Synergistic Reconstruction Symposium
Chester UK, 2019

PET-enabled Dual-energy CT:

Exploring a New Way of Spectral Imaging Using Synergistic Reconstruction

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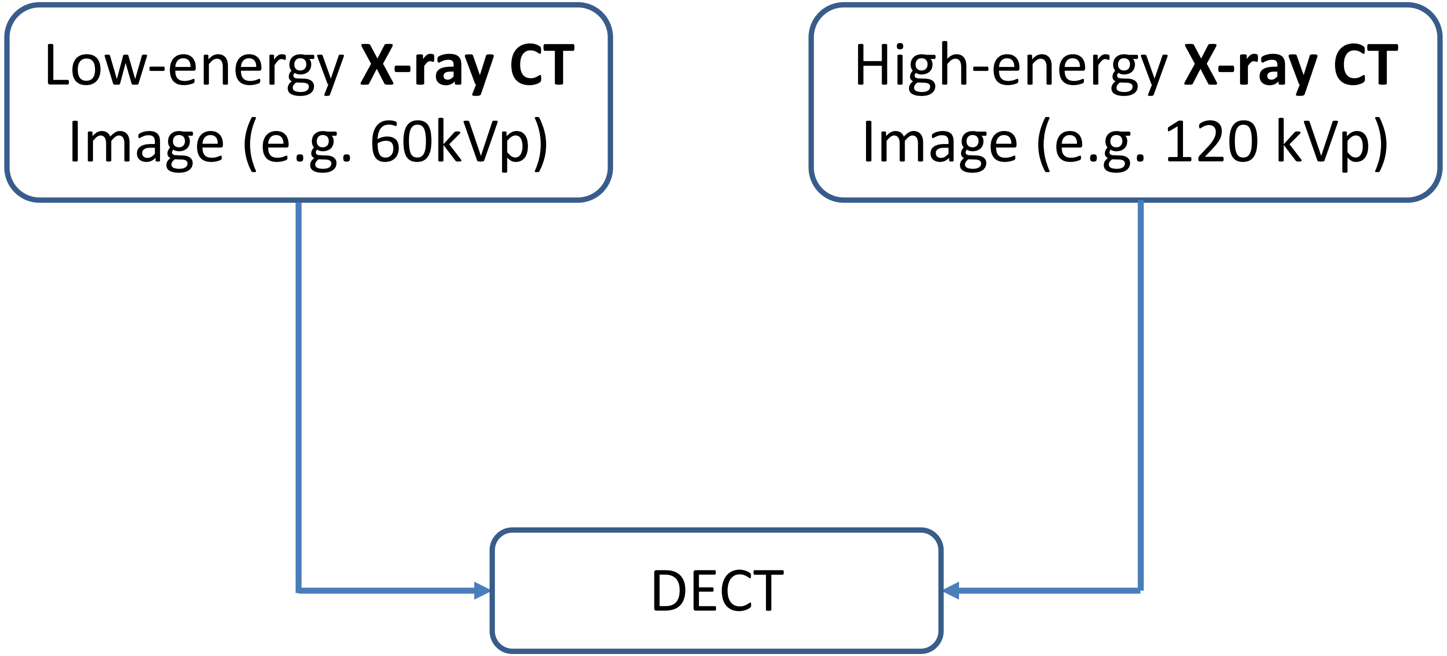
Dual-energy (DE) CT for Quantitative Imaging

- DECT uses two different energies to obtain quantitative material decomposition
- First introduced in late 1970’s
- Renewed interests since ~2005
  - *New implementations*: fast kVp switching, dual-source, spectral detectors

Existing X-ray DECT Paradigm



Low-energy **X-ray CT** Image (e.g. 60kVp)

High-energy **X-ray CT** Image (e.g. 120 kVp)

