Cardiac Motion Compensation from Short Scan CT Data: A Comparison of Three Algorithms

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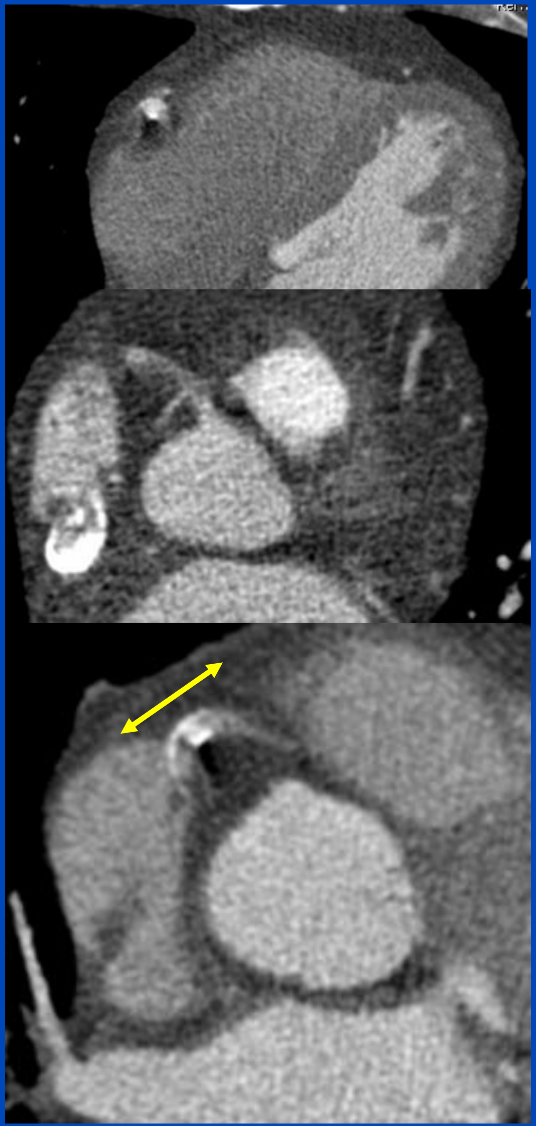
Introduction

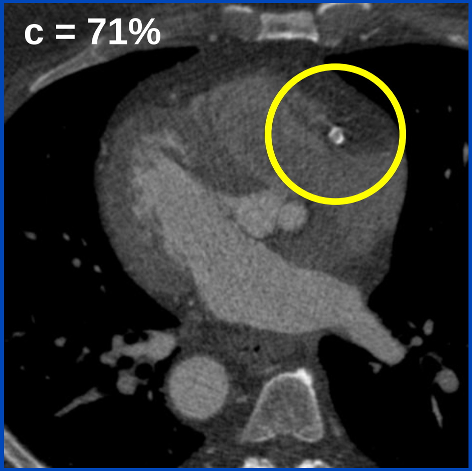


- **In cardiac CT, the imaging of small and fast moving vessels places high demands on the spatial and temporal resolution of the reconstruction.**
- **Mean displacements of** $$d \approx \frac{t_{rot}}{2}\,\bar{v} \approx \frac{250}{2}\,\mathrm{ms}\;50\,\frac{\mathrm{mm}}{\mathrm{s}} = 6.25\ \mathrm{mm}$$ **(RCA mean velocity measurements[1,2,3,4]) are possible.**
- **Standard FDK-based cardiac reconstruction might have an insufficient temporal resolution introducing strong motion artifacts.**

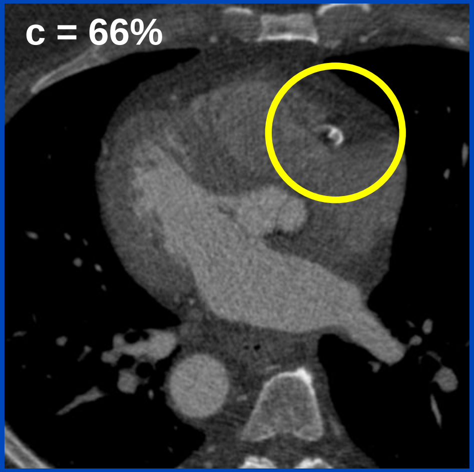
[1] Husmann et al. Coronary Artery Motion and Cardiac Phases: Dependency on Heart Rate - Implications for CT Image Reconstruction. Radiology, Vol. 245, Nov 2007.
[2] Shechter et al. Displacement and Velocity of the Coronary Arteries: Cardiac and Respiratory Motion. IEEE Trans Med Imaging, 25(3): 369-375, Mar 2006
[3] Vembar et al. A dynamic approach to identifying desired physiological phases for cardiac imaging using multislice spiral CT. Med. Phys. 30, Jul 2003.
[4] Achenbach et al. In-plane coronary arterial motion velocity: measurement with electron-beam CT. Radiology, Vol. 216, Aug 2000.

Reducing Motion Artifacts in Cardiac CT

- For single source systems, several software-based solutions have been developed.
- In view of dose optimized scan protocols, we want to focus on methods, which utilize only the data needed for the reconstruction of a single cardiac phase (short scan data ≙ 180° + fan angle).
- Especially beneficial in cases of patients with high or irregular heart rates or non-optimally chosen gating positions.



"Best phase"



Non-optimally chosen gating position

Reducing Motion Artifacts in Cardiac CT

- **Two approaches as a software-based solution to increase the temporal resolution are common:**

- **Reconstruction using less data than needed for the reconstruc-tion of a single cardiac phase**
  - **Iterative reconstruction algorithms dealing with limited angle artifacts:**
    1. **Temporal Resolution Improvement Method (TRIM[1])**

       [1] Schöndube et al. "Evaluation of a novel CT image reconstruction algorithm with enhanced temporal resolution" Medical Imaging 2016 : Physics of Medical Imaging, Proc. of SPIE Vol. 7961: 1605-7422

- **Reducing motion artifacts by compensating for motion occurring during data acquisition**
  - **Motion compensation (MoCo) algorithms from short-scan data:**
    2. **Motion Artifact Metric method (MAM[2])**

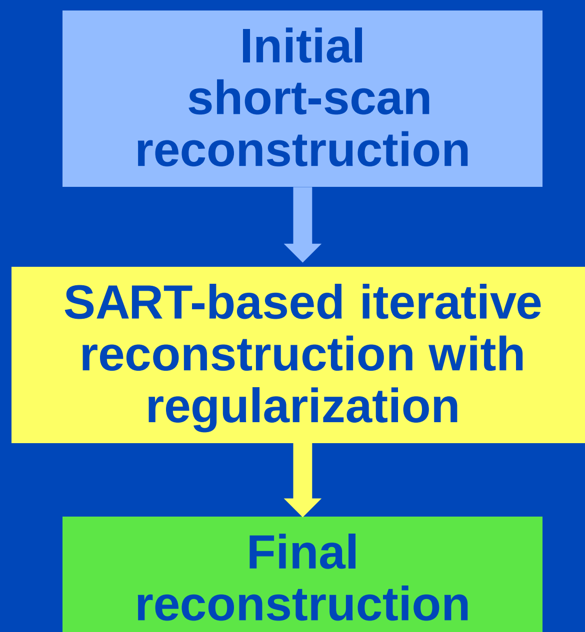
       [2] Rohkohl et al., „Improving best-phase image quality in cardiac CT by motion correction with MAM optimization" , Med. Phys. 40 (3): 0094-2405, March 2013.

    3. **Motion Compensation based on Partial Angle Reconstructions (PAMoCo[3])**

       [3] Hahn et al., „Reduction of motion artifacts in cardiac CT based on partial angle reconstructions from short scan data", Medical Imaging 2016 : Physics of Medical Imaging, Proc. of SPIE

TRIM

- **Aim: Enhance the temporal resolution in CT beyond the short-scan limit.**
- **Idea: Increase the temporal resolution by using less than the short scan data for reconstruction ($p' \approx 120°$).**
- **Workflow:**



TRIM

- **Minimize the cost function**

$$C(f) = \underbrace{C_1(f)}_{\text{raw data fidelity}} + \beta \underbrace{C_2(f)}_{\text{regularization term}}.$$

- **Extension of an SART, which optimizes the raw data fidelity**

$$C_1(f) = |\mathsf{X}\underbrace{f}_{\text{Image to be reconstructed}} - p'|^2.$$

by a regularization term $C_2(f)$.

- **Optimizing only $C_1(f)$ would introduce limited angle artifacts, since $p´ \approx 120°$ covers only a subset of the short-scan range.**

TRIM

- To prevent from limited angle artifacts, local histograms of the initial short-scan reconstruction are calculated.
- After each SART step, the pixel values of the reconstruction, which are far away from a maximum in the histogram, are slightly pushed towards the closest maximum.
- Hence, the regularization term appears as a histogram constraint

$$C_2(f) = -\sum_i \log(\omega(f(x_i))).$$

Probability density defined by the local histograms

MAM

- Aim: Improve the image quality of best phase images.
- Idea: Estimate motion vectors by measuring the amount of motion artifacts in the reconstruction (adaption of auto focus concept from photography).
- Workflow:

Initialization

↓

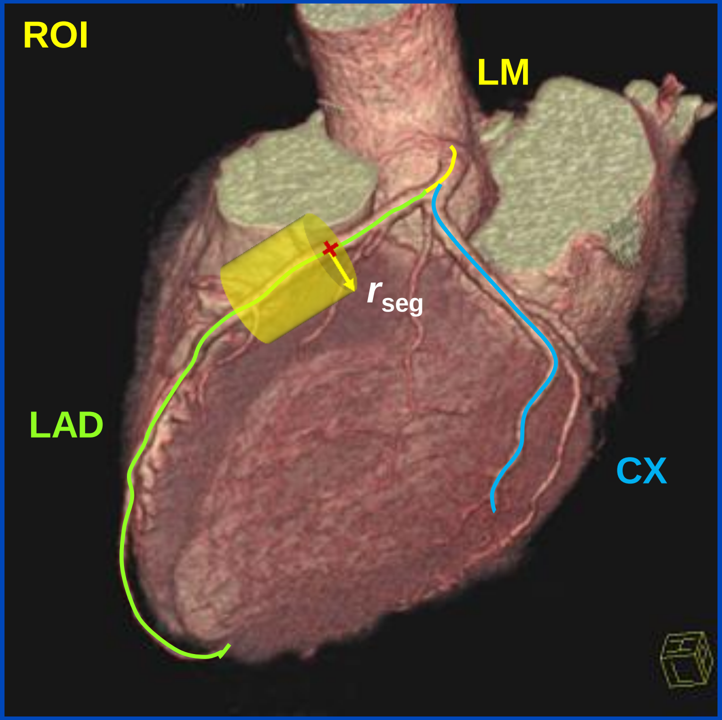
Motion model
Motion estimation
Motion compensation

