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Removing Blooming Artifacts with Binarized Deconvolution

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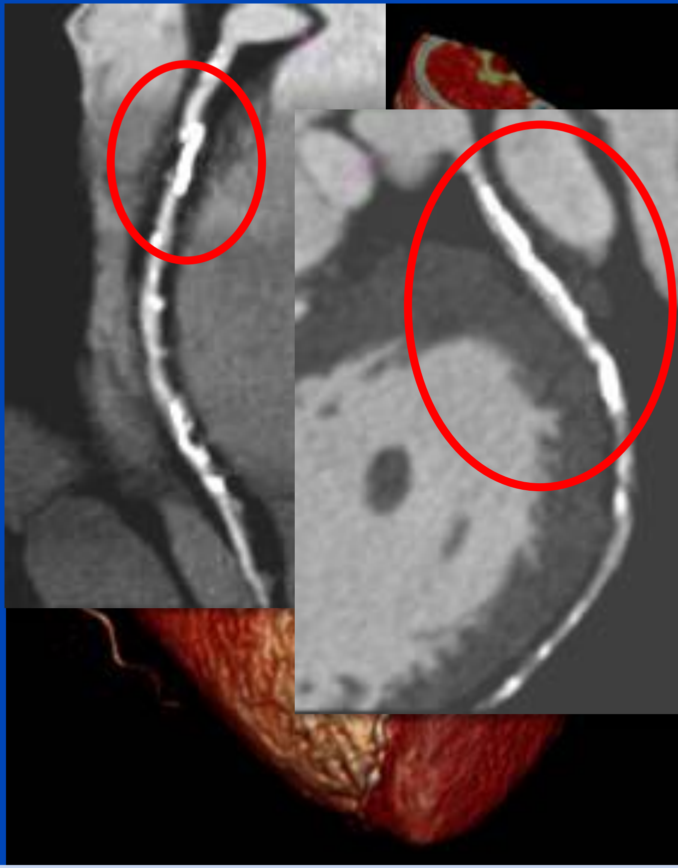
DEUTSCHES KREBSFORSCHUNGSZENTRUM IN DER HELMHOLTZ-GEMEINSCHAFT

Cardiac CT – Motivation

- The coronary artery disease is still one of the most dominating causes of death in the western world.
- Cardiac CT is a desirable non-invasive alternative to invasive coronary angiography.
- Imaging system limits spatial resolution → PSF of the system generates blooming artifacts which reduce the contrast at high contrast structures.
- Blooming artifacts arising from calcified vessels lead to an over-estimation of the degree of luminal narrowing and to loss of the plaque's morphology.

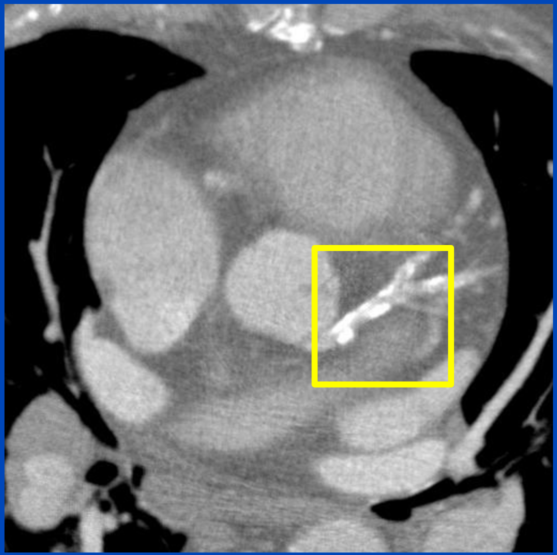


Data courtesy of Dr. Michael Lell, Erlangen, Germany

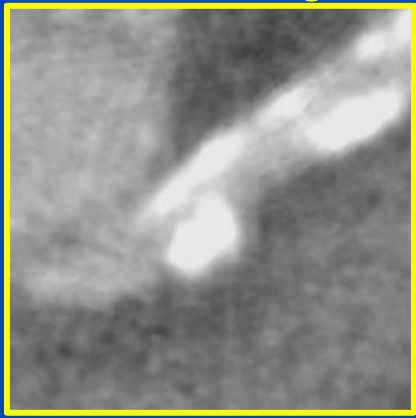
Aim

- To estimate the morphology and the CT-value of the calcification without the adverse effect of the PSF of the imaging system by an image based deconvolution approach.
- The proposed method estimates the correct CT-value and the correct morphology of the calcification by assuming that one calcification consists of a compact homogeneous region with a constant CT-value.

