the experimental phantom with Gammex and Cirs inserts for RGN and U-Net.

of the regularization parameter. For further insight into the selection of the regularization parameter for RGN, we refer to [13] and [14]. Although fine tuning can generally lead to an acceptable compromise between noise and regularization for variational approaches, its benefits are mitigated by the fact that regularization functionals are not optimal.

In this work, we have proposed a learning approach for material decomposition in the projection domain which has been the preferred approach for empirical calibration methods and for model based approaches, as they are supported by the physics. Empirical calibration methods used low-order polynomials to perform material decomposition in a pixel-by-pixel basis. Similarly, model-based approaches such as Maximum-Likelihood estimation have been used for material decomposition in a pixel-by-pixel basis. However, these approaches do not exploit the information of neighbor pixels as in the regularization approaches. In [50], RGN method was shown to outperform a Maximum-Likelihood method, which was very sensitive to noise. In this work, we compare RGN to a CNN method, which implicitly regularizes the problem, as previously discussed.

Several deep learning approaches have been previously proposed for material decomposition. Most of them perform material decomposition in the image domain [30,31]. These approaches reconstruct first each energy bin and then perform material decomposition on the image domain assuming that images are monochromatic, which may lead to beam hardening artefacts. However, CNN learning approaches may correct for this. A comparison between CNN projection-based and CNN image-based decomposition methods is still an open question but it is out of the scope of this work. Recently a multilayer perceptron was proposed for solving the material decomposition problem in the projection domain [33]. In this work, we proposed a CNN approach to assess the potential of deep learning not only for solving the material decomposition problem but for implicit regularization. We have also compared deep learning with a variational approach while previous works compared their methods with more classical approaches, eg. calibration methods. In addition, we have shown that a Sim2Real approach that learns on numerical human phantoms is feasible. This is particularly interesting for a CNN method, as it can implicitly learn the optimal regularization for human data. We remark that previous approaches have learned on experimental

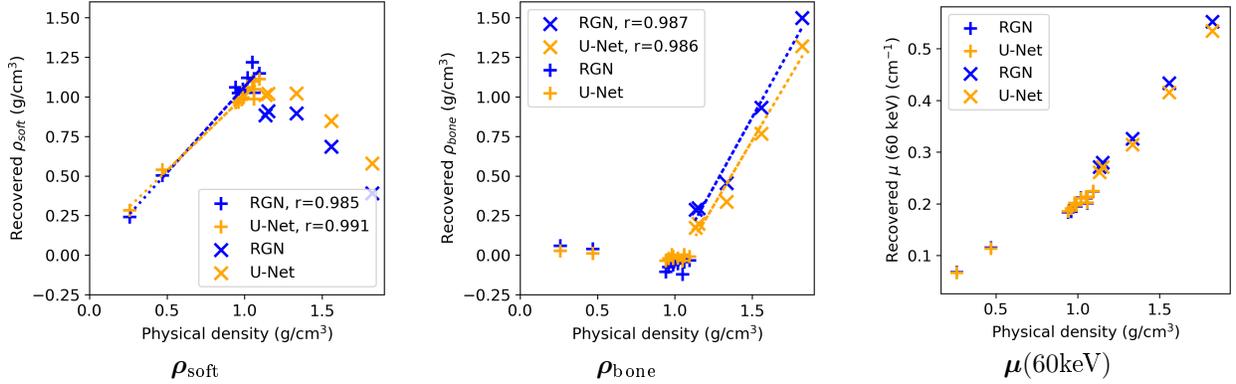


Figure 11: Quantification of (Gammex and Cirs) inserts for RGN (blue) and U-Net (orange) from decomposed soft tissue image ρ_{soft} (left), decomposed bone image ρ_{bone} (center) and 60 keV image $\mu(60\text{keV})$ (right). Inserts are marked with the symbol + for soft tissue like materials (lung, breast, muscle, solid water, adipose tissue, brain, muscle) and with the symbol \times for bone like materials (cortical bone (CB), inner bone, CB 30%, CB 50%).

Table 1: Material decomposition (Mat. Dec.) and tomographic reconstruction (Tomo. Rec.) times for synthetic data and for the human thorax data set. CPU, 4 cores: Parallelization using 4 cores. BS = Batch Size.

	Simulated data		Human thorax	
	Mat. Dec.	Tomo. Rec.	Mat. Dec.	Tomo. Rec.
RGN (CPU, 4 cores)	44 min	25 s	155 min	50 min (GPU)
U-Net (CPU)	70 s	""	25 min	""
U-Net (GPU)	6 s (BS = 16)	""	45 s (BS = 1024)	""

during training led to almost complete removal of artefacts, as expected. Nevertheless, as most of these artefacts can be considerably reduced with appropriate data calibration, we do not address this issue in this work.