↓

Final
reconstruction

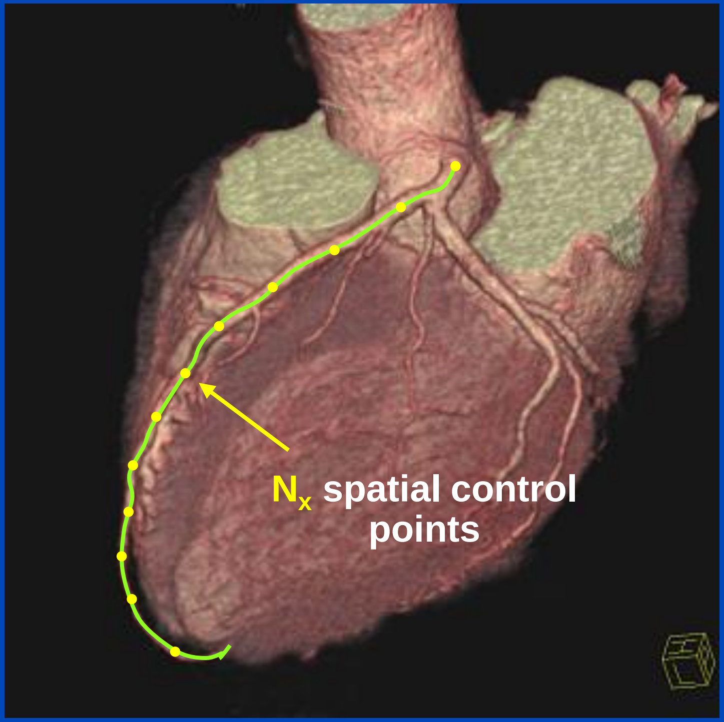
MAM

Initial Reconstruction and Segmentation



- Perform an initial short scan reconstruction of the complete volume.
- Segmentation of one of the main coronary artery branches using an in-house algorithm.
- Generation of a region of interest (ROI) Ω_s incorporating all motion artifacts associated to the chosen coronary artery.

MAM
Motion Model



- Motion inside Ω_s is modeled by a 4D motion vector field (MVF) sub-sampled in time and space, with N_t temporal and N_x spatial control points:

$$M(t, x, s) = x + s_{t,x}.$$

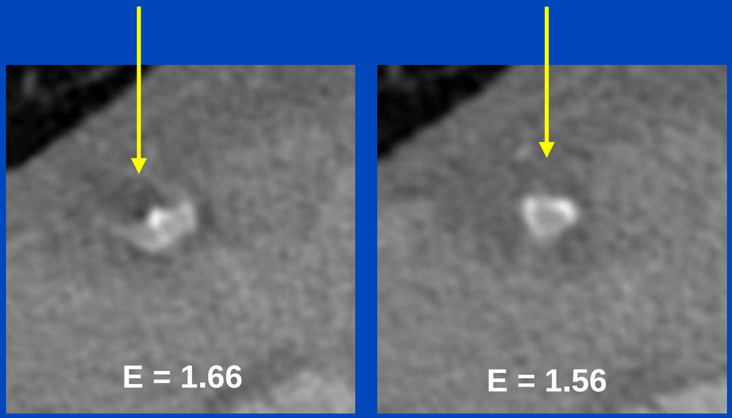
Parameter vector (↑ s) — Shift at time t and voxel x (↑ $s_{t,x}$)

- Between the control points, the MVF is approximated by spline interpolation and a dense MVF is generated.

MAM

Motion Estimation

“negative” values “non-negative” values



High entropy Low entropy

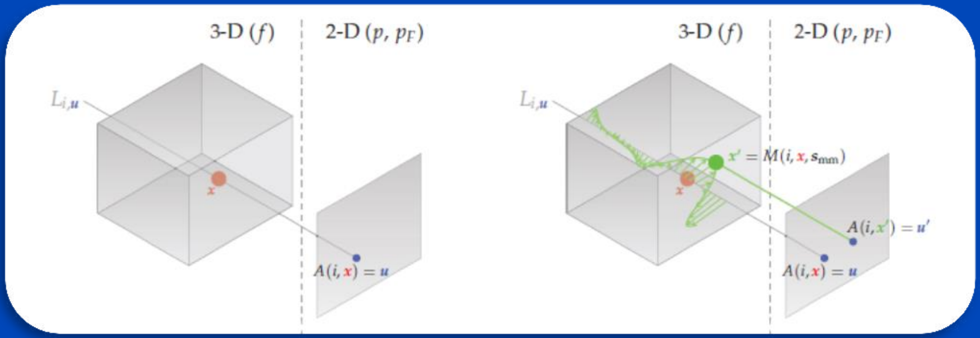
- The MVFs are subject to the cost function optimization:

$$\hat{s} = \arg\min_{\boldsymbol{s} \in \mathbb{R}^{N_p}} \mathcal{L}$$

- Entropy and Positivity are used as image artifact measuring cost functions and are optimized in an alternating manner using a gradient descent method.

MAM
Motion Compensation

- Motion-compensated reconstruction is obtained by taking the motion during the backprojection step into account (Schäfer's method[1]).



Backprojection along new trajectories

- The final reconstruction is created by replacing the segmented, motion-compensated region Ω_s in the original reconstruction.

[1] D. Schäfer et al. "Motion-compensated and gated cone beam filtered back-projection for 3-D rotational x-ray angiography," IEEE Trans. Med. Imaging 25(7), 898–906 (2006).

PAMoCo

- **Aim: Improve the image quality in the region of the coronary arteries for images featuring motion artifacts of different severity.**
- **Idea: Apply motion vectors on partial angle reconstructions to generate a fast reconstruction pipeline.**
- **Workflow:**

Initialization

+ Generation of partial angle reconstructions (PARs)

↓

Motion model
Motion estimation
Motion compensation

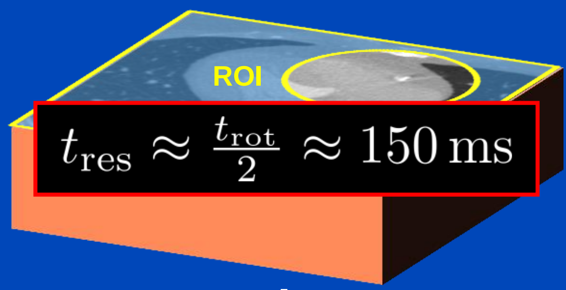
↓

Final reconstruction

PAMoCo

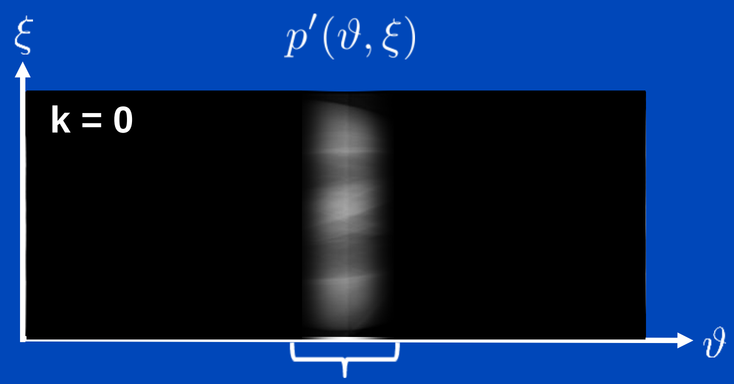
Generation of 2K+1 PARs

Initial segmented stack volume



$$t_{\text{res}} \approx \frac{t_{\text{rot}}}{2} \approx 150\,\text{ms}$$

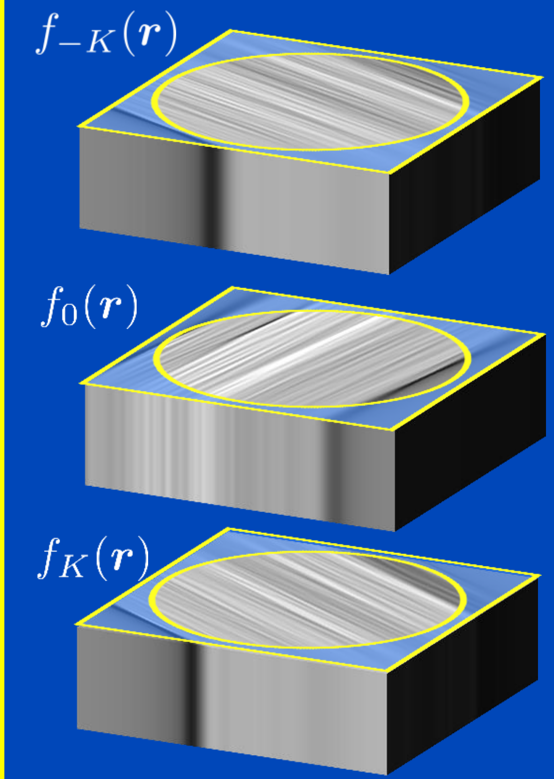
Subdivide the projection data $p'(\vartheta,\xi)$ into 2K + 1 overlapping sectors



$$p_k(\vartheta,\xi) = w_k(\vartheta)p'(\vartheta,\xi)$$

$$w_k(\vartheta) = \Lambda((\vartheta-\vartheta_k)/2\Delta\vartheta)$$

Partial angle reconstructions $f_k(\boldsymbol{r})$

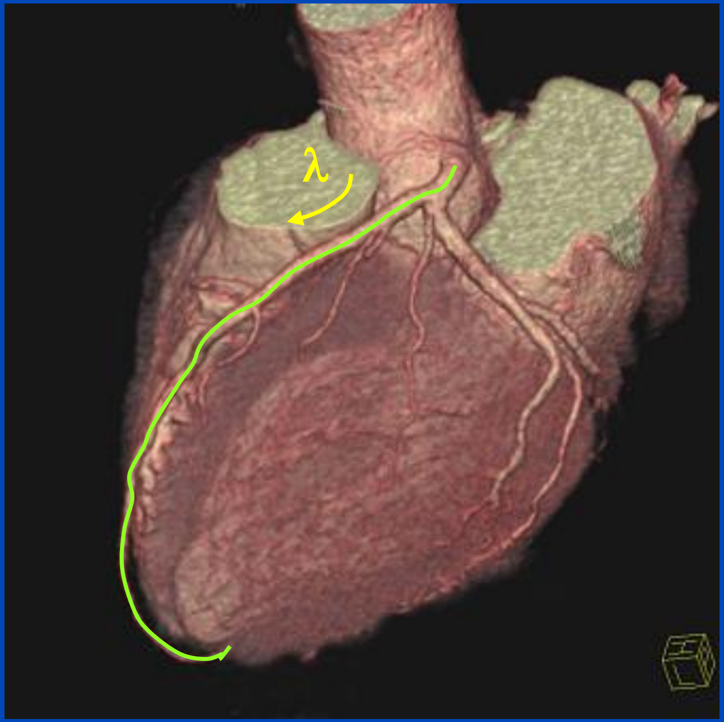


$$t_{\text{res}} \approx \frac{t_{\text{rot}}/2}{(2K+1)/2} \approx 10\,\text{ms}$$