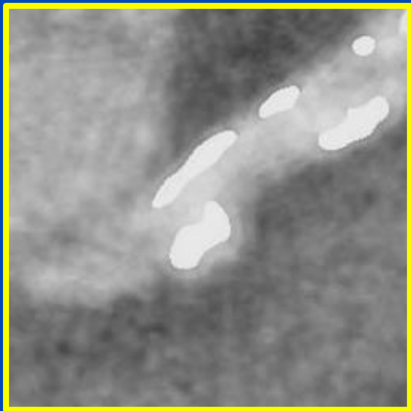
Cardiac reconstruction



Calcification + Blooming



Estimated calcification



C **= 0 HU,** ***W*** **= 1000 HU**

Basic Idea of Binarized Deconvolution

- In CT the formation of the observed image g can be described as a convolution of the real image f with the PSF K of the imaging system:

$$K \cdot f = g$$

- The observed image g can be split into background g_{BG} and the calcification plus blooming g_C so that:

$$g_{\mathrm{C}} = g - g_{\mathrm{BG}}$$

- The proposed method makes use of the assumption that a single calcification g_C consists of a continuous region with almost constant CT-value:

$$K \cdot f_{\mathrm{B}} \cdot c = g_{\mathrm{C}}$$

  - f_B = binary image → shape
  - c = factor for CT-value

- Task: find f_B and c.

Binarized Deconvolution – Workflow

Initialization

↓

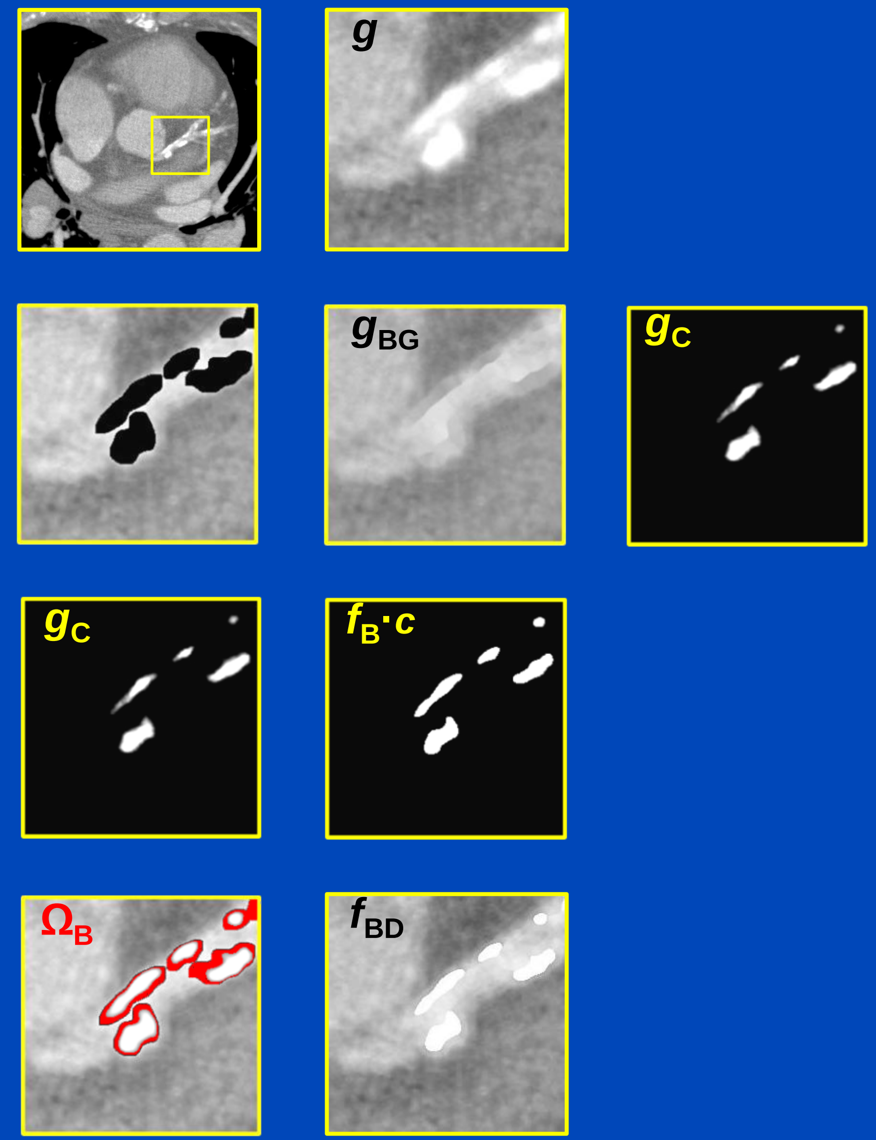
Separation of calcification and background

↓

Removal of blooming artifacts

↓

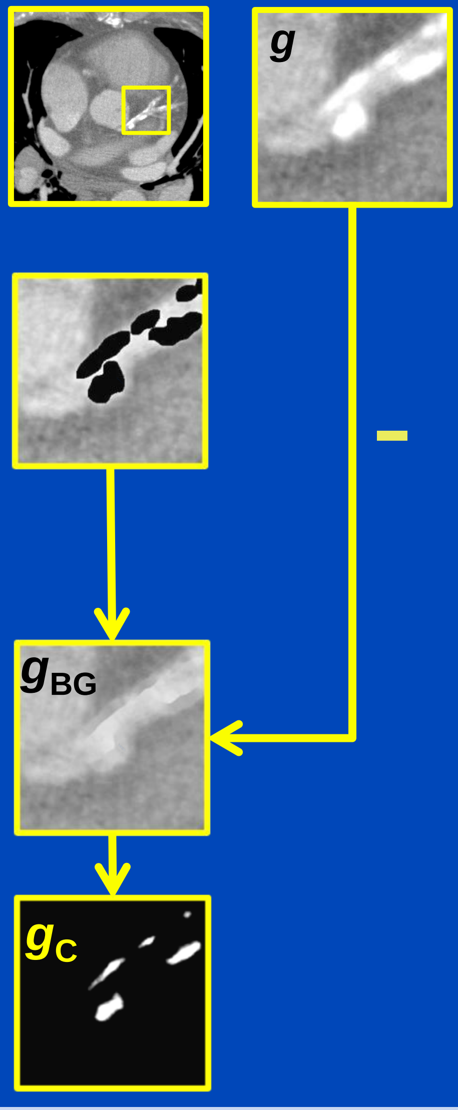
Combination of calcification and background