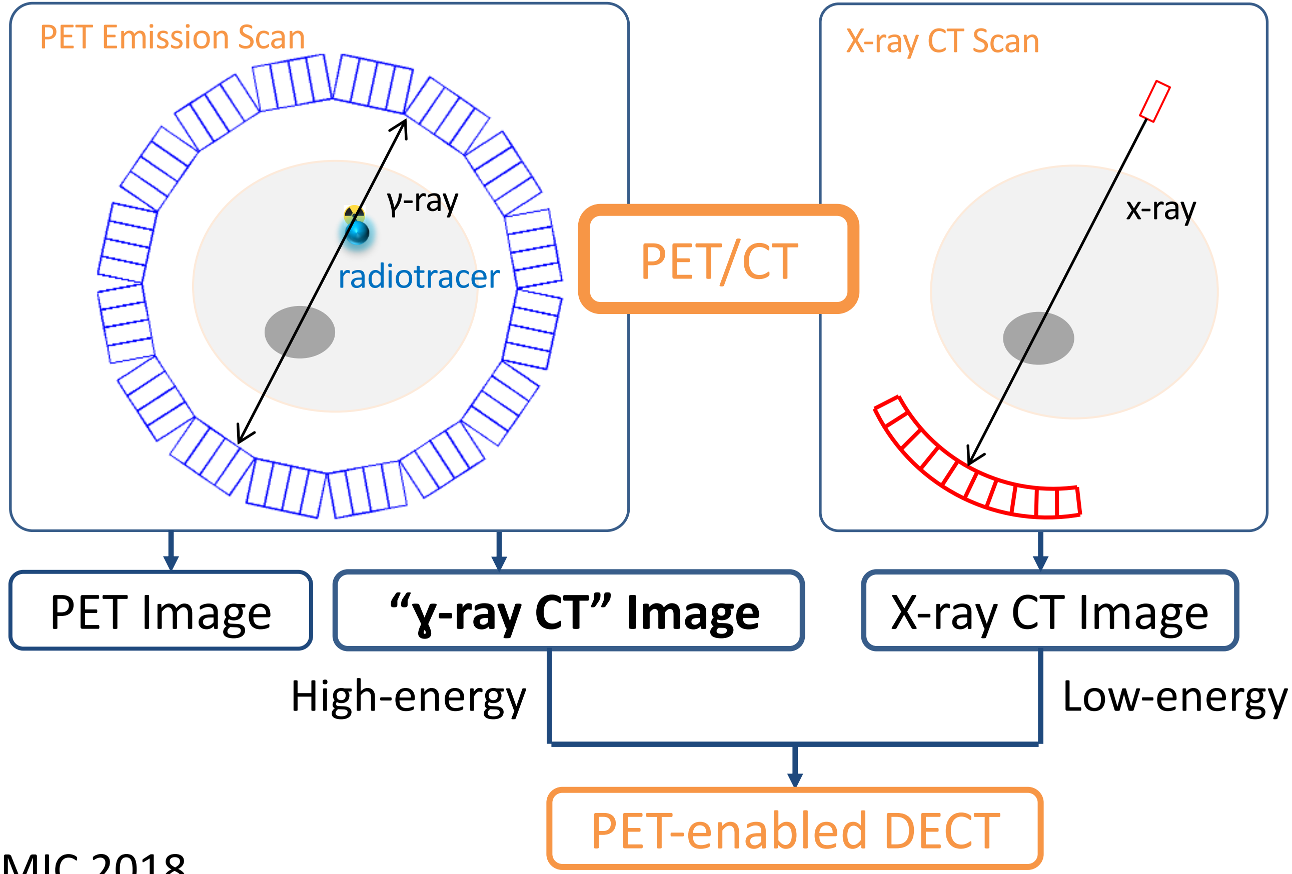
DECT

Implementation on PET/CT Scanners?

- PET/DECT

would require hardware upgrade
or increase radiation dose

Our Proposed Method: PET-enabled DECT



PET/CT

PET Image

“ɣ-ray CT” Image

X-ray CT Image

High-energy

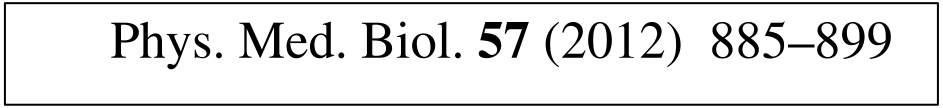
Low-energy

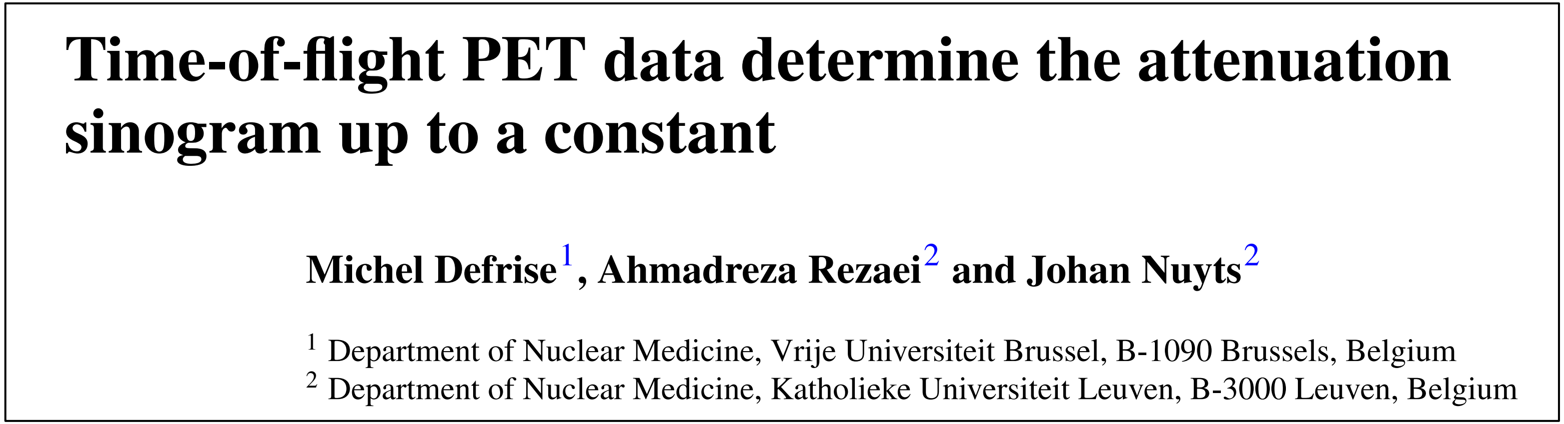
PET-enabled DECT

Wang, IEEE-MIC 2018

Two Ways to Obtain "ɣ-ray CT" from PET Data

- Method 1: Synergistic reconstruction of PET activity and ɣ-ray CT attenuation at 511 keV - known as "MLAA"