FWHM = $\Delta\vartheta$ **K = 15**

PAMoCo

Motion Model



- Motion inside Ω_s is modeled by a 4D vector field

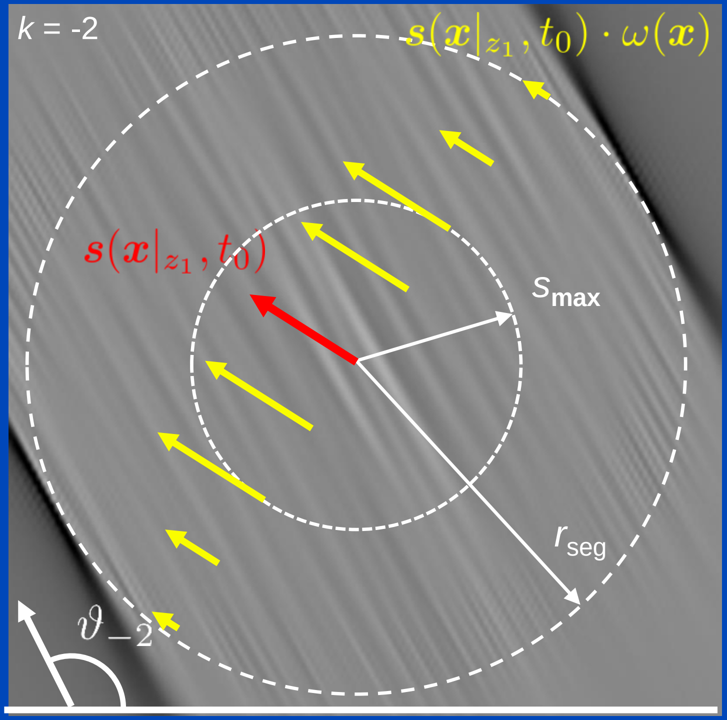
$$M(s, x, t) = x + d(s, x, t),$$

whose temporal and spatial dependence along the center line is modeled by two low degree polynomials (P,L ≤ 3)

$$d(s, \lambda, t) = \sum_{p=0}^{P} \sum_{l=0}^{L} s_{lp} (\lambda - \lambda_0)^l (t - t_0)^p.$$

- A dense MVF is generated by assigning a center line parameter λ to each voxel.

PAMoCo
Motion Model

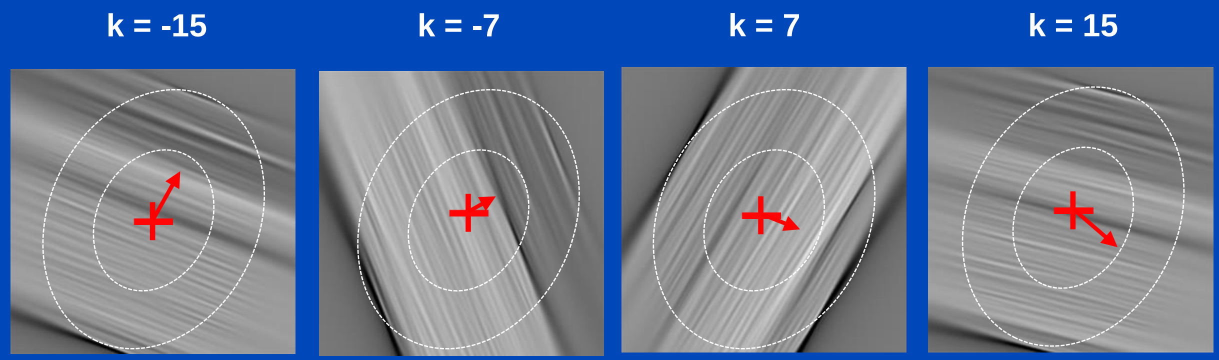


- **Furthermore, in order to enforce a smooth transition from the sub-volume, which is subject to MoCo and the original reconstruction a dense MVF is created, which drops to zero at the borders of the volume by introducing the weighting term**

$$\omega(x) \equiv \omega(|x - c(\lambda)|).$$

PAMoCo
Motion Compensation

- **Since the time frames of the PARs correspond to the different partial angle reconstructions, a motion-compensated reconstruction can be obtained by applying the MVF M on the PARs f_k**

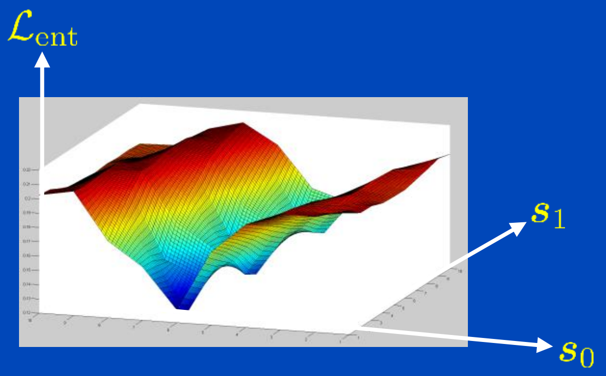


$$f_{\mathrm{MoCo}}(\boldsymbol{x}, \boldsymbol{s}) = \sum_{k=-K}^{K} f_k \left(\boldsymbol{M}(\boldsymbol{s}, \boldsymbol{x}, t_k) \right) .$$

and adding the warped images.

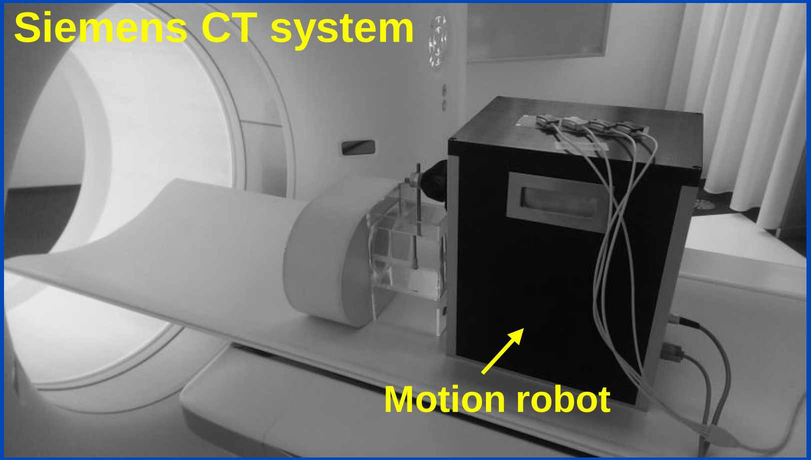
PAMoCo
Motion Estimation

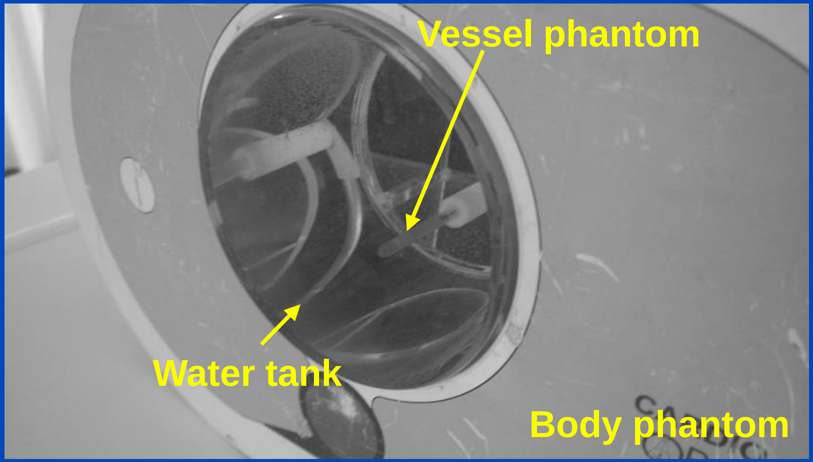
- As motion artifact measuring cost functions the images entropy and the entropy of the absolute gradient image are chosen.
- The motion estimation routine is separated into two parts:
- **Step 1: Brute force search**
  - Crude scan of the parameter space assuming a linear motion pattern using the images entropy as cost function.



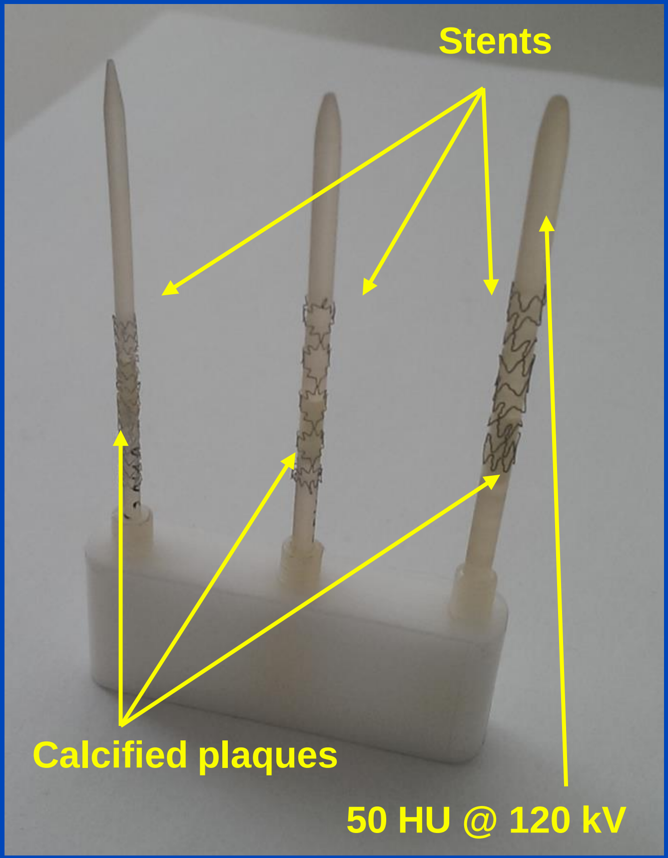
- **Step 2: Optimization**
  - Since the cost function is non-convex, the optimization is re-initialized multiple times at local minima found in step 1.
  - For optimization an implementation of the gradient-free Powell's algorithm is used.