Result

Separation of Calcification and Background

- **Detect and select calcification**
  - **Manually chosen (to date)**
- **Upsample to fine voxel grid ($\Delta x = \Delta y = 0.1$ mm)**
  - **For a more exact extraction and deconvolution**
- **Extract calcification and blooming**
  - **Thresholding approach proposed by Steckmann et al[1].**
- **Calculate background g_{BG} by inpainting the thresholded image**
  - **Erosion-based method which uses the local surroundings of the borders of the defect which are to be continued into the defect region.**
- **Subtract g_{BG} from g to obtain calcification + blooming: $g_C = g - g_{BG}$**



[1]Steckmann,S. and Kachelrieß M., "Blooming artifact reduction for cardiac CT," Nuclear Science Symposium Conference Record (NSS/MIC) , 2030–2035 (Oct. 2010).

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Removal of Blooming Artifacts

$$C(\boldsymbol{f}_\mathrm{B}, c) = ||\boldsymbol{K} \cdot \boldsymbol{f}_\mathrm{B} \cdot c - \boldsymbol{g}_\mathrm{C}||_2^2$$

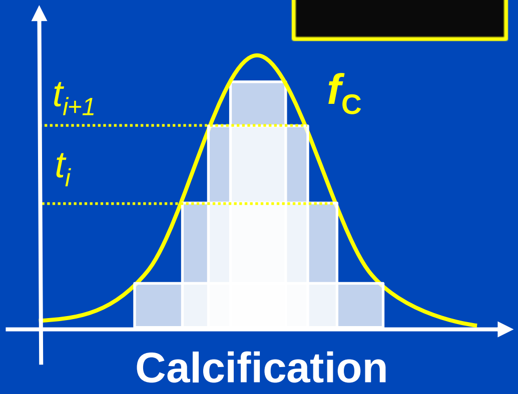
- f_B = binary image → shape
- c = factor for CT-value

Deconvolve g_C with TV penalty → f_C

$$\mathrm{C}_\mathrm{TV}(\boldsymbol{f}_\mathrm{C}) = ||\boldsymbol{K} \cdot \boldsymbol{f}_\mathrm{C} - \boldsymbol{g}_\mathrm{C}||^2 + \beta \mathrm{TV}(\boldsymbol{f}_\mathrm{C})$$

Binarize f_C with thresholds t → $f_B(t)$
t = [0, max(f_C)]

$$f_{\mathrm{B}j}(t) = \begin{cases} 1, & \text{if } f_{\mathrm{C}j} > t \\ 0, & \text{if } f_{\mathrm{C}j} \leq t \end{cases}$$



For all thresholds t calculate c minimizing $C(f_B(t), c)$

$$c = \arg\min_{c} ||\boldsymbol{K} \cdot \boldsymbol{f}_\mathrm{B}(t) \cdot c - \boldsymbol{g}_\mathrm{C}||_2^2$$

$$\Rightarrow \quad c = \frac{(\boldsymbol{K} \boldsymbol{f}_\mathrm{B}(t)) \cdot \boldsymbol{g}}{(\boldsymbol{K} \boldsymbol{f}_\mathrm{B}(t))^2}$$

Use the set c^*, f_B^* with the lowest cost $C(f_B(t^*), c^*)$

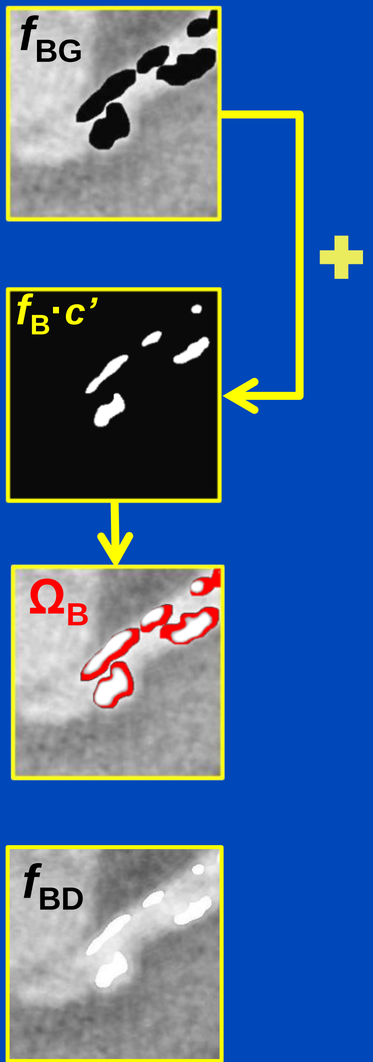
$$(c^*, t^*) = \arg\min_{c,t} \; C(\boldsymbol{f}_\mathrm{B}(t), c)$$

