Reduction of Motion Artifacts in Cardiac CT Based on Partial Angle Reconstructions from Short Scan Data

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Motivation

- Cardiac CT imaging is routinely practiced for the diagnosis of cardiovascular diseases like coronary artery disease.
- The imaging of small and fast moving vessels places high demands on the spatial and temporal resolution of the reconstruction.
- Insufficient temporal resolution leads to motion artifacts, whose occurrence might require a second scan increasing the dose applied to the patient.

Temporal Resolution in Cardiac CT

- For the right coronary artery (RCA) mean velocities varying between 35 mm/s and 70 mm/s have been measured.[1,2,3,4)]
- Assume a constant mean velocity of 50 mm/s during scan

	Single Source	Dual Source
t_{rot}	250 ms	250 ms
t_{res}	125 ms	63 ms
Displacement	6.2 mm	3.1 mm

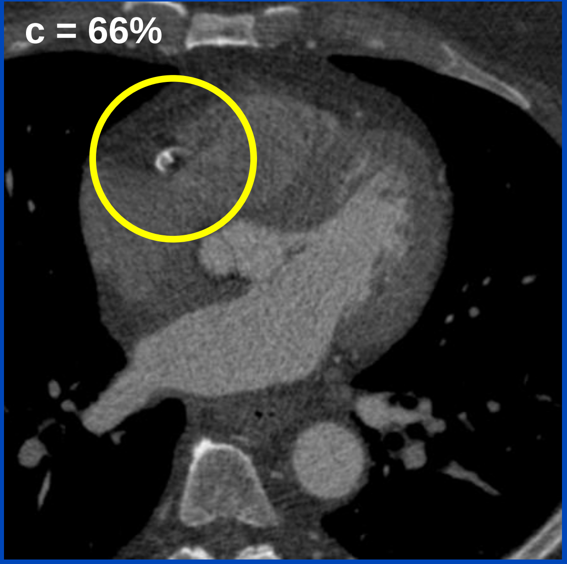
- Large displacement for an object of ~ 1-5 mm diameter.

→ Occurrence of strong motion artifacts especially in case of single source systems!

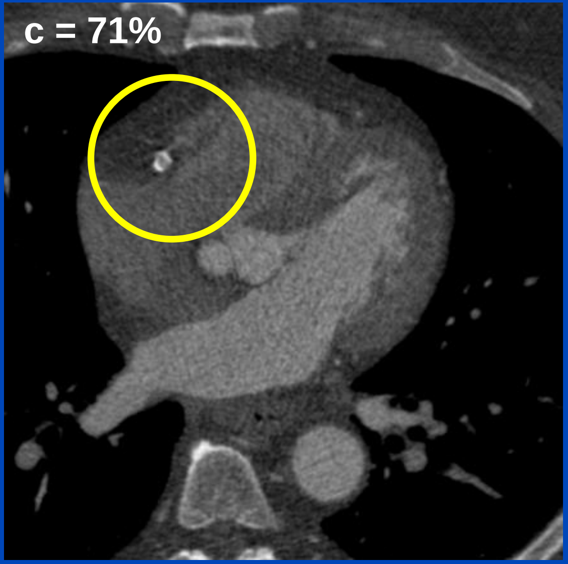
[1)]Husmann et al. Coronary Artery Motion and Cardiac Phases: Dependency on Heart Rate - Implications for CT Image Reconstruction. Radiology, Vol. 245, Nov 2007.
[2]Shechter et al. Displacement and Velocity of the Coronary Arteries: Cardiac and Respiratory Motion. IEEE Trans Med Imaging, 25(3): 369-375, 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.

Aim

- Increase the temporal resolution in cardiac CT in the region of the coronary arteries for data acquired with single source systems.
- Especially beneficial in cases of patients with high or irregular heart rates or non-optimally chosen gating positions.
- In view of dose optimized scan protocols, we want to utilize only the data needed for a single short scan reconstruction.