Phys. Med. Biol. **57** (2012) 885–899


Time-of-flight PET data determine the attenuation sinogram up to a constant

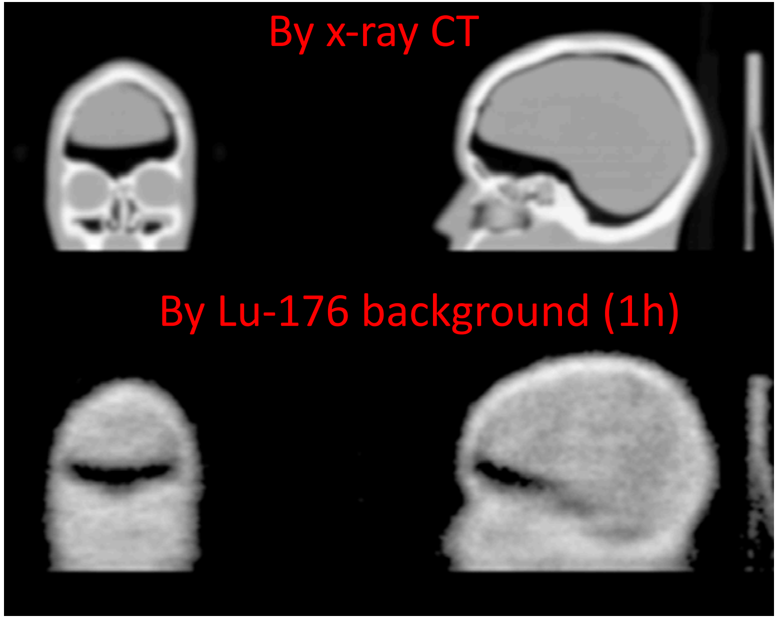
Michel Defrise[1], **Ahmadreza Rezaei**[2] **and Johan Nuyts**[2]

[1] Department of Nuclear Medicine, Vrije Universiteit Brussel, B-1090 Brussels, Belgium
[2] Department of Nuclear Medicine, Katholieke Universiteit Leuven, B-3000 Leuven, Belgium

MLAA implementation: Rezaei *et al* 2012; Erdogan & Fessler 1999

Two Ways to Obtain “ɣ-ray CT” from PET Data

- Method 2: Reconstruction of ɣ-ray CT attenuation from intrinsic background radiation of PET LSO scintillators



Rothfuss *et al.* PMB 2014

Comparison with Previous Works in PET/CT

- Previous attention: To break up with x-ray CT

Transmission-less PET **attenuation correction**

- Our interest: To keep the x-ray CT

PET-enabled **DECT** imaging

Quantitative Material Decomposition

- Example: 3-material decomposition (*fat, soft tissue,* and *calcium*)

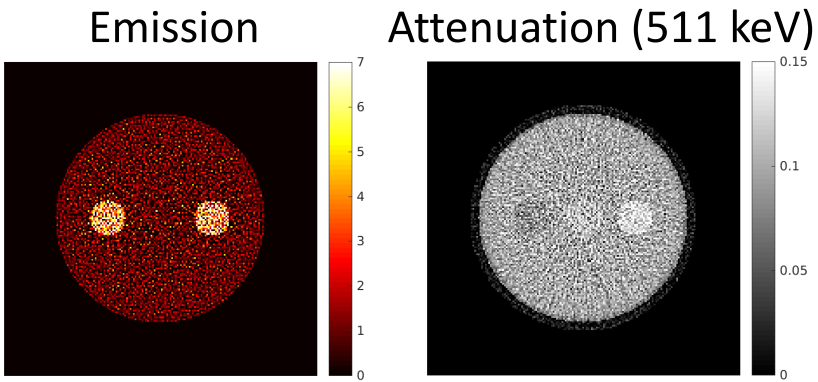
$$\begin{array}{cccc} \begin{bmatrix} u^L \\ u^H \end{bmatrix} & = & \begin{bmatrix} u_F^L & u_S^L & u_C^L \\ u_F^H & u_S^H & u_C^H \end{bmatrix} & \begin{bmatrix} \rho_F \\ \rho_S \\ \rho_C \end{bmatrix} \\ \boldsymbol{u} & & \boldsymbol{B} & \boldsymbol{\rho} \\ \text{DECT} & & \text{Basis Functions} & \text{Fraction} \end{array}$$

- Material fraction $\boldsymbol{\rho}$ is estimated using optimization

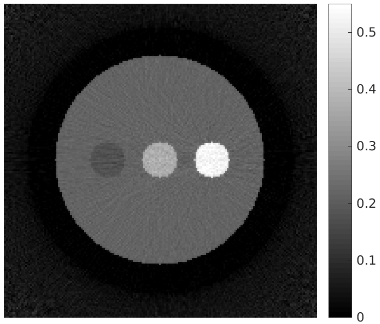
$$\hat{\boldsymbol{\rho}} = \text{argmin}_{\rho} \, ||\boldsymbol{u} - \boldsymbol{B}\boldsymbol{\rho}||^2$$