Phantom Measurement

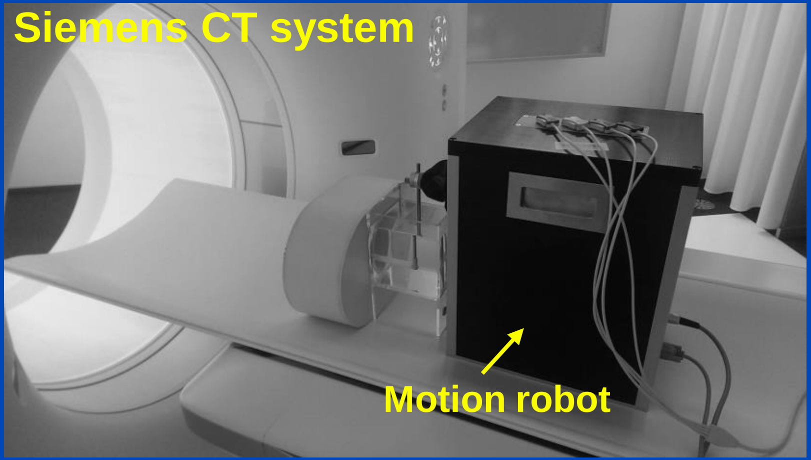


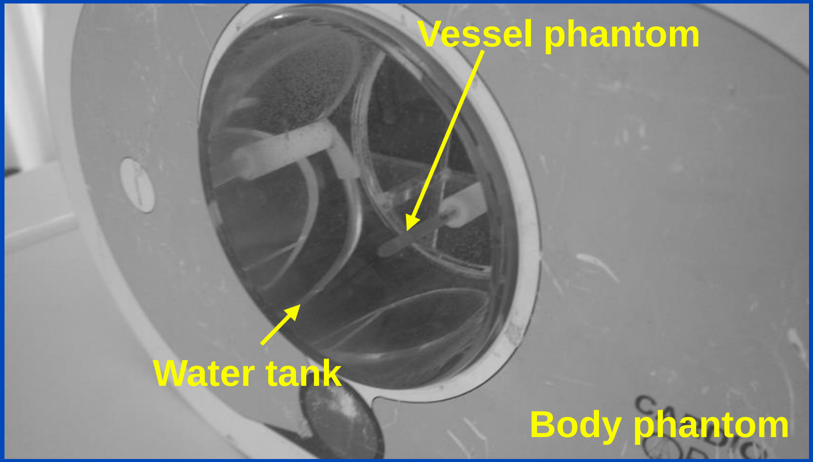


d = 1.5 mm 2.5 mm 3 mm

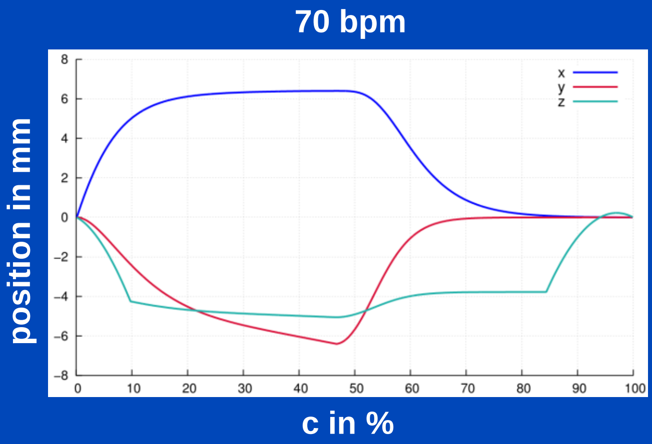


Phantom Measurement





Applied motion

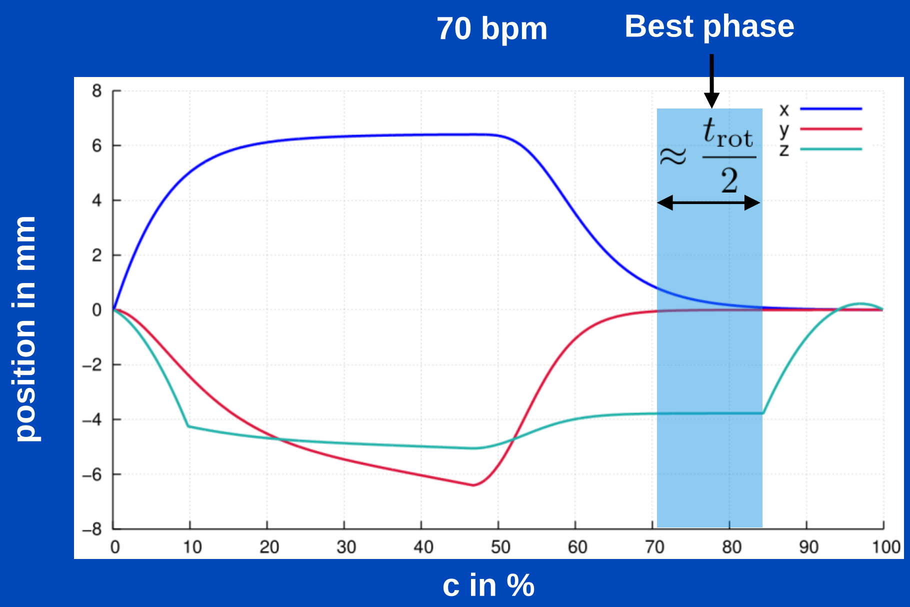


Data acquisition

- Rotation Time t_{rot} = 250 ms
  → t_{res}(FBP) ≈ 125 ms
- Low pitch spiral scanning: p ≈ 0.2
- → Reconstruction of multiple cardiac phases possible.

Phantom Measurement

- We characterize the best phase, by the simulated phase featuring least absolute motion:

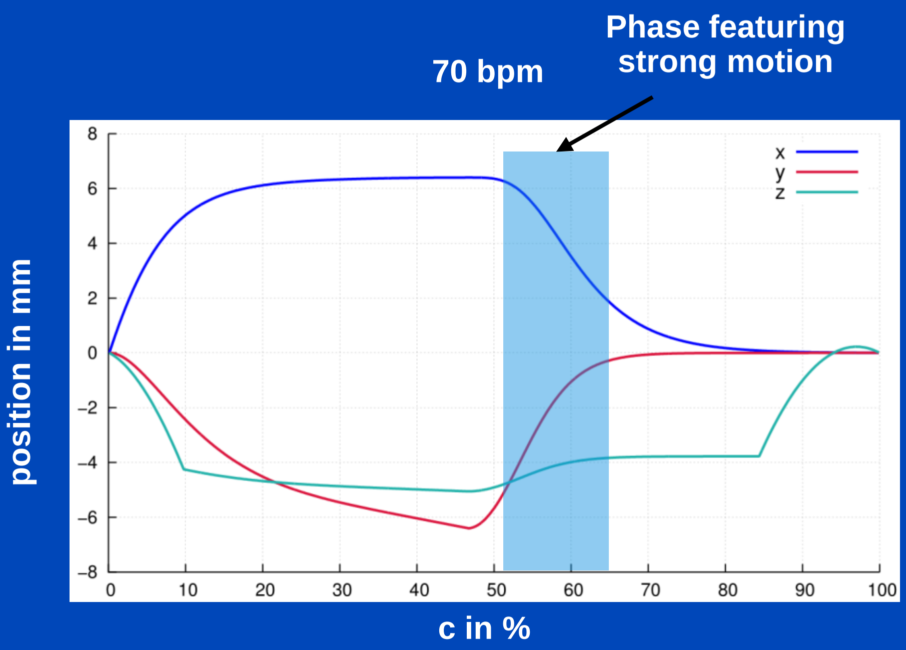


- MAM and TRIM aim at increasing the image quality close to the best phase.

- → Perform reconstructions at slightly shifted cardiac phases.

Phantom Measurement

- We characterize the best phase, by the simulated phase featuring least absolute motion:



- MAM and TRIM aim at increasing the image quality close to the best phase.

- → Perform reconstructions at slightly shifted cardiac phases.

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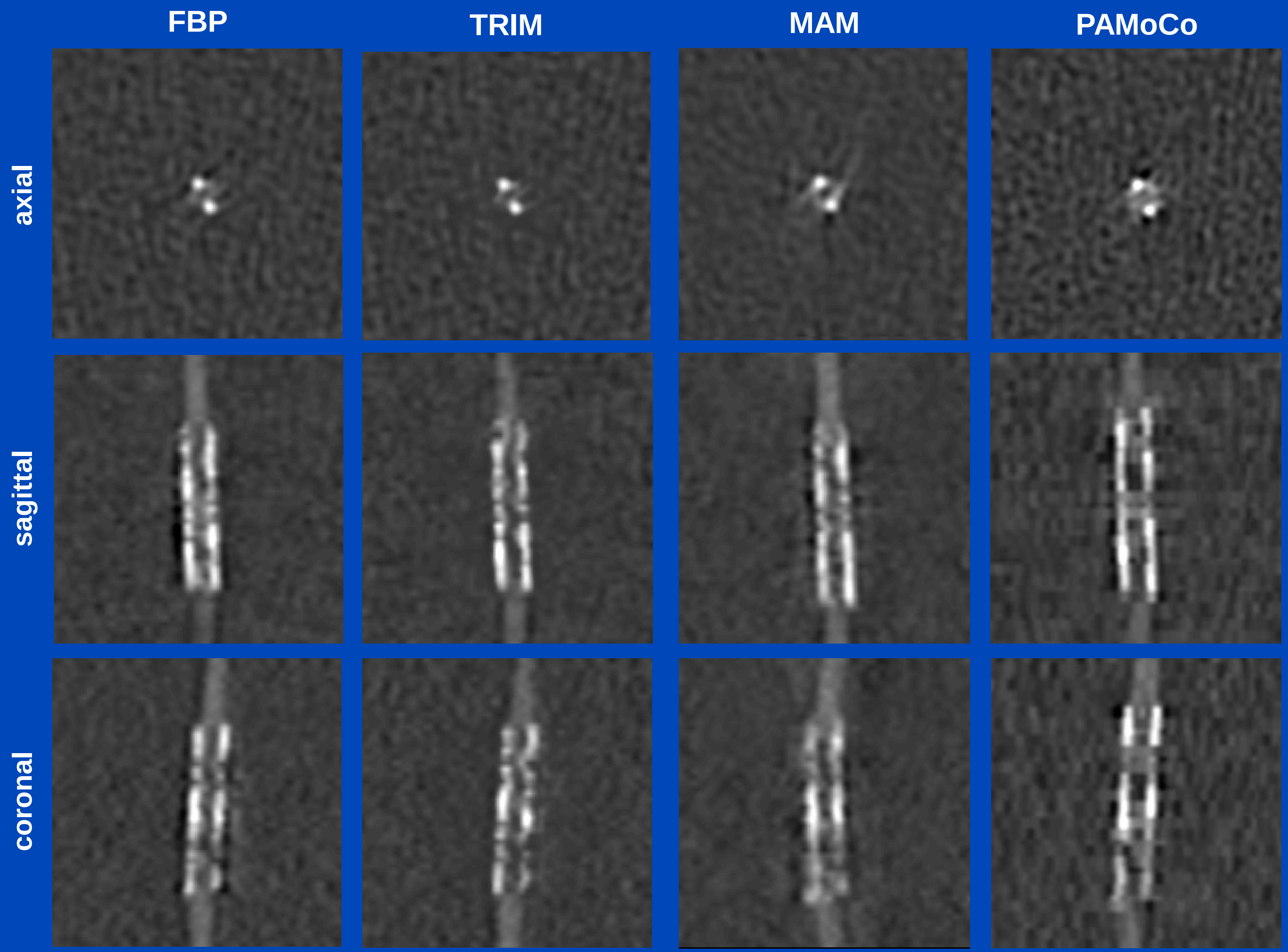
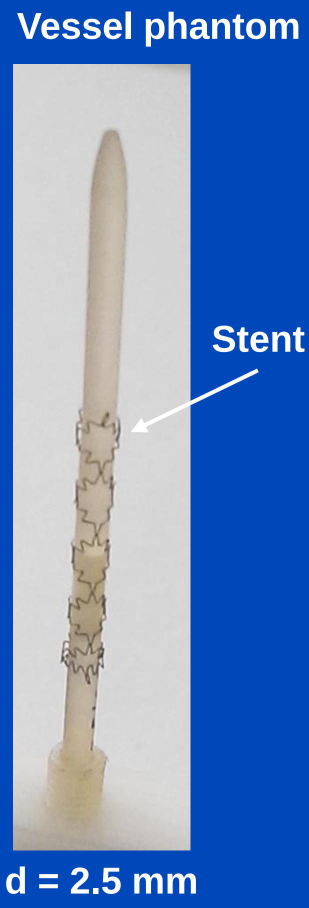
Results
Best Phase