Non-optimally chosen gating position



“Best phase”

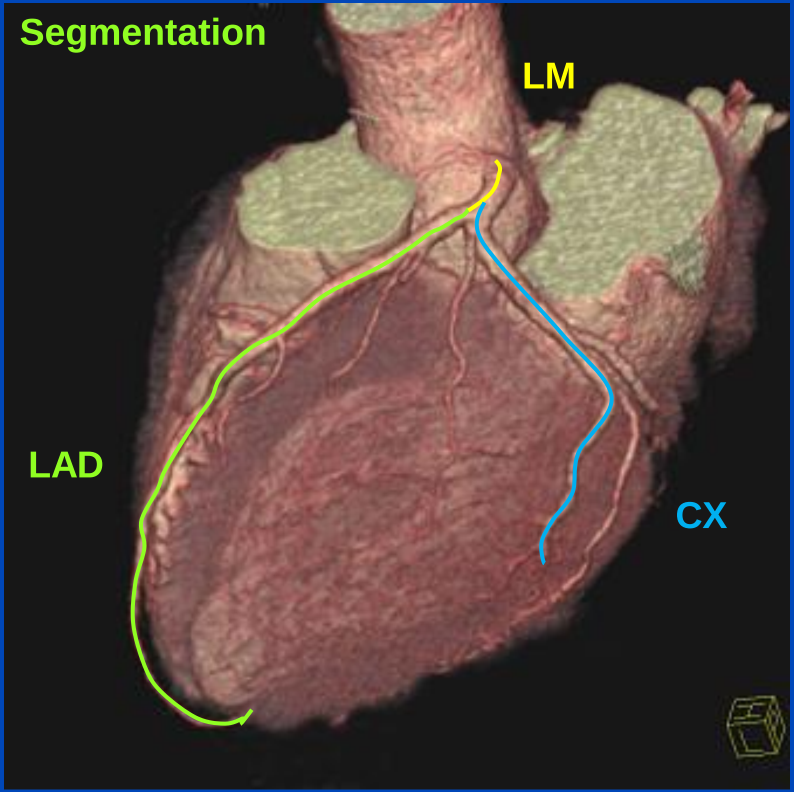
C = 300 HU; *W* = 1500 HU

PAMoCo
Workflow

- Reconstruction and segmentation of sub-volumes from a phase-correlated data-set
- Generation of 2K+1 partial angle reconstructions (PARs)
- Motion compensation based on PARs (PAMoCo)
  - Motion model
  - Cost function optimization