Preliminary Simulation Results for Proof of Concept

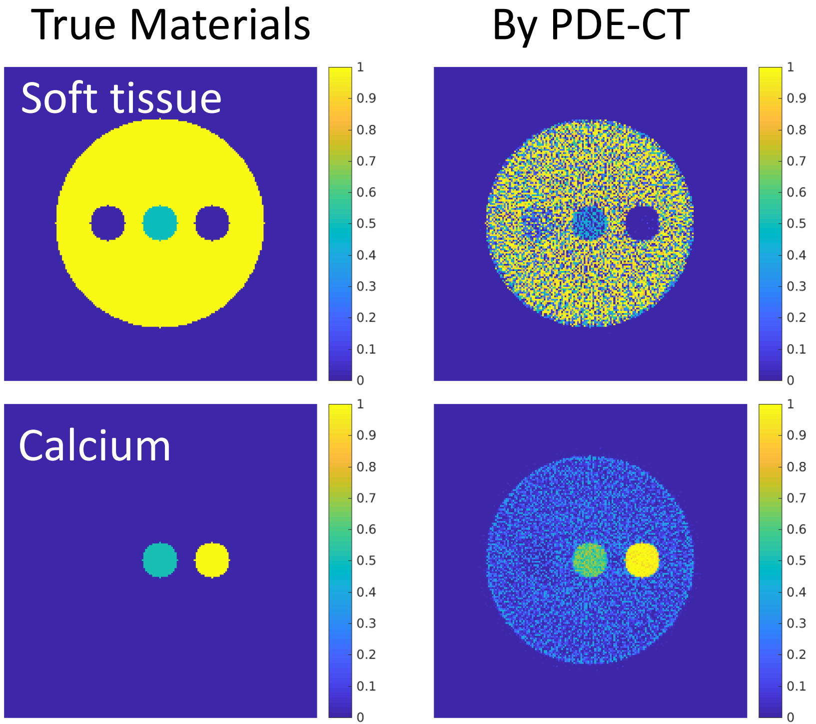
A. MLAA reconstruction of TOF-PET data


Emission
Attenuation (511 keV)

B. X-ray CT (60 keV)



C. Material Decomposition


True Materials
By PDE-CT

Wang, IEEE-MIC 2018

Noise Challenge and Solution

- Challenge

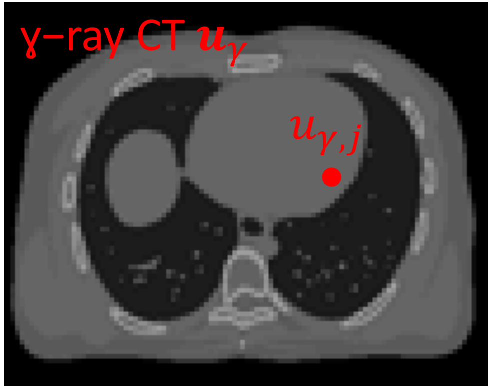
  High noise associated with PET emission data

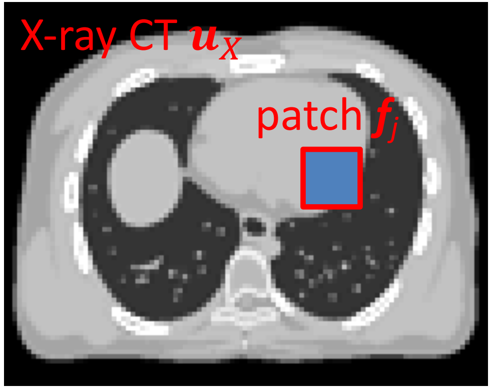
- Solution: Guide the reconstruction of ɣ-ray CT from PET data using the available **x-ray CT** data
  - Good initial estimate
  - Anatomical prior

Kernel Representation for ɣ-ray CT Image

- Pixel j of a ɣ-ray image $\boldsymbol{u}_\gamma$ is associated with a feature vector $\boldsymbol{f}_j$ on the x-ray CT image $\boldsymbol{u}_X$
- Describe the intensity ("label") $u_{\gamma,j}$ as a linear function in the kernel space using "kernel trick"

$$u_{\gamma,j} = \sum_{l=1}^{N} \alpha_l \kappa(\boldsymbol{f}_j, \boldsymbol{f}_l) = [\boldsymbol{K\alpha}]_j$$





More details on the kernel methods: Wang&Qi ISBI 2013; Wang&Qi IEEE T-MI 2015; …

Kernelized MLAA

- Standard MLAA (maximum likelihood estimation of activity and attenuation) for time-of-flight PET emission data

$$\hat{\boldsymbol{x}}, \hat{\boldsymbol{u}} = \arg\max_{\boldsymbol{x},\boldsymbol{u}} \; L(\boldsymbol{y}^{\mathrm{TOF}} | \bar{\boldsymbol{y}}\,(\boldsymbol{x}, \boldsymbol{u}))$$