Combination of Calcification and Background

- The factor for the CT-value c^* has to be background corrected due to the subtraction of the background before.
- For that we calculate the mean of the set of voxels of the inpainted image g_{BG} which corresponds to the blooming corrected calcification.
- Add thresholded image and $f_B \cdot c'$.
- Choose the blooming corrected $f_{\Omega B}$ region best solving the cost function:

$$\boldsymbol{f}_{\Omega_\mathrm{B}} = \underset{f_j,\, j \in \Omega_\mathrm{B}}{\arg\min} ||\boldsymbol{K} \cdot (\boldsymbol{f} + \boldsymbol{f}_\mathrm{B}(t^*) \cdot c' + \boldsymbol{f}_\mathrm{BG}) - \boldsymbol{g}||_2^2 + \beta \mathrm{TV}(\boldsymbol{f}).$$

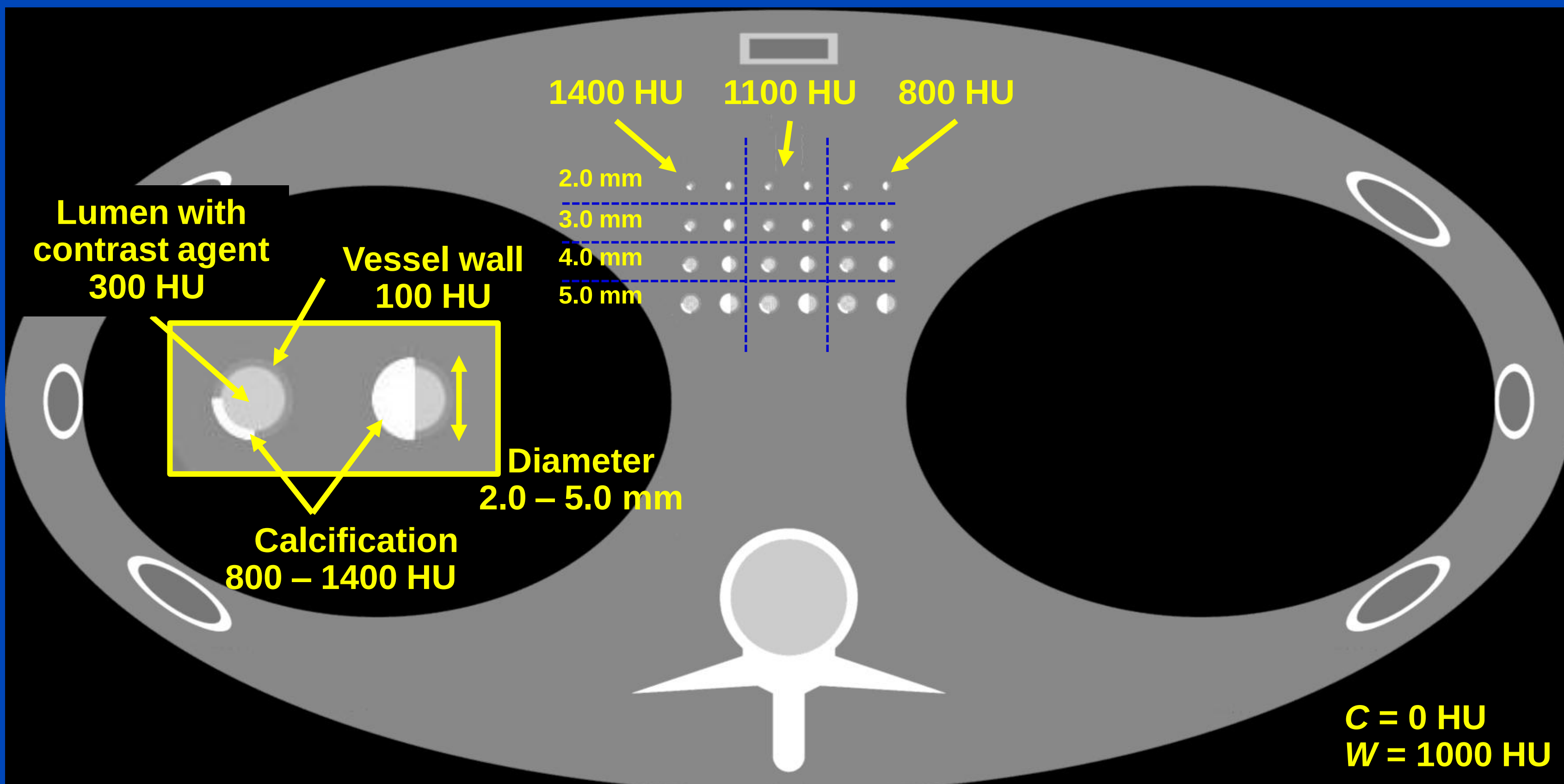
→ Here only voxel values $j \in \Omega_B$ are being updated.