PAMoCo Step 1

Initial Reconstruction and Segmentation

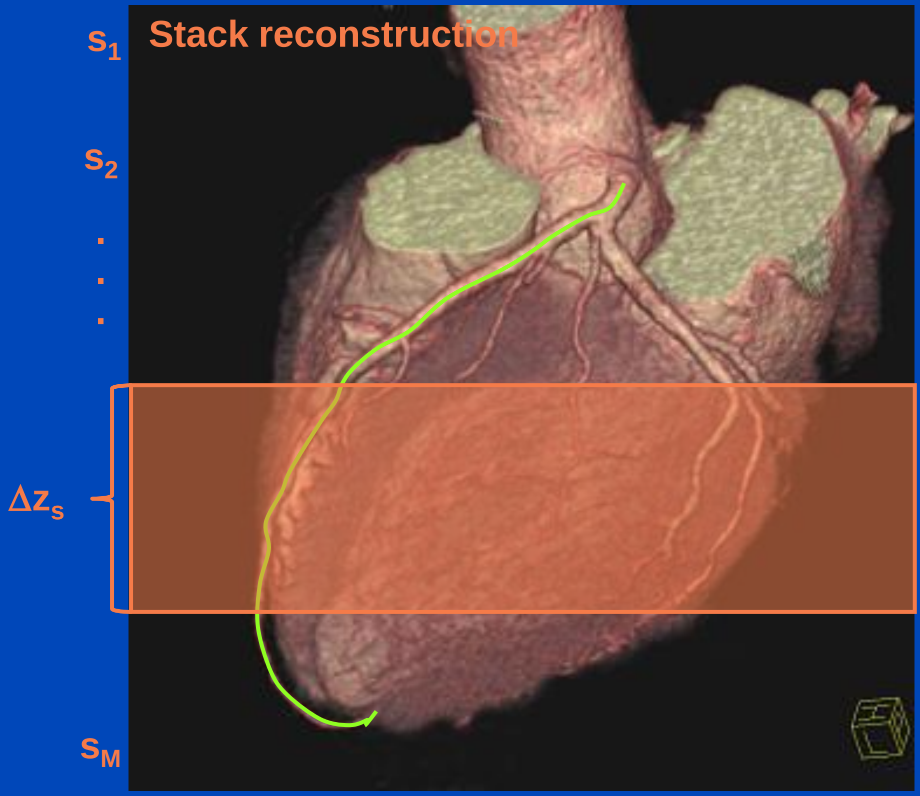


Data courtesy of Dr. Stephan Achenbach

- **Perform an initial short scan reconstruction of the complete volume.**
- **Segmentation of one of the main coronary artery (CA) branches (RCA, LM, LAD, CX) by an in-house algorithm.**
- **In case of spiral scan and sequential scans a discontinuity of the time coordinate ϑ in the z-direction is implied.**

PAMoCo Step 2

Reconstruction of Stacks

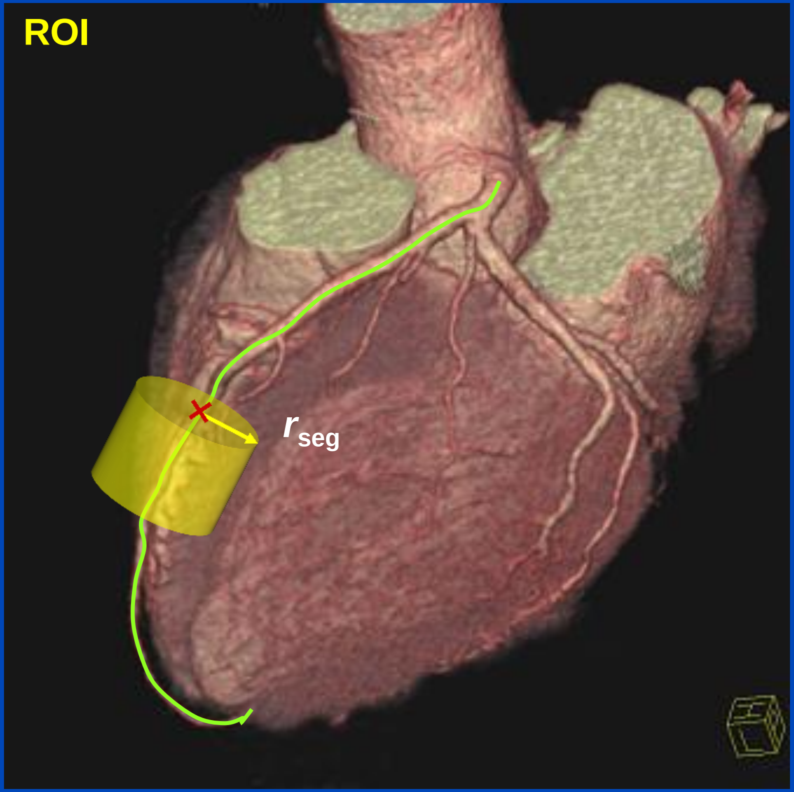


Data courtesy of Dr. Stephan Achenbach

- **We subdivide the volume into several overlapping stacks, whose extent Δz_s and quantity M depends on the detector size.**
- **For the reconstruction of each stack only short scan data acquired during one heart beat are used.**
- **Each stack is processed independently.**

PAMoCo Step 3

Stack Segmentation



Data courtesy of Dr. Stephan Achenbach

- For each stack, a region of interest (ROI) Ω_{seg} is defined by creating a tube of radius r_{seg} around the segmented centerline.
- This region should incorporate all motion artifacts caused by the motion of the CAs.
- We estimated r_{seg} with the help of coronary artery velocity measurements[1)]: $v_{max} \approx 100$ mm/s

$$\rightarrow r_{\text{seg}} = 2d_{\text{max}} = 2(v_{\text{max}} t_{\text{rot}}/2)$$

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[1)]Vembar et al. A dynamic approach to identifying desired physiological phases for cardiac imaging using multislice spiral CT. Med. Phys. 30, Jul 2003.