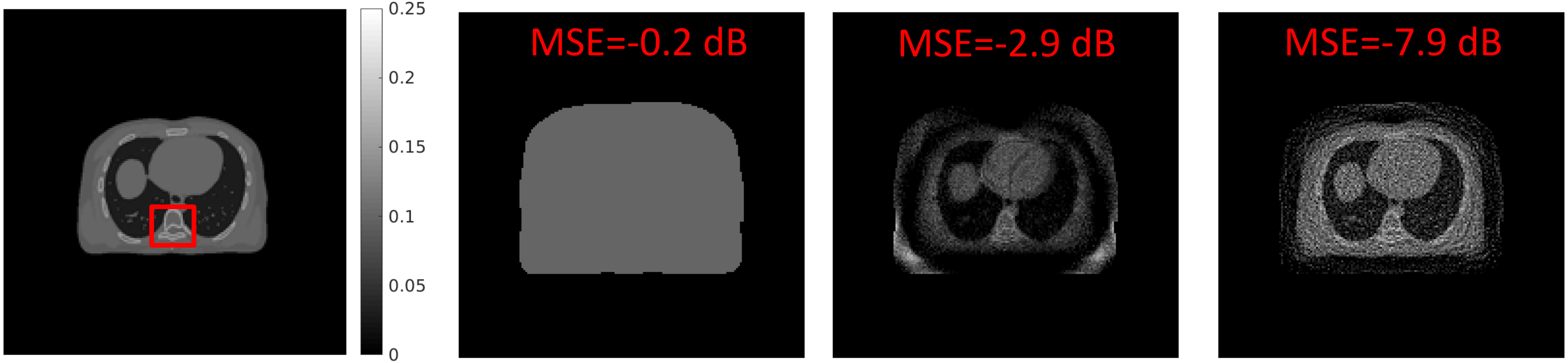
- Kernelized MLAA

$$\hat{\boldsymbol{x}}, \hat{\boldsymbol{\alpha}} = \arg\max_{\boldsymbol{x},\alpha} \; L(\boldsymbol{y}^{\mathrm{TOF}} | \bar{\boldsymbol{y}}\,(\boldsymbol{x}, \boldsymbol{K}\boldsymbol{\alpha}))$$

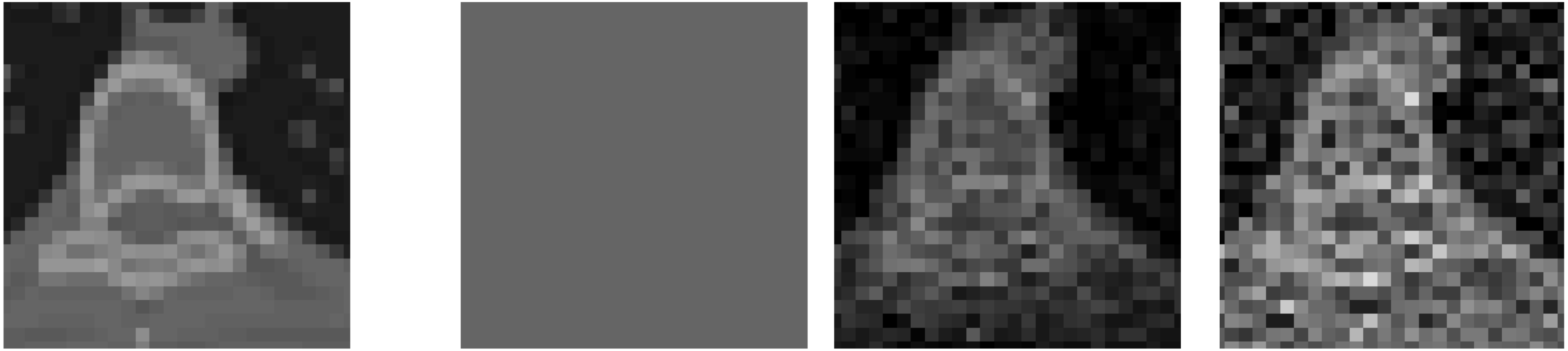
where the kernel matrix $\boldsymbol{K}$ is derived from x-ray CT image.

- Final image estimate from K-MLAA is $\hat{\boldsymbol{u}} = \boldsymbol{K}\hat{\boldsymbol{\alpha}}$