Result

Phantom with Calcified Vessels

- Blood vessels were simulated into the heart region of the Forbild thorax phantom and high density calcium deposits were simulated into the vessels.
- The center portion of large lesions is typically between 800 – 1400 HU.



Assessment of Image Quality

- **Image quality was quantified by computing the normalized cross correlation with ground truth,**

$$\mathrm{NCC} = \frac{1}{L-1} \sum_{x,y \in \Omega} \frac{(f(x,y) - \bar{f})(g(x,y) - \bar{g})}{\sigma_f \sigma_g},$$

  - » ***f*** **= reconstructed image,** ***g*** **= ground truth**
  - » $\boldsymbol{\sigma_f}$**,** $\boldsymbol{\sigma_g}$ **= corresponding standard deviations**
  - » **Ω region for NCC analysis**

- **and with the root mean square deviation to the ground truth:**

$$\mathrm{RMSD} = \sqrt{\frac{1}{|\Omega|} \sum_{x,y \in \Omega} (f(x,y) - g(x,y))^2}.$$

- **Estimated calcium size: voxels exceeding the threshold of 70 % of the CT-value of the calcification are counted as calcified voxels.**

Compared Algorithms

- **Ground truth:**

  Noise-free ten-fold spatial resolution analytical reconstruction of our analytical phantom.

- **FBP reference reconstruction:**

  Ram-Lak kernel (ramp filter till Nyquist frequency).

- **Richardson Lucy deconvolution (RL)[1]:**

  Standard deconvolution technique (often used in CT literature).

$$\boldsymbol{f}^{n+1} = \boldsymbol{f}^{n}\boldsymbol{K}^{T}\left(\frac{\boldsymbol{g}}{\boldsymbol{K}\boldsymbol{f}^{n}}\right)$$

  – g = image to be deconvolved
  – K = convolution kernel
  – K^T = transpose kernel
  – f^n = result after iteration n

- **Proposed binarized deconvolution (BD):**

$$C(\boldsymbol{f}_{\mathrm{B}}, c) = ||\boldsymbol{K} \cdot \boldsymbol{f}_{\mathrm{B}} \cdot c - \boldsymbol{g}_{\mathrm{C}}||_2^2$$

[1]Al-Ameen, et al., "Utilizing a Laplacian–Richardson Lucy procedure for deblurring CT medical images degraded by Gaussian blur," In proceeding of: 5th conference and exhibition on computer communication (Jul. 2012).

Simulation and Reconstruction Setting for Phantom Simulations

Rawdata:

- Siemens SOMATOM Definition Flash Geometry
- $N_{360} = 1160$
- Monoenergetic x-ray spectrum with 80 keV
- 30 HU Poisson noise (in water equivalent tissue)

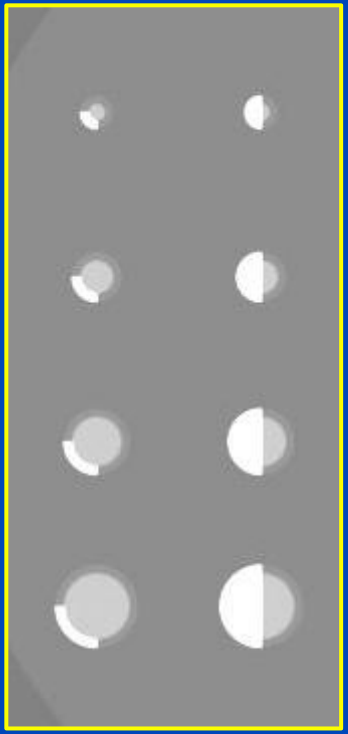
Reconstruction setting:

- Field of view = 250 mm
- $N_x = N_y = 512 \rightarrow \Delta x = \Delta y = 0.5$ mm

BD Algorithm:

- For the deconvolution the PSF was approximated as a spatial invariant Gaussian function
- The PSF is determined with a delta object, simulated into the phantom: $\text{FWHM}_{\text{FBP}} \approx 1.01$ mm $\rightarrow \sigma_{\text{FBP}} \approx 0.43$ mm

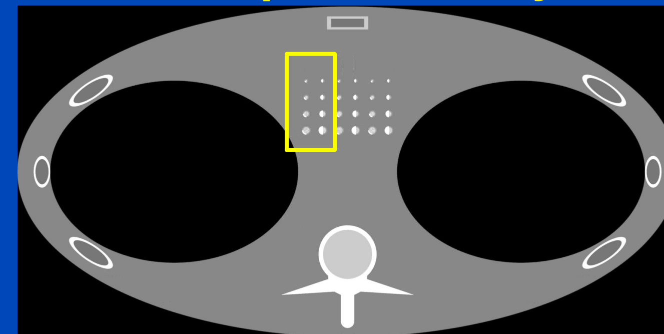
Ground truth
(Calcification = 1400 HU)