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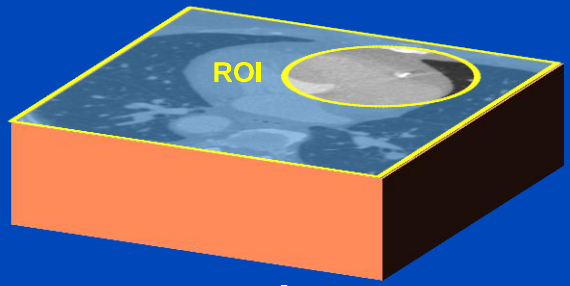
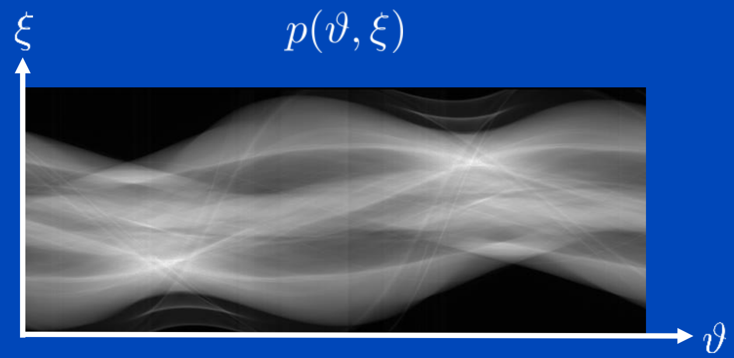
PAMoCo Step 4

Create 2K+1 Partial Angle Reconstructions (PARs)

Initial segmented stack volume



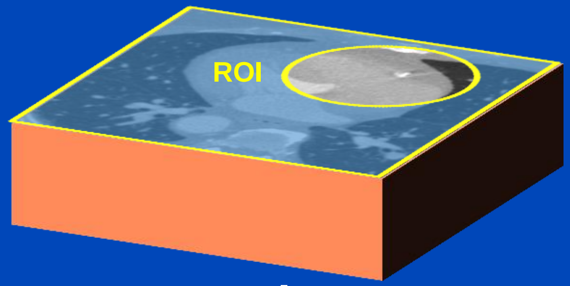
Subdivide the projection data $p(\vartheta, \xi)$ into 2K + 1 overlapping sectors



PAMoCo Step 4

Create 2K+1 Partial Angle Reconstructions (PARs)

Initial segmented stack volume



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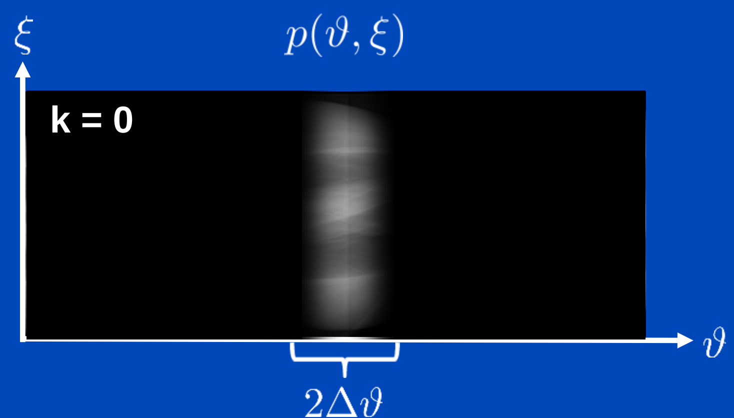


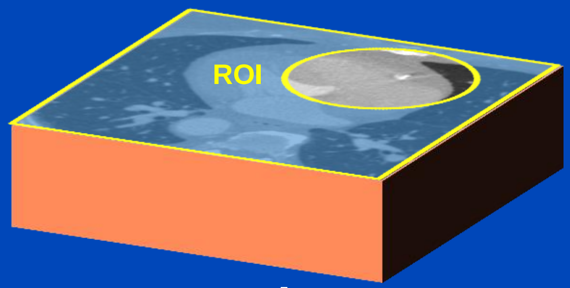
$$p_k(\vartheta, \xi) = w_k(\vartheta) p(\vartheta, \xi)$$