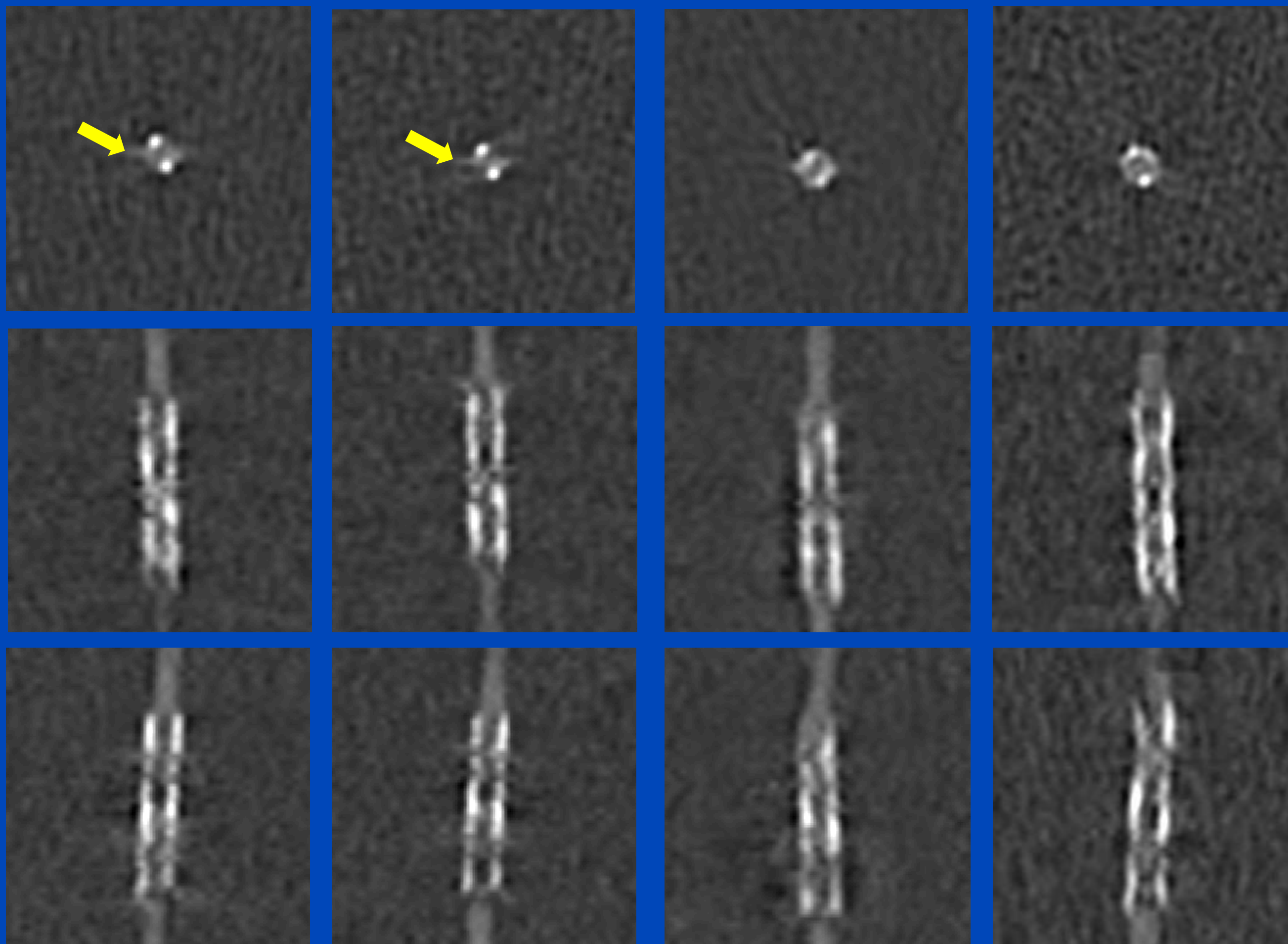
$\overline{HR}$ = 70 bpm, c = 72%,
C = 400 HU, W = 1500 HU