Phantom Results

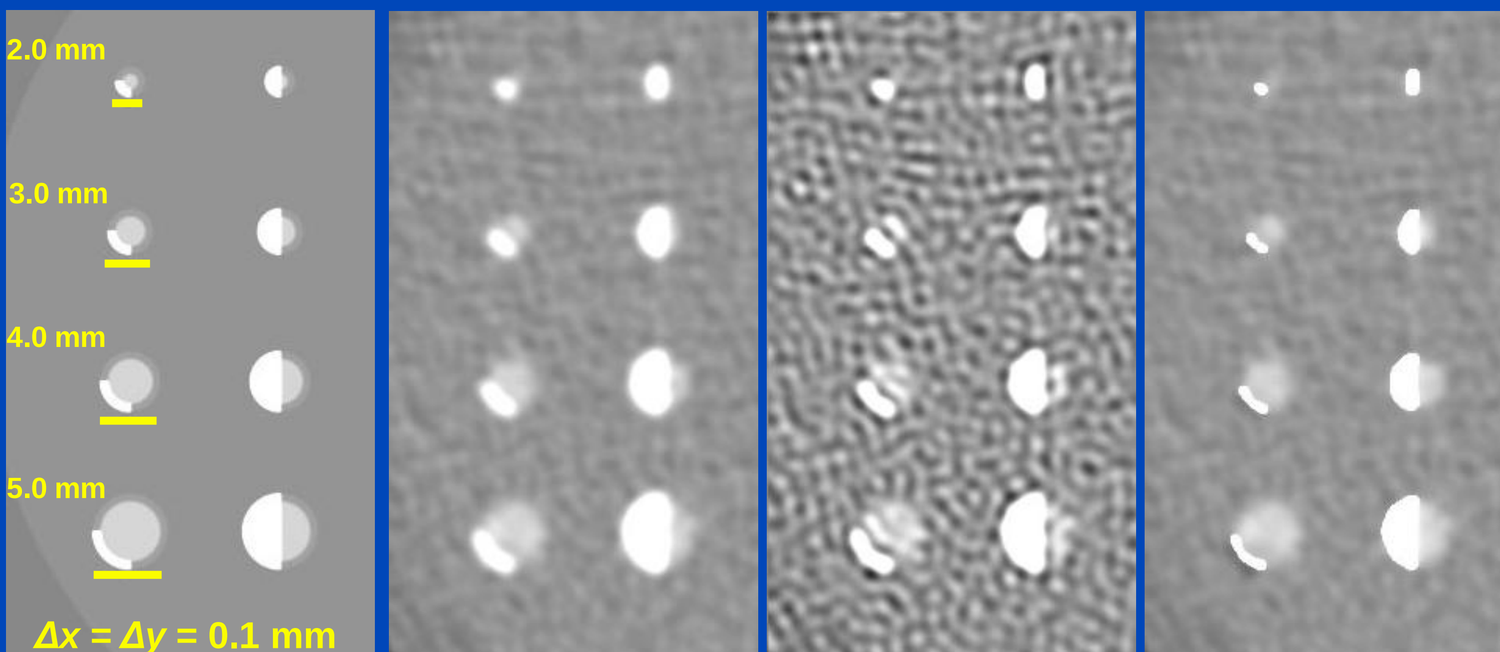
- Reference reconstruction: **FBP** (Ram-Lak)
- Richardson Lucy (**RL**) deconvolution
- Binarized deconvolution (**BD**)

1400 HU pattern analyzed



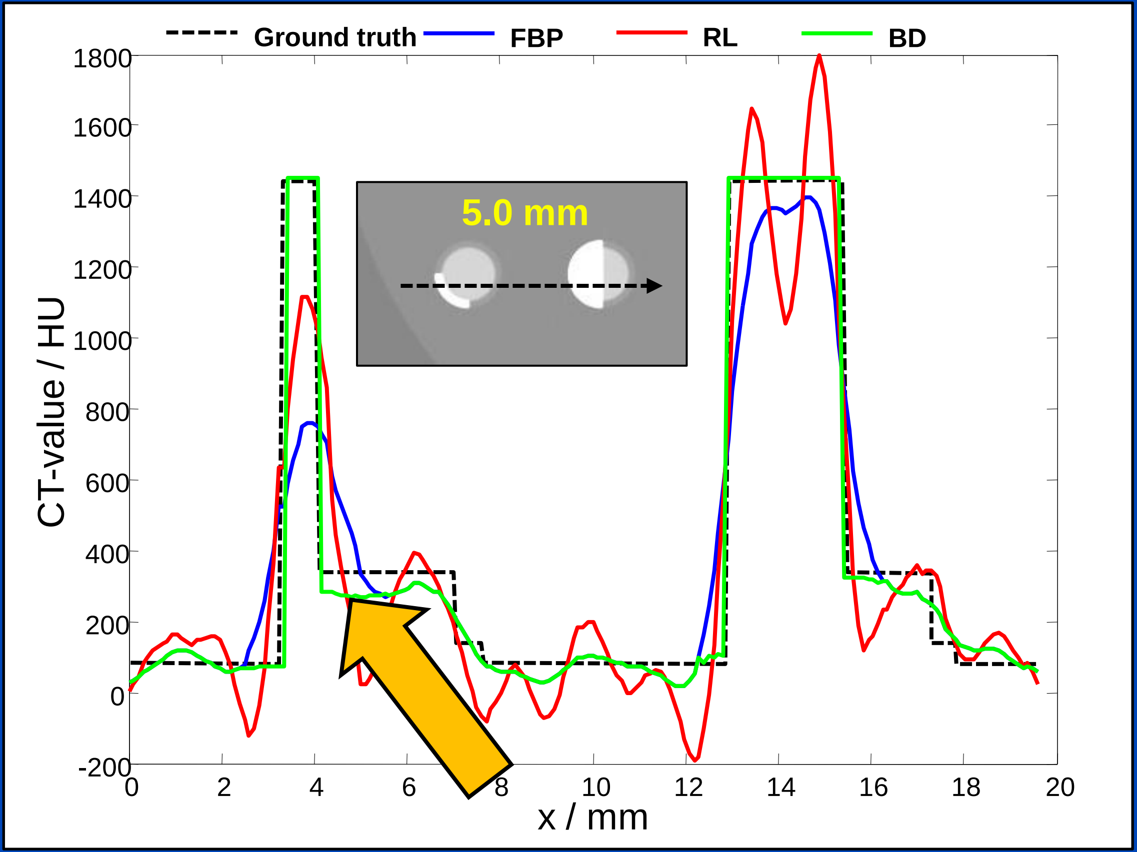
Ground truth
(Calcification = 1400 HU)