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

PAMoCo Step 4

Create 2K+1 Partial Angle Reconstructions (PARs)

Initial segmented stack volume



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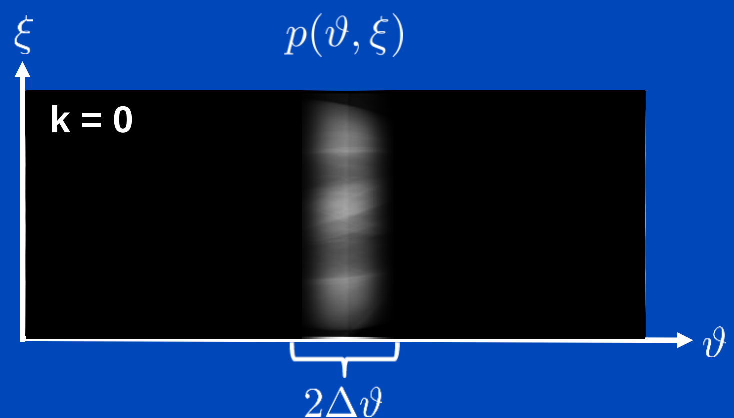

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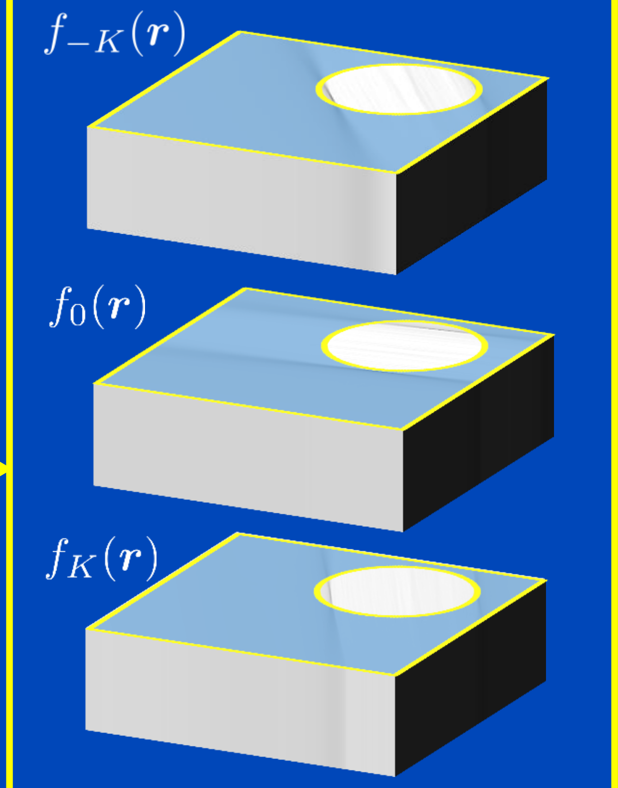
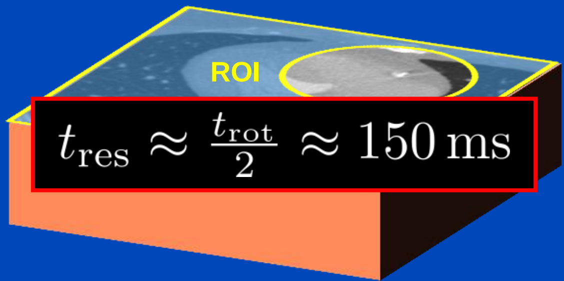
Partial angle reconstructions $f_k(\boldsymbol{r})$