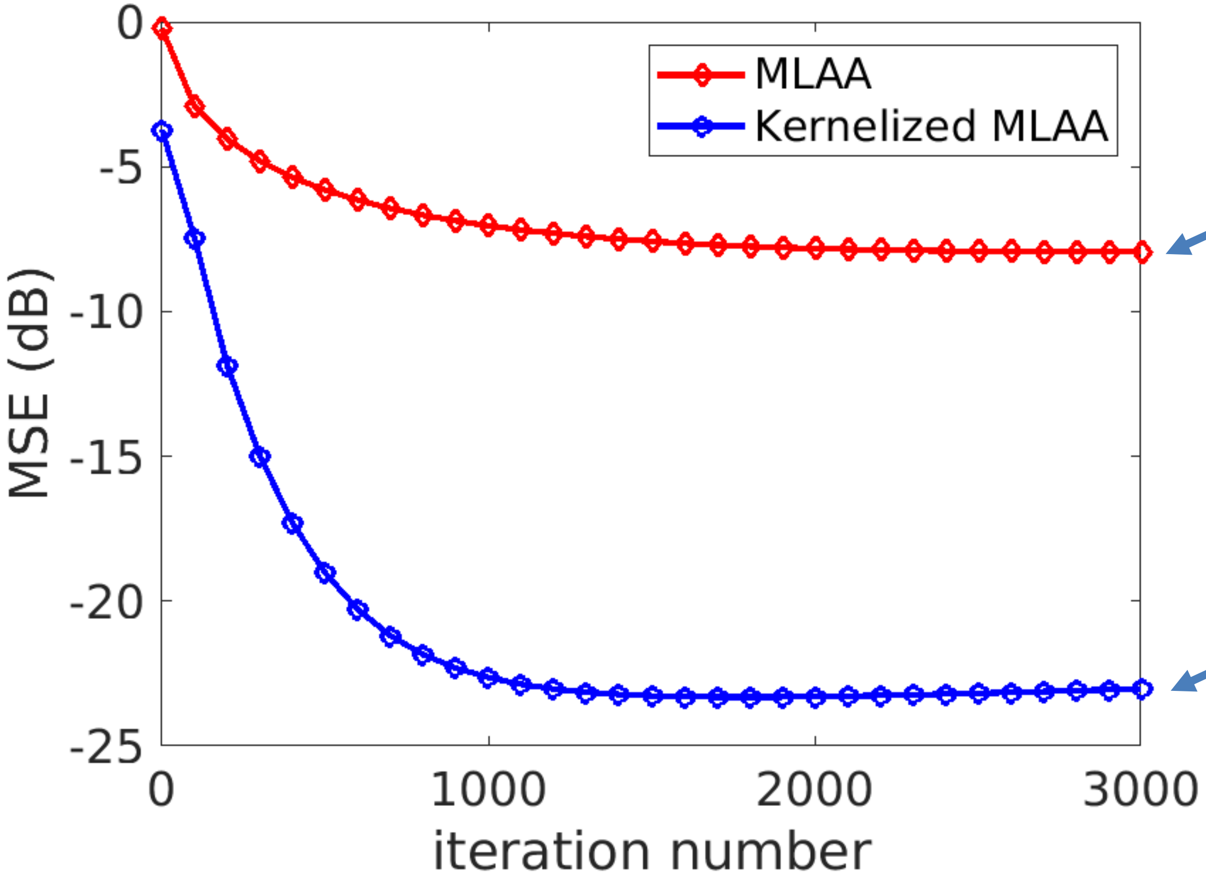
Results from a Computer Simulation: MLAA

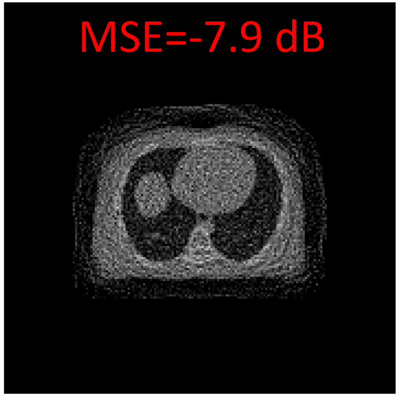


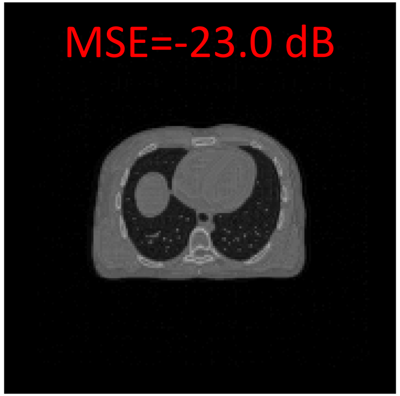
Ground truth **Iteration 0** **Iteration 100** **Iteration 3000**