This work is subject to a few limitations. Results may depend on the learning model. Here we have used a U-Net architecture, which has led to outstanding results for diverse CT applications [22, 29]. However, for this problem a simpler architecture may be sufficient. The material decomposition has a pixel-wise nature, so it is possible that simpler ConvNets can learn the prior distributions of the materials without the need of a hierarchical-based architecture, which is more designed to interpret information across the entire image. Besides the architecture, deep iterative methods, also called model-based learning methods, proposed for solving inverse problems could provide further benefits [24, 27, 53]. In particular, these iterative methods include a data consistency condition, which could mitigate the appearance of artefacts that can potentially lead to mislead diagnosis. Actually, in this work we found that both RGN with large regularization or U-Net can lead to 'fake' artefacts that may look real, given the improved quality in the surrounding regions. Although the proposed method based on U-Net leads to overall high image quality, deep iterative methods may be required to provide learning methods that are robust for the clinical use. These methods can also reduce further the need of learning power as well as the required amount of training data. Another interesting solution could be to consider Bayesian neural networks that also provide an uncertainty estimate [43]. With regard to the amount of training data, we have not evaluated the minimum requirements for this model. In addition, we have assumed that the energy response of the scanner is perfectly known. Although this has been provided by the manufacturer, some deviations from ideality conditions may occur. Further work could also fine tune a pretrained network using experimental phantom data as suggested in [19]. However, a previous work found only small improvements using a fully-connected network [33]. In this work, we have assessed the feasibility of the proposed method for material decomposition for two materials only. However, this methodology can be easily extended to three or more materials, including k-edge imaging, as shown on numerical data in [19].

In conclusion, deep learning methods show a great potential for spectral CT as they provide almost real time material decomposition and yield competitive results in comparison with current variational approaches.

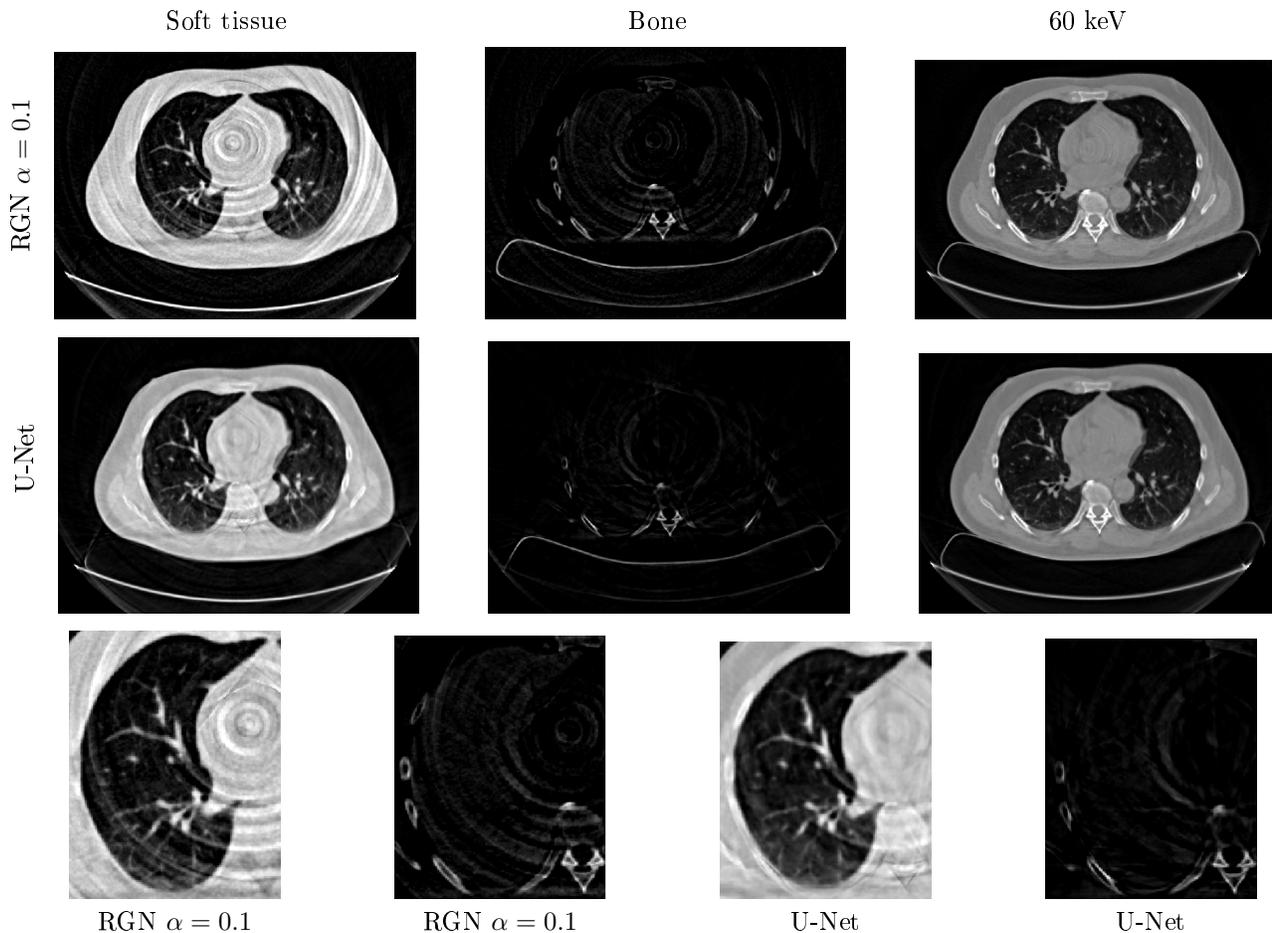


Figure 12: Top: Tomographic reconstruction of a human thorax for RGN and U-Net. Gray scales are (0, 35), (0, 15), and (0, 0.4) for soft tissue, bone and 60 keV, respectively. We remark that data calibrations and corrections included in the scanner manufacturer pipeline have not been included in this work. Bottom: zoomed detail for soft tissue and bone.

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