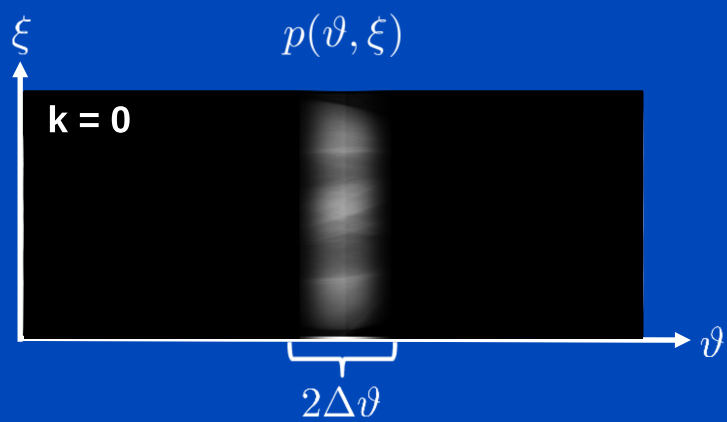
PAMoCo Step 4

Create 2K+1 Partial Angle Reconstructions (PARs)

Initial segmented stack volume



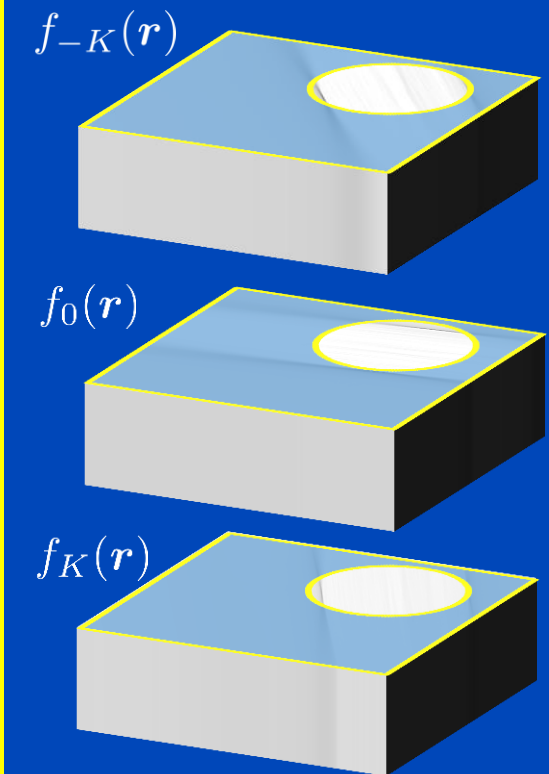
Subdivide the projection data $p(\vartheta, \xi)$ into 2K + 1 overlapping sectors



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

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

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



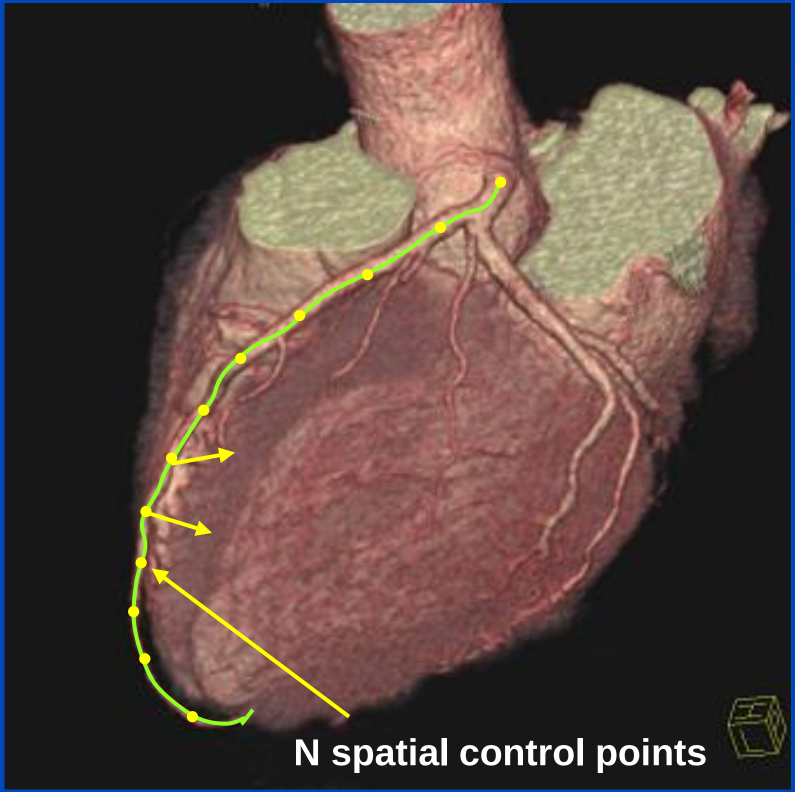
$$t_{\text{res}} \approx \frac{t_{\text{rot}}/2}{(2K+1)/2} \approx 10\,\text{ms}$$