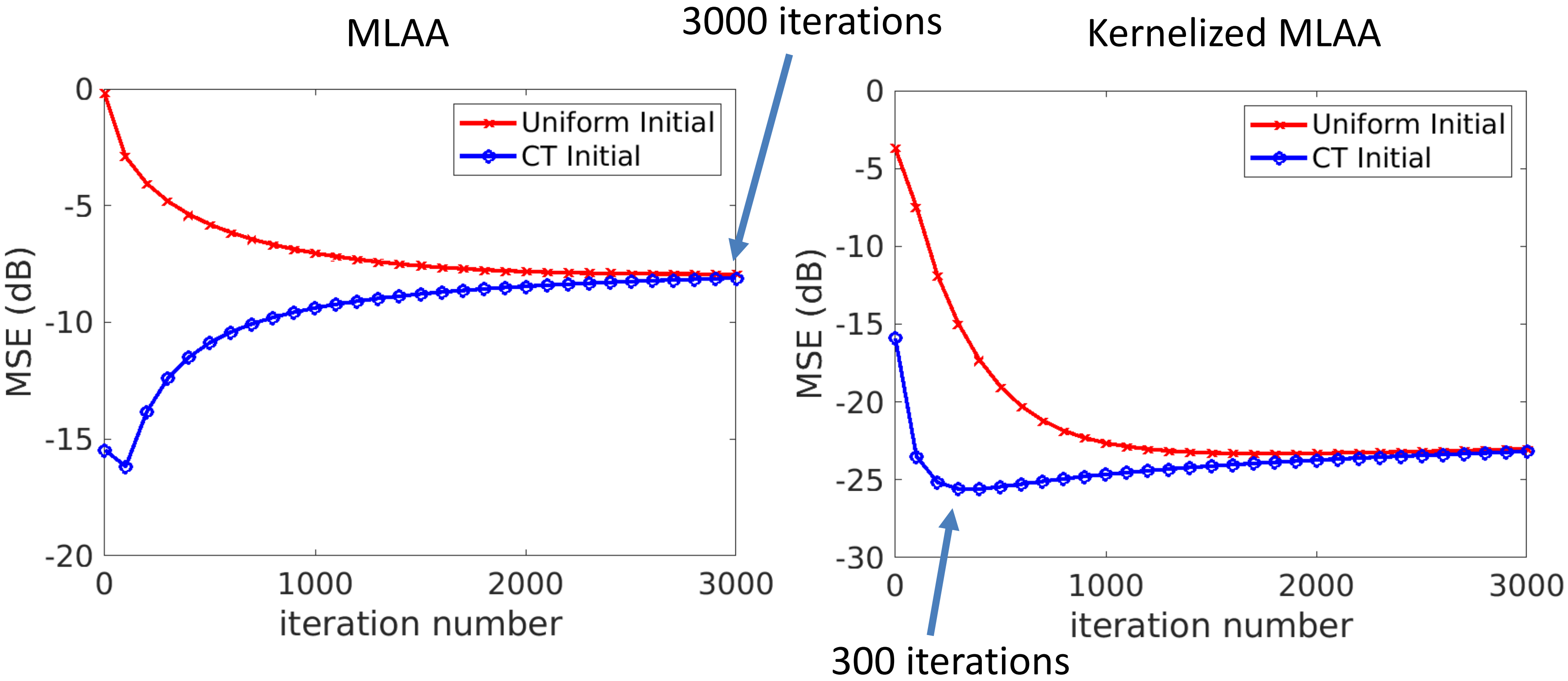
Kernelized MLAA Dramatically Improves Image Quality