Results

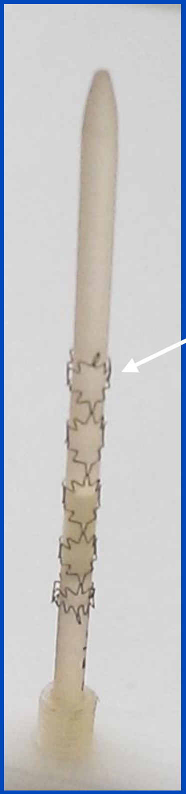
5% off Best Phase

FBP TRIM MAM PAMoCo

axial, sagittal, coronal



Vessel phantom

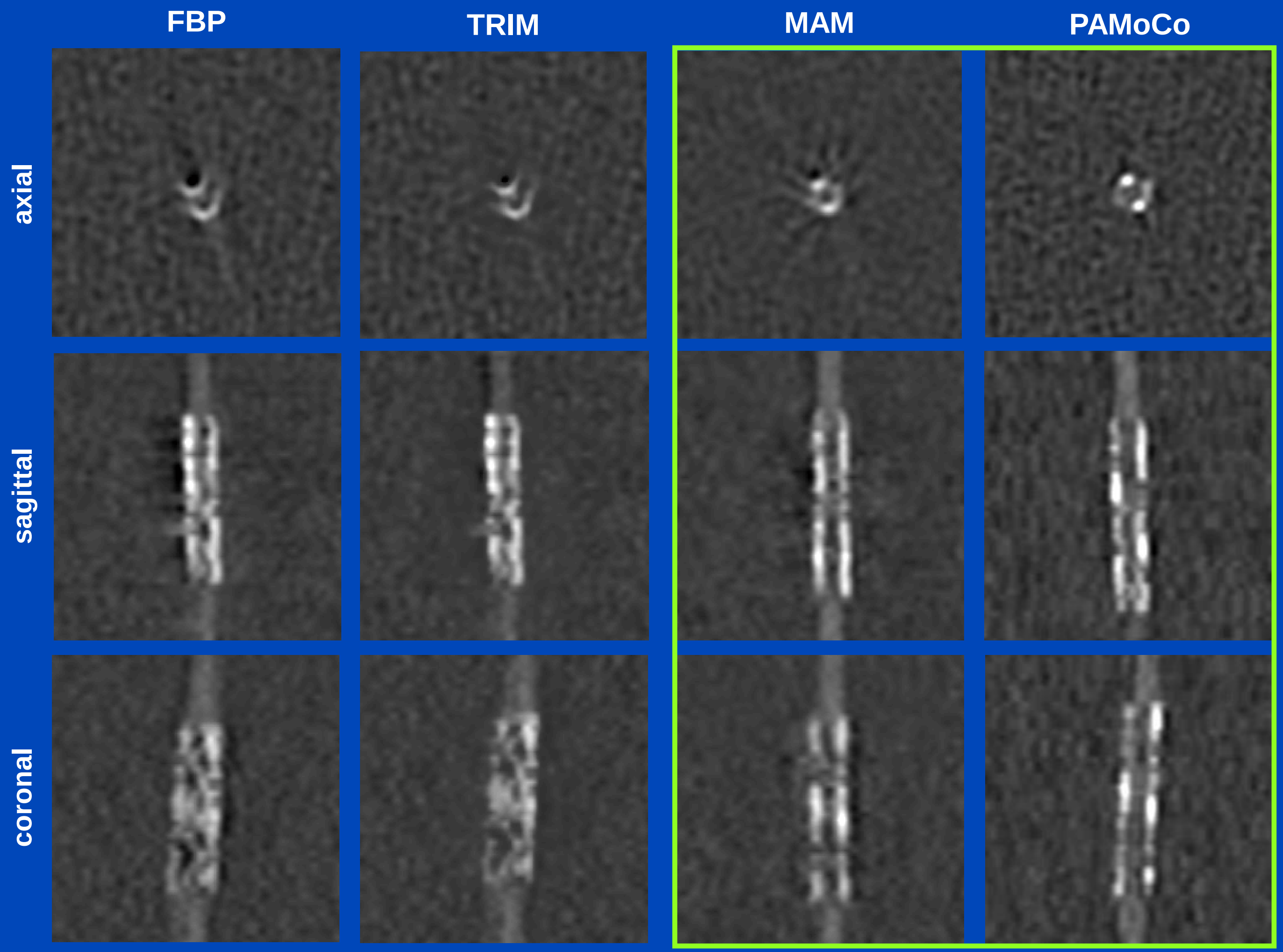


Stent

d = 2.5 mm

$\overline{HR}$ = 70 bpm, c = 67%,
C = 400 HU, W = 1500 HU

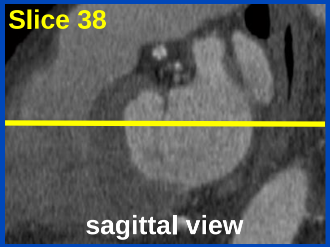
Results
10% off Best Phase



Vessel phantom

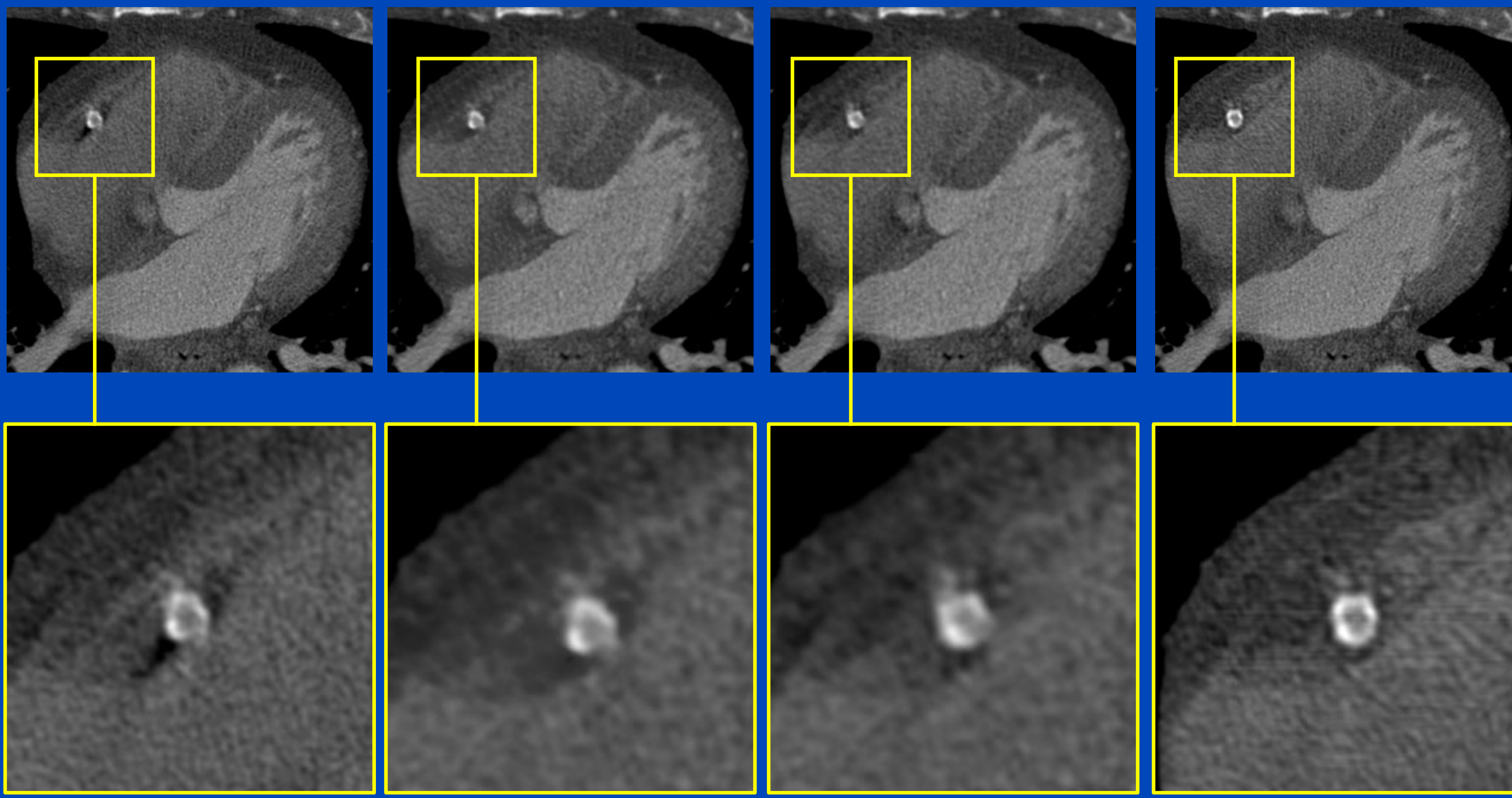
d = 2.5 mm

$\overline{HR}$ = 70 bpm, c = 62%,
C = 400 HU, W = 1500 HU



Clinical Case

FBP | TRIM | MAM | PAMoCo



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$\overline{HR}$ = 74 bpm, c = 74%,
C = 400 HU, W = 1500 HU

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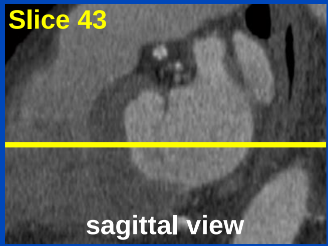
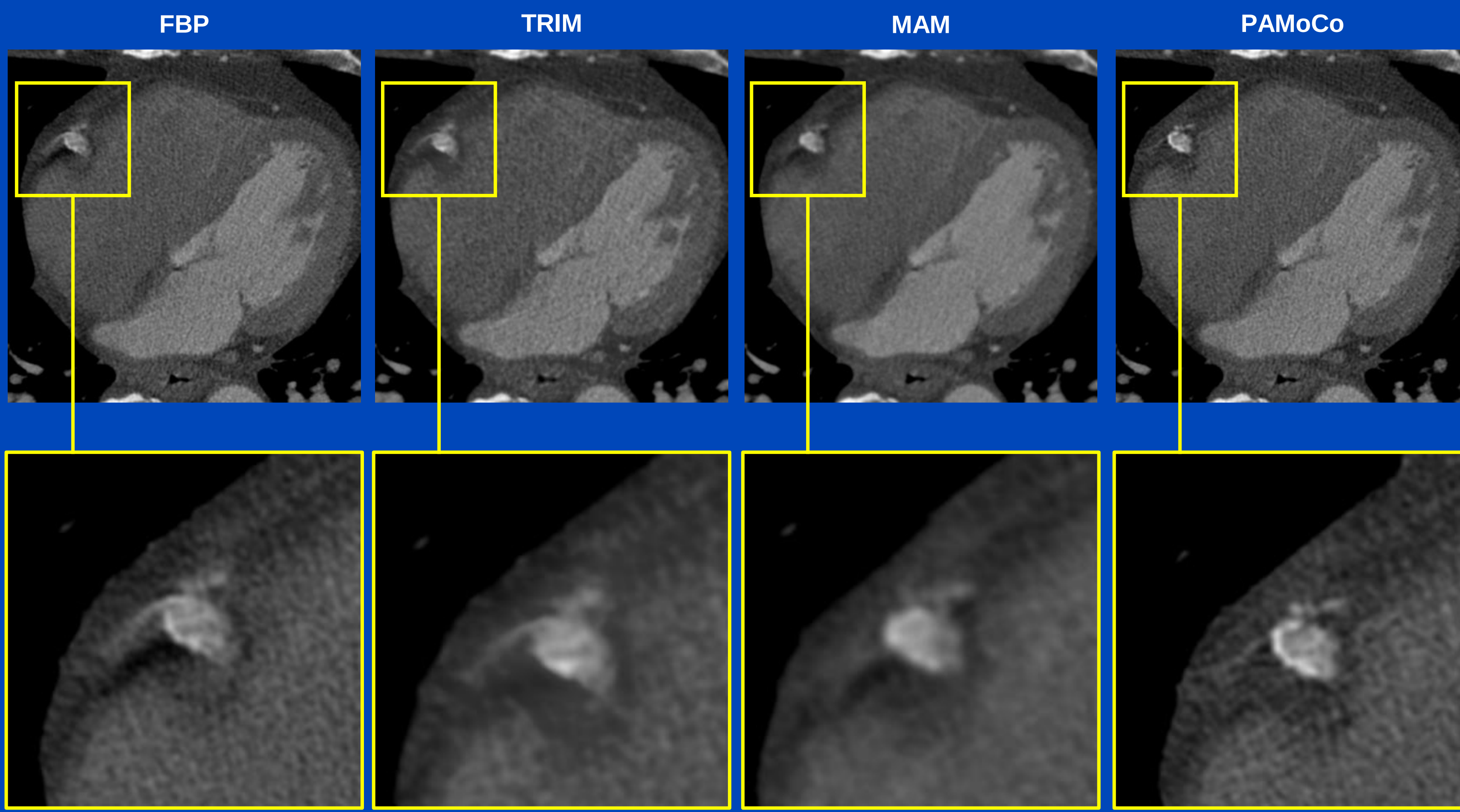


Clinical Case



$\overline{HR}$ = 74 bpm, c = 74%,
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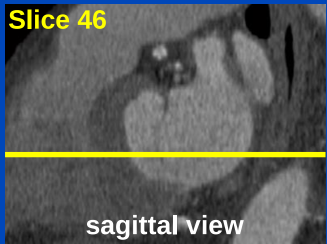
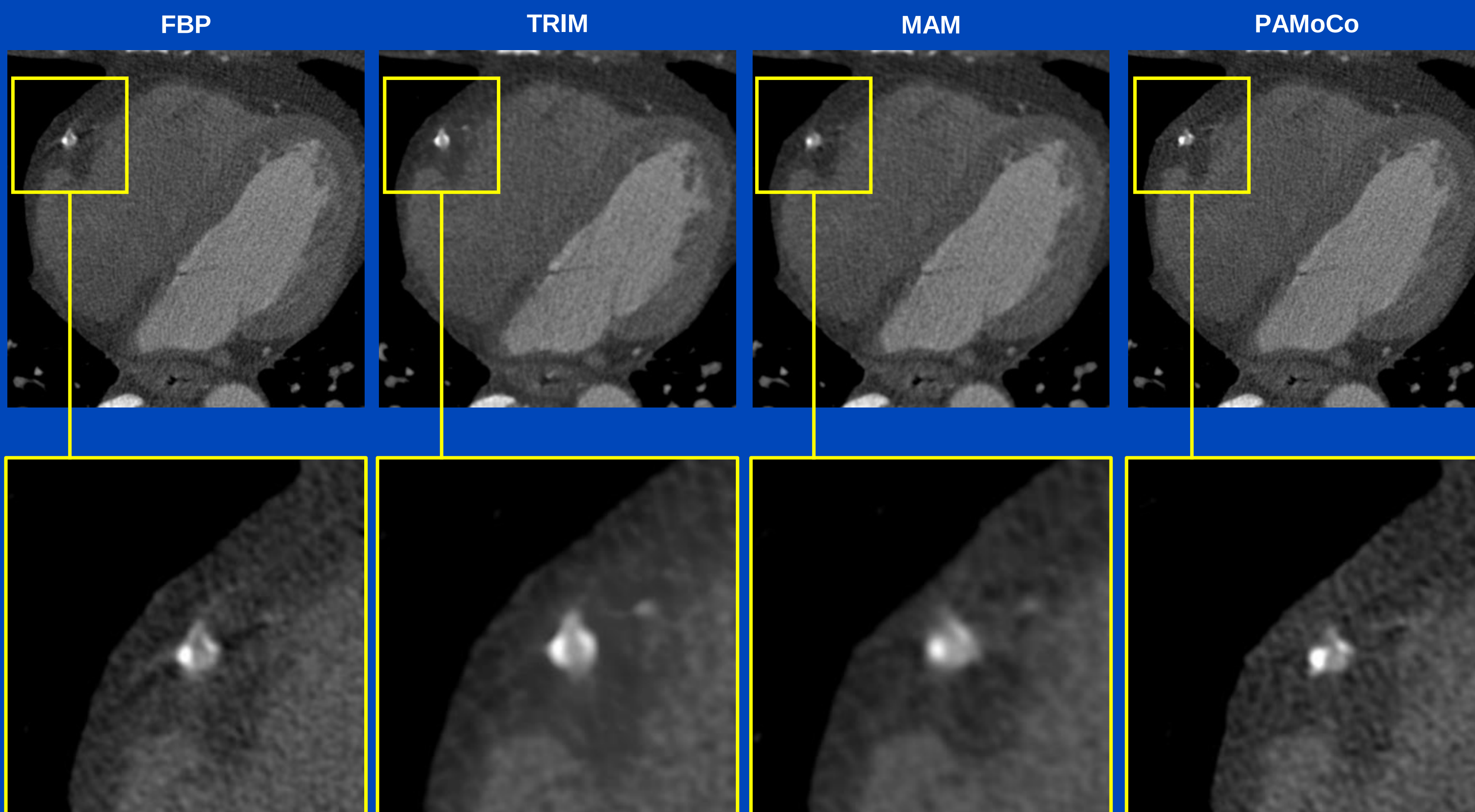
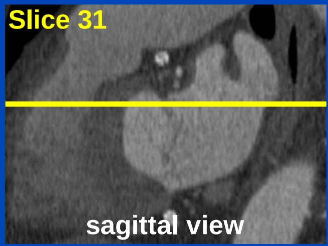