FWHM = $\Delta\vartheta$

K = 15

Algorithmic Concept

Motion Model



Data courtesy of Dr. Stephan Achenbach

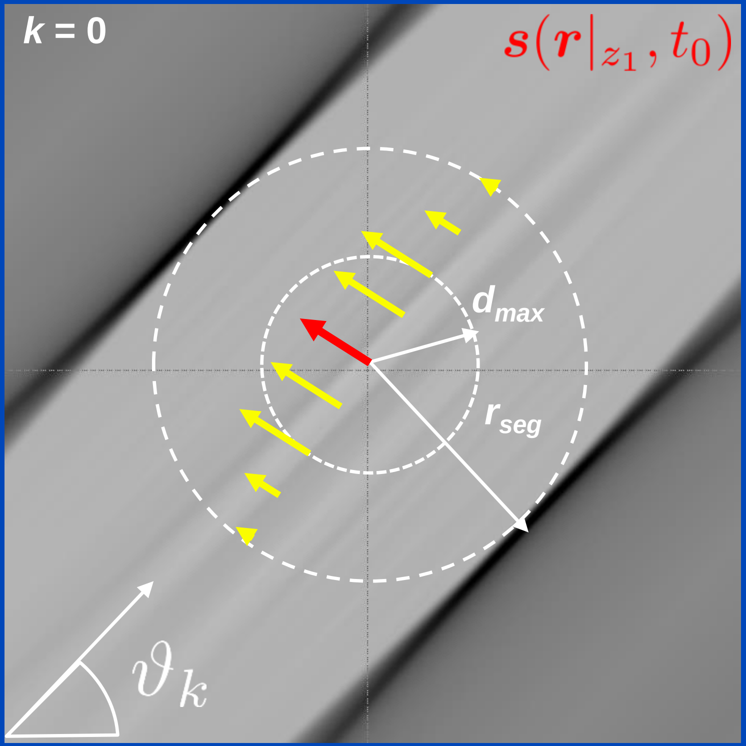
- Motion model: Motion is modeled by a motion vector field (MVF) $\boldsymbol{s}(\boldsymbol{r}, t)$ sub-sampled in time and space, whose time dependence we parameterize by a low degree polynomial ($P \leq 2$)

$$\boldsymbol{s}(\boldsymbol{r}, t) = \sum_{p=1}^{P} \boldsymbol{a}_p(\boldsymbol{r})(t - t_0)^p.$$

- For each artery, each stack and each control point incorporated in the latter a set of parameters $\boldsymbol{a}_p(\boldsymbol{r}_n)$ is determined separately.

- Between the control points, the MVF is approximated by linear interpolation.

Algorithmic Concept
Motion Compensation



- **Create a dense MVF, which drops to zero at the borders of the segmented region.**
- **Motion compensation (MoCo): Apply MVF on *2K + 1* PARs $f_k(\boldsymbol{r})$ and add them to obtain the motion-compensated reconstruction**

$$f_{\mathrm{MoCo}}(\boldsymbol{r}, \boldsymbol{s}) = \sum_{k=-K}^{K} f_k(\boldsymbol{r} + \boldsymbol{s}(\boldsymbol{r}, t_k))$$