FBP | **RL** | **BD**



C = 0 HU, W = 1000 HU

Quantitative Results



Algorithm	RMSD / 10^{-3}	NCC	Calcification size in %
Ground truth	0.0	1.0	100
FBP	0.35	0.53	55.8
RL	0.30	0.62	83.1
BD	0.26	0.71	93.9

Reconstruction Setting for the Patient Data

Rawdata:
- Siemens SOMATOM Definition Flash scanner
- Dual source spiral scan, retrospectively gated scan
- Scan parameters: N_{360} = 1160, tube voltage = 100 kV

Reconstruction setting:
- Reference reconstruction: EPBP[1]
- Retrospective reconstruction, 70% R-R interval reconstruction
- Field of view = 250 mm
- $N_x = N_y = 512 \rightarrow \Delta x = \Delta y = 0.5$ mm

BD algorithm:
- Estimate PSF in Reconstruction:
  - Measure several edge profiles an calculate averaged FWHM
  - d/dr (ESF(r)) = LSF(r) → FWHM by fitting Gaussian to the LSF
  - FWHM ≈ 1.3 mm → $\sigma_{EPBP} \approx 0.56$ mm

[1] Kachelrieß, M., Knaup, M., and Kalender, W., "Extended parallel backprojection for standard three dimensional and phase-correlated four-dimensional axial and spiral cone-beam CT with arbitrary pitch, arbitrary cone-angle, and 100% dose usage," Med. Phys. 31, 1623–1641 (Jun. 2004).

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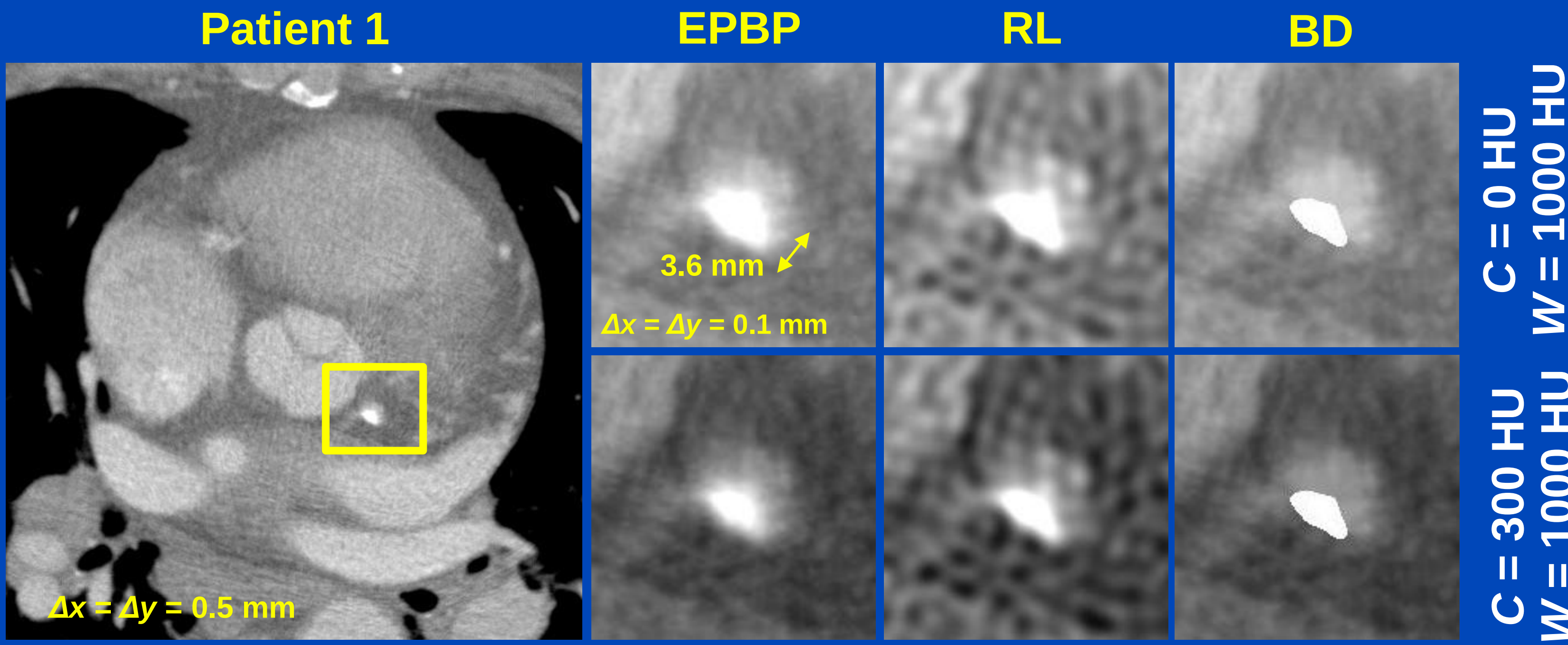
Simple Calcification Patient Data