Clinical Case

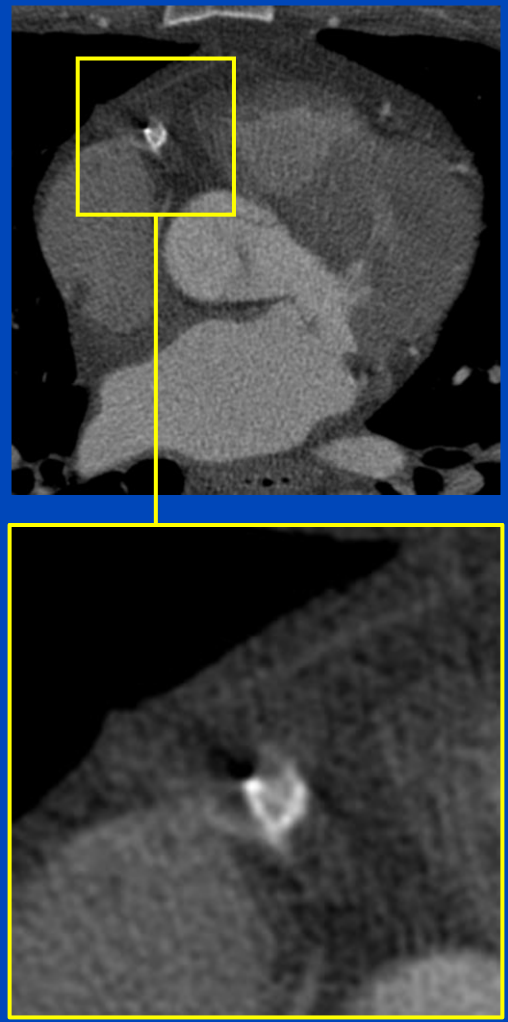


$\overline{\mathrm{HR}}$ = 74 bpm, c = 74%,
C = 400 HU, W = 1500 HU

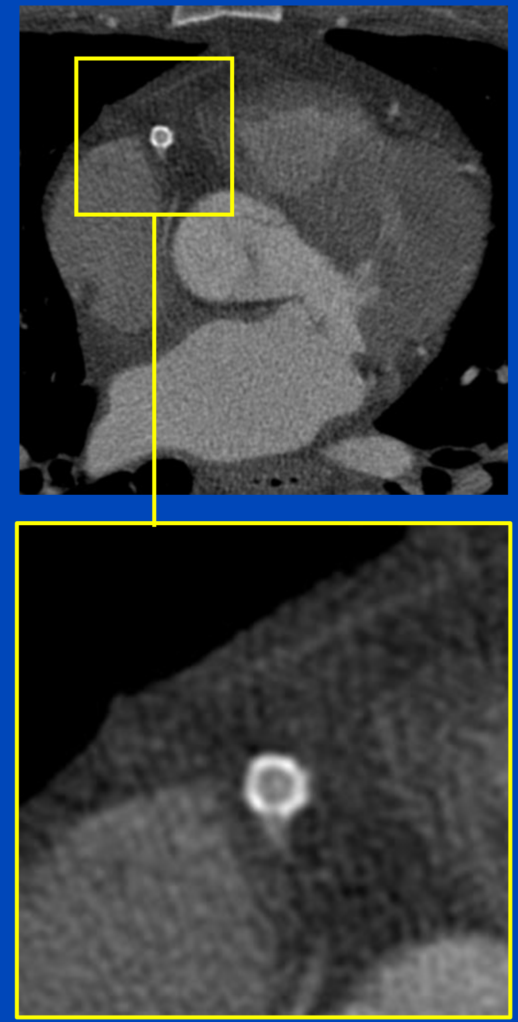


Clinical Case

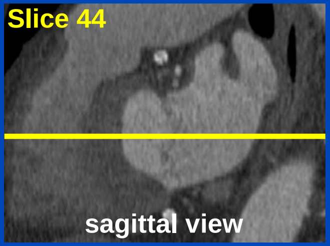
FBP



PAMoCo

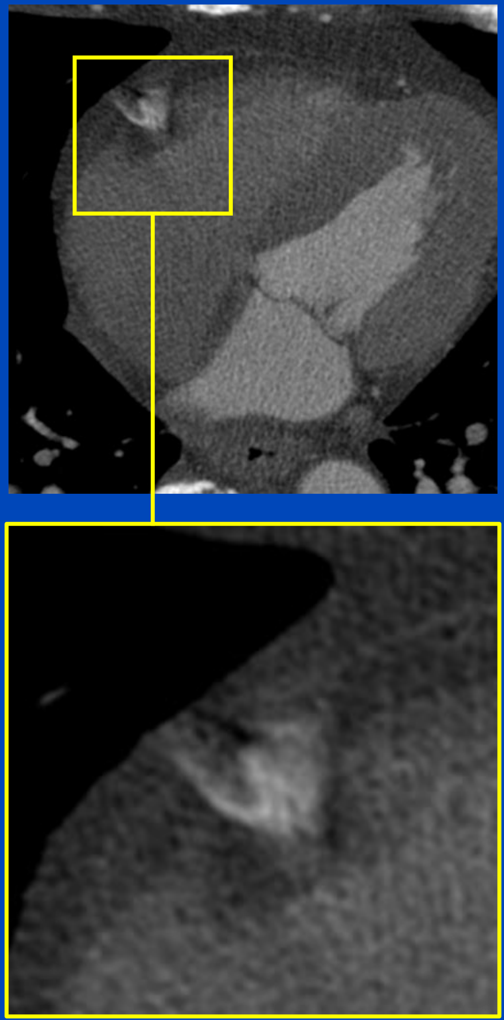


$\overline{\mathrm{HR}}$ = 74 bpm, c = 30%,
C = 400 HU, W = 1500 HU

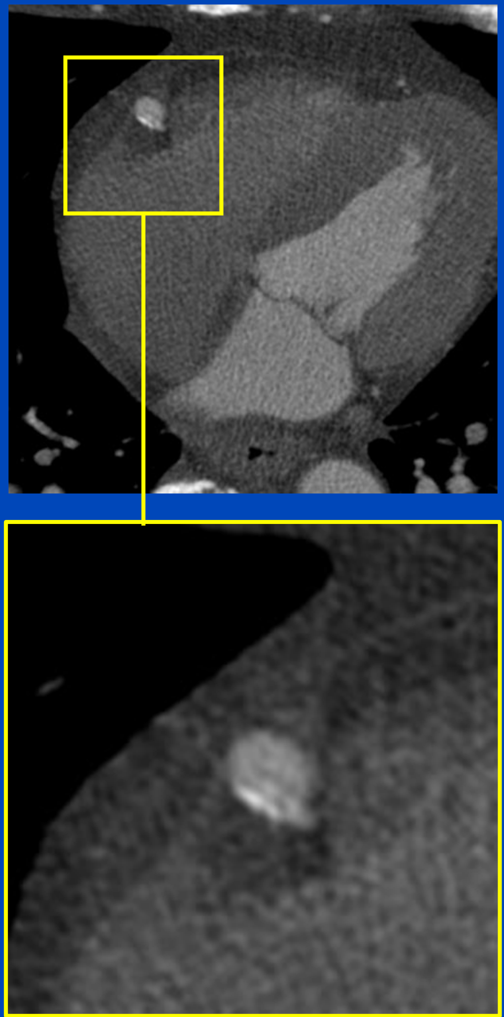


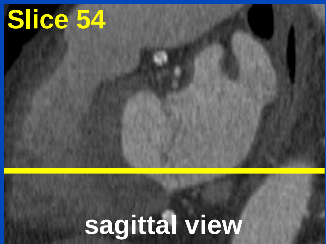
Clinical Case

FBP



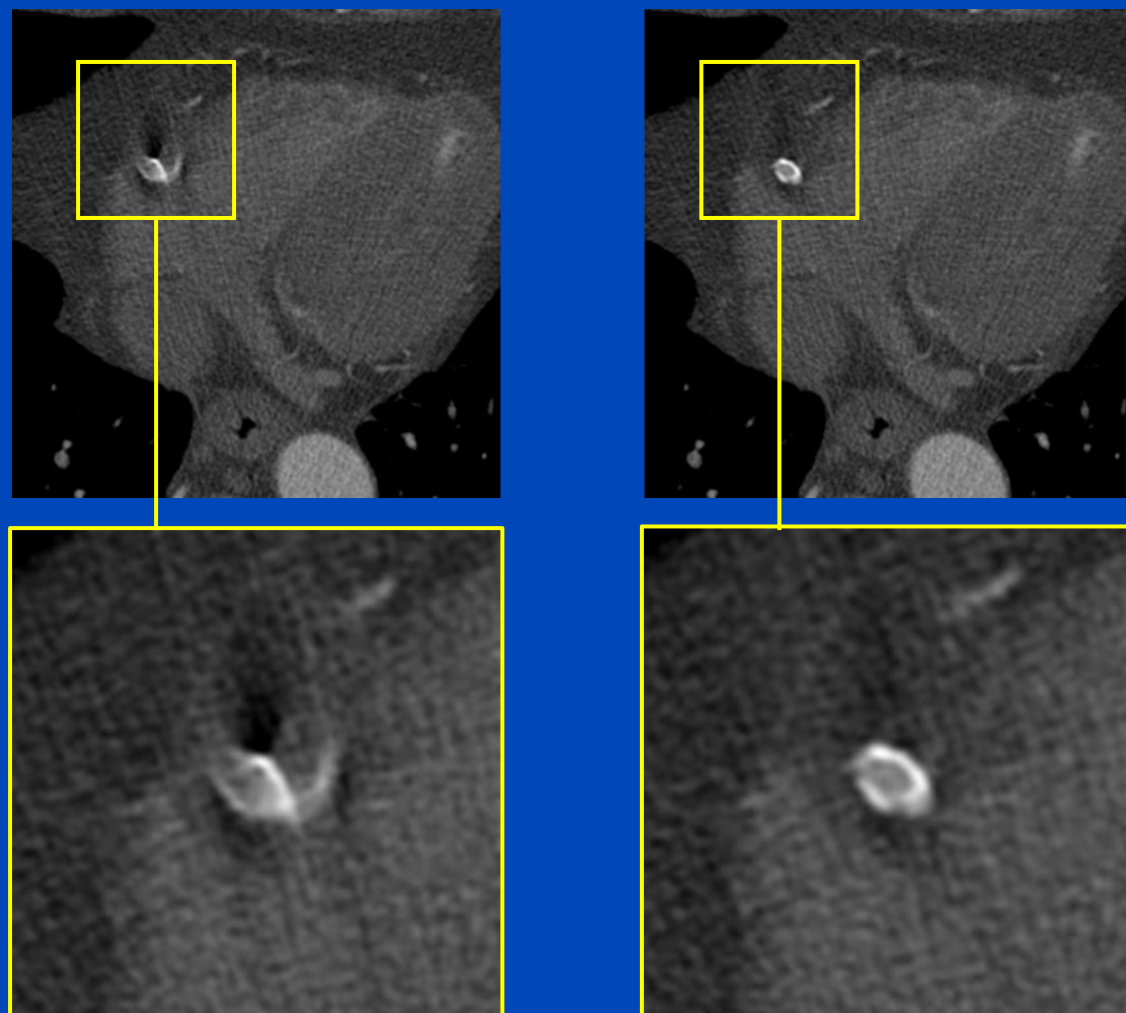
PAMoCo





Clinical Case

FBP | PAMoCo



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$\overline{HR}$ = 74 bpm, c = 30%,
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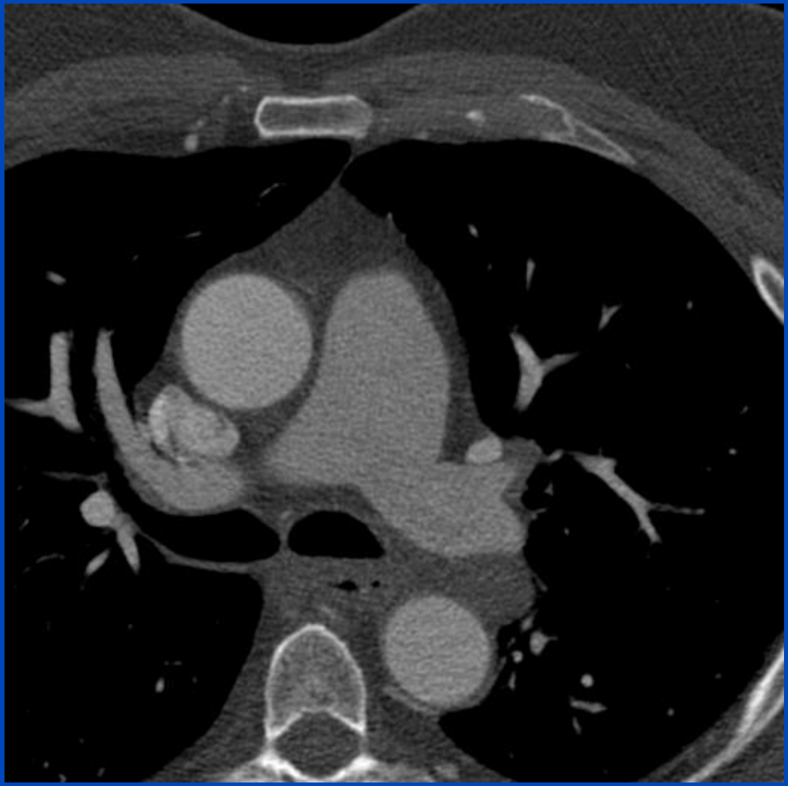
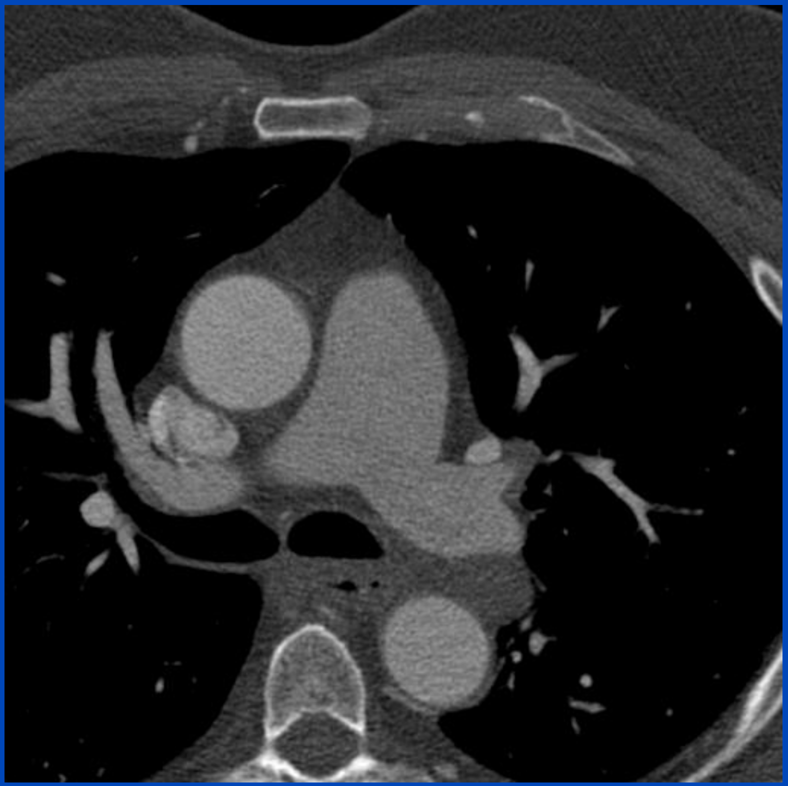
Clinical Case

FBP



PAMoCo