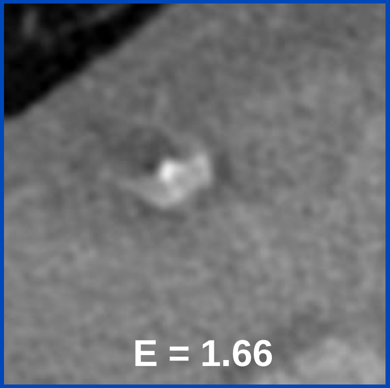
Algorithmic Concept

Motion Estimation

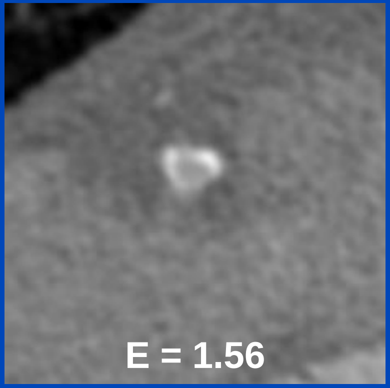
- Motion estimation: The MVFs are subject to the cost function optimization:

$$\hat{\boldsymbol{s}} = \arg \min_{\boldsymbol{s} \in \mathbb{R}^{PND}} E,$$

- As image artifact measuring cost function, we chose the image's entropy.



High entropy



Low entropy

- The cost function is only evaluated inside the ROI.

For 3D MoCo, N = 25, P = 2 → 150 parameters

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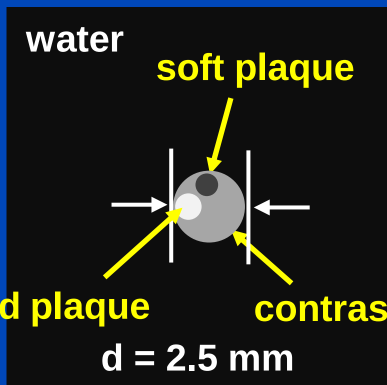
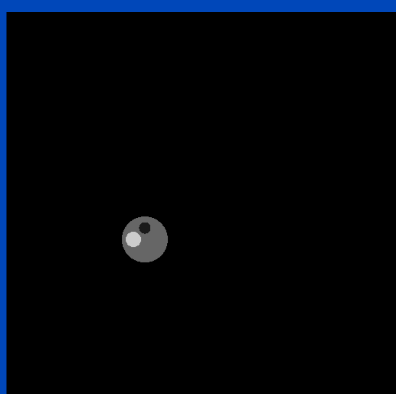
Simulation Study



motion



- **Settings:**
  - Low pitch spiral scanning: p ≈ 0.2
    → Reconstruction of multiple cardiac phases possible.
  - Rotation time t_{rot} = 300 ms
  - Heart rate 70 bpm
  - Noise
- **For the evaluation of the algorithm we choose *P* = 2.**

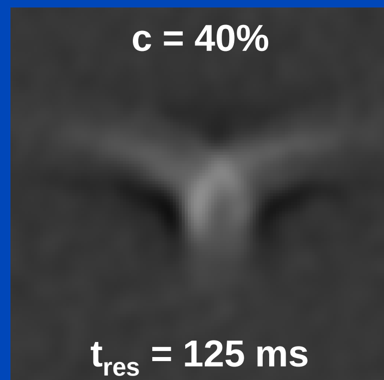
C = 400 HU; *W* = 750 HU

Results

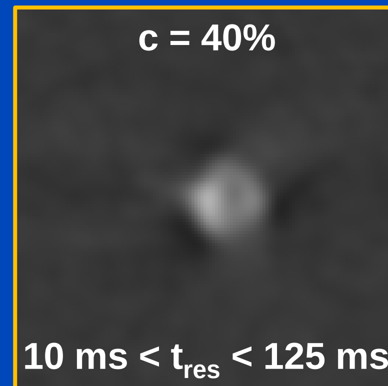
Simulation Study (70 bpm)

Vessel phantom static reference

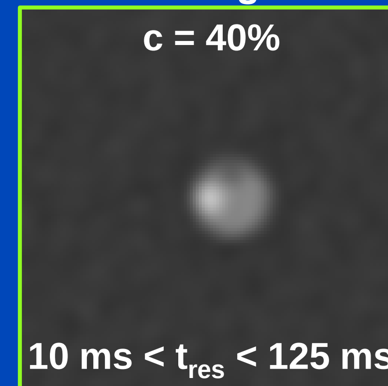
Standard FBP reconstruction