- The estimation of the calcification size and the degree of the luminal narrowing is now independent from the window level due to the removed blooming.



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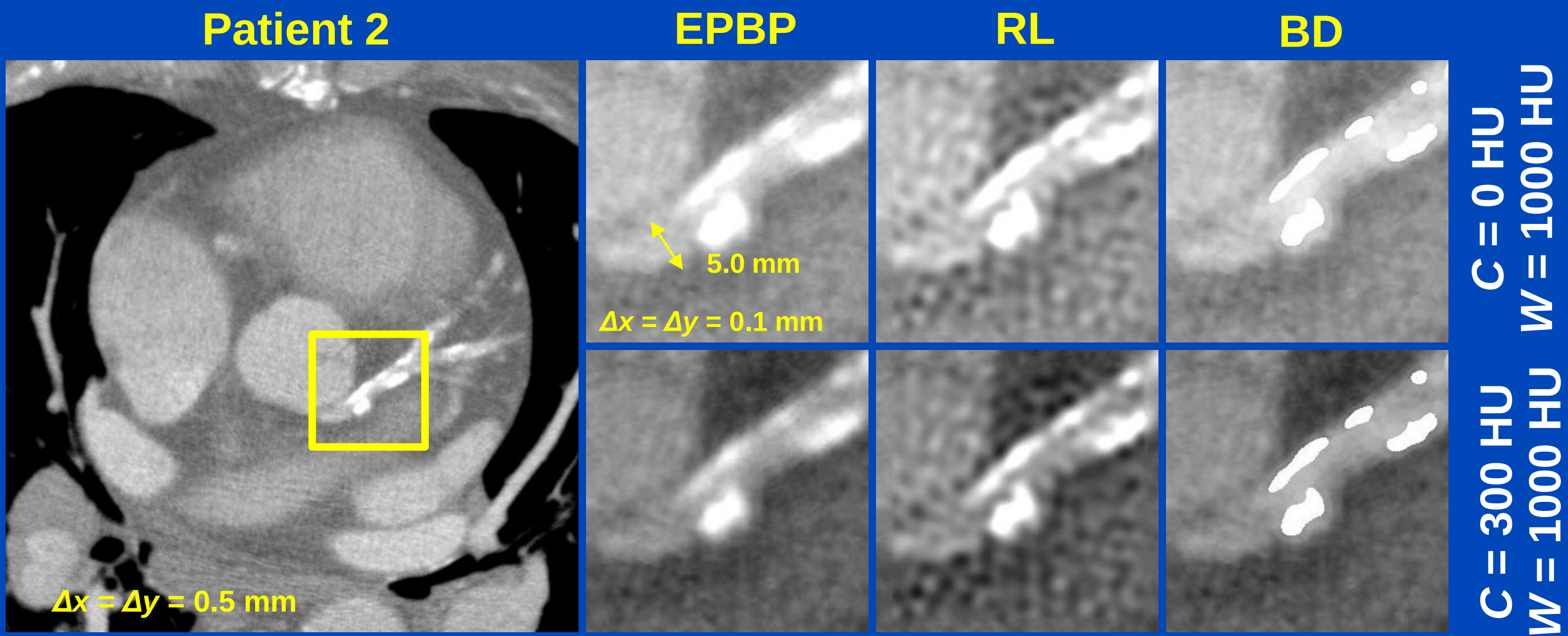
Multiple Calcifications Processed at Once

- The initial patient data results are promising, but a μ-CT study has to be done to verify the correctness of the CT-values and the morphology.



Summary & Conclusion

- The phantom study shows good results → the correct CT-value and the correct morphology of the calcification can be restored.
- The visibility of the calcification and its borderline to the lumen is independent from the window level.
- The patient data results are promising.
- A µ-CT study has to be done to verify the correctness of the CT-values and the morphology and to tune the various parameters of the method.

Thank You!

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This presentation will soon be available at www.dkfz.de/ct.