Use of CT Initial Accelerates Kernelized MLAA



MLAA

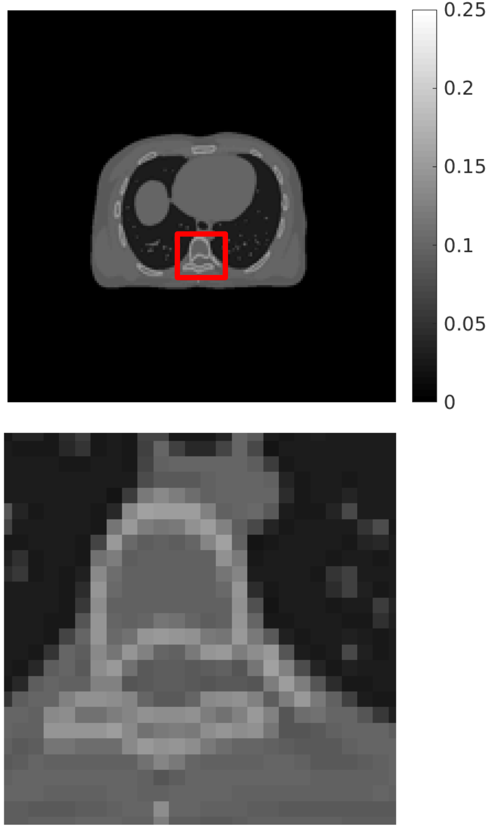
Kernelized MLAA

3000 iterations

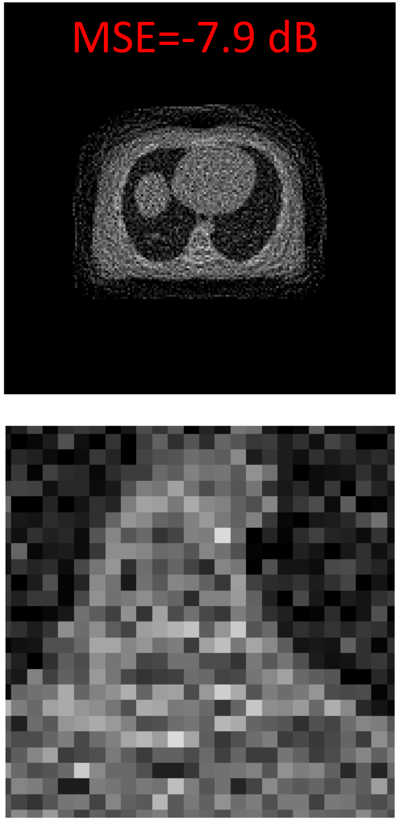
300 iterations

Demonstration of Improved Image Quality

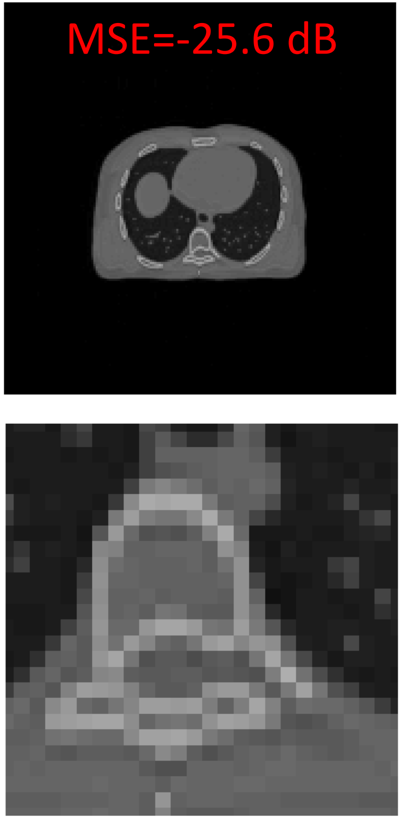
Ground truth



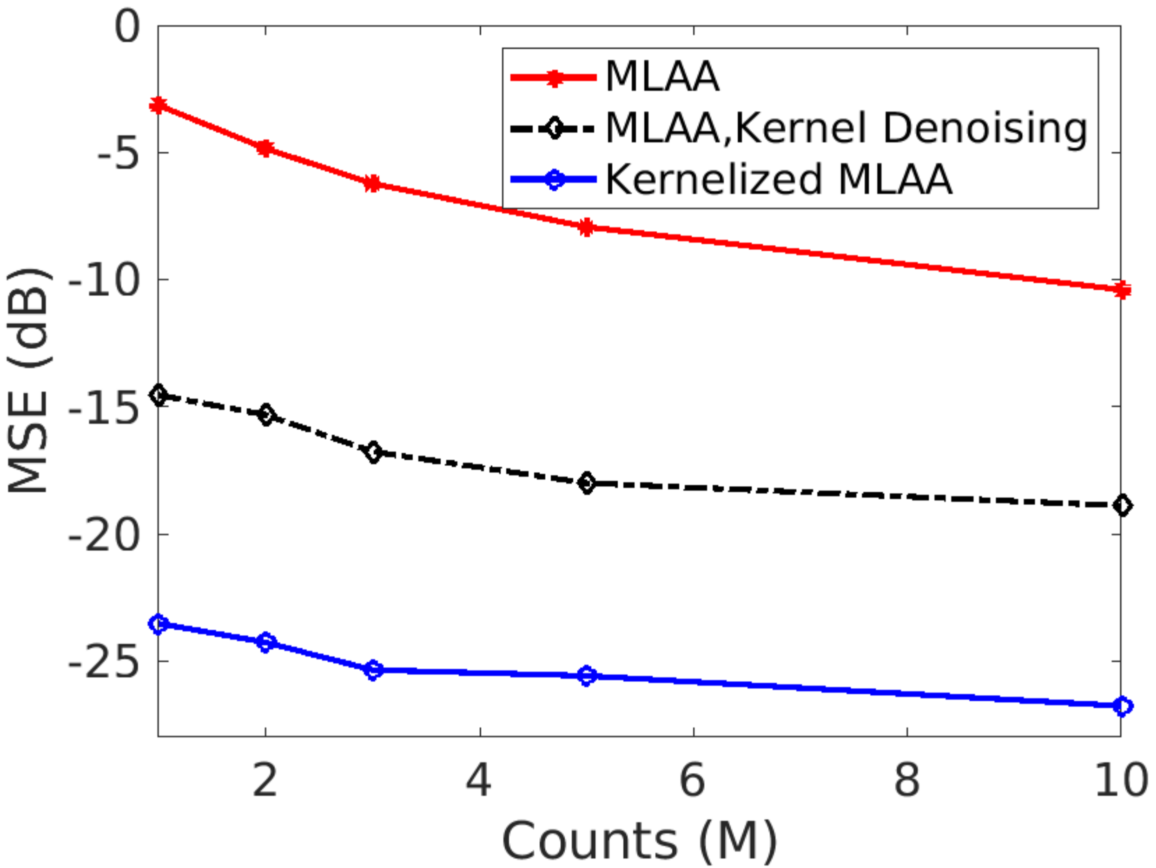
Existing Method



Proposed Method



Quality Improvement Across Different Count Levels



Summary and Future Directions

- We have shown the feasibility of a PET-enabled dual-energy spectral CT imaging method
- Many more algorithm development opportunities
  - PET-enabled spectral CT (PS-CT) imaging (511 keV, 307 keV, 202 keV γ-rays and ≤ 140 keV x-ray)
  - Super-resolution PS-CT
  - Synergistic and direct reconstructions

Acknowledgements

- NIH Funding: R21 EB027346-01A1
- Valuable discussions with Drs. Jinyi Qi, Simon R. Cherry, Ramsey D. Badawi, and John M. Boone at UC Davis

- MATLAB code to be released at:

https://wanglab.faculty.ucdavis.edu/code