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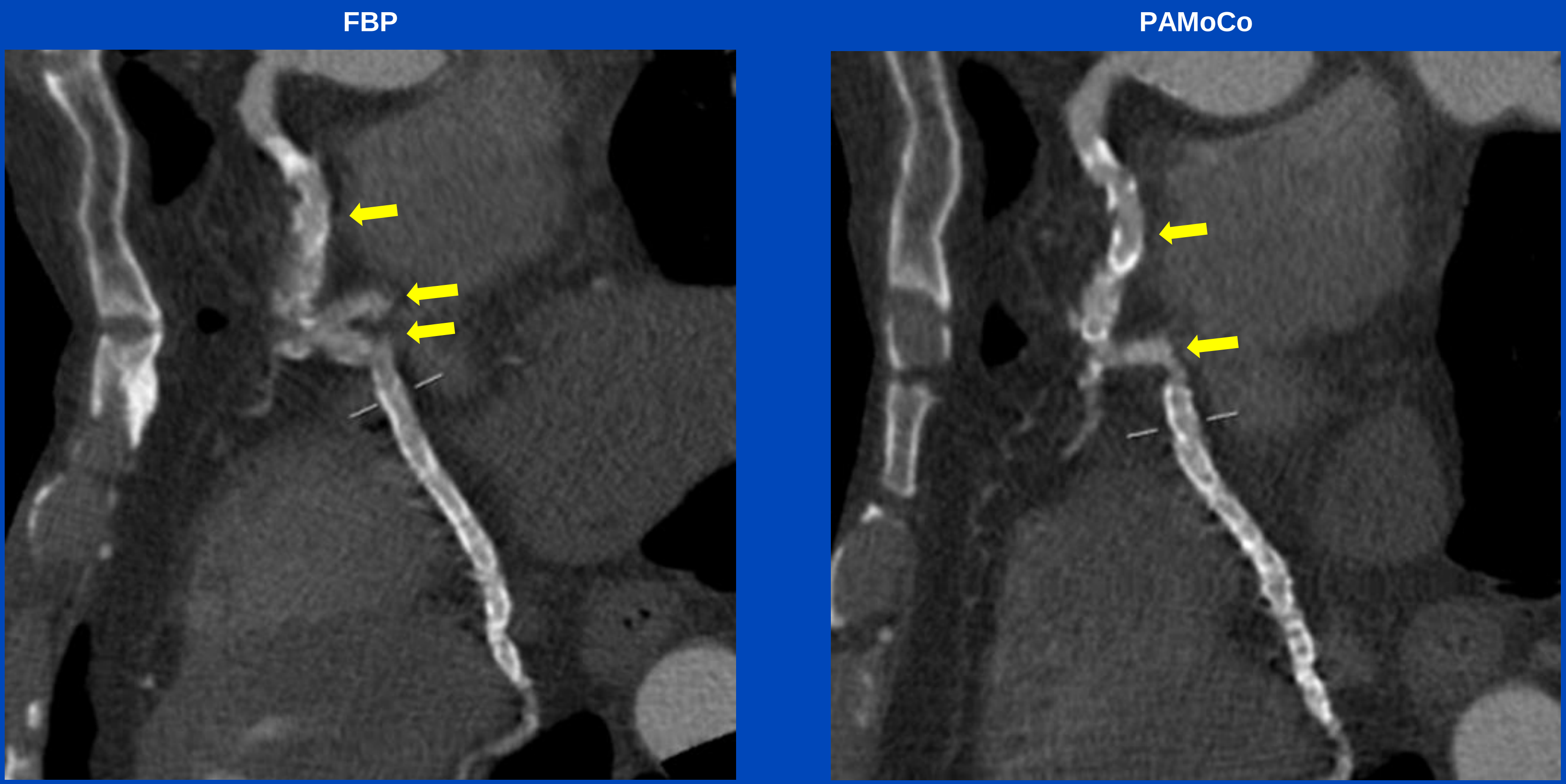
Clinical Case



curved MPRs of the RCA

Summary and Conclusion

- We see an increased sharpness of the coronary arteries in case of the phantom and the real data in cardiac phases close to the best phase in case of all three algorithms.
- Stepping further away from the best phase, the MoCo algorithms are able to enhance the temporal resolution beyond the TRIM limit.
- The PAMoCo algorithm is also able to correct for slightly more severe motion artifacts than MAM due to its global optimization character.
- Hence we conclude

FBP < TRIM < MAM < PAMoCo

in their ability of increasing the temporal resolution.

Thank You!

This study was supported by Siemens Healthcare GmbH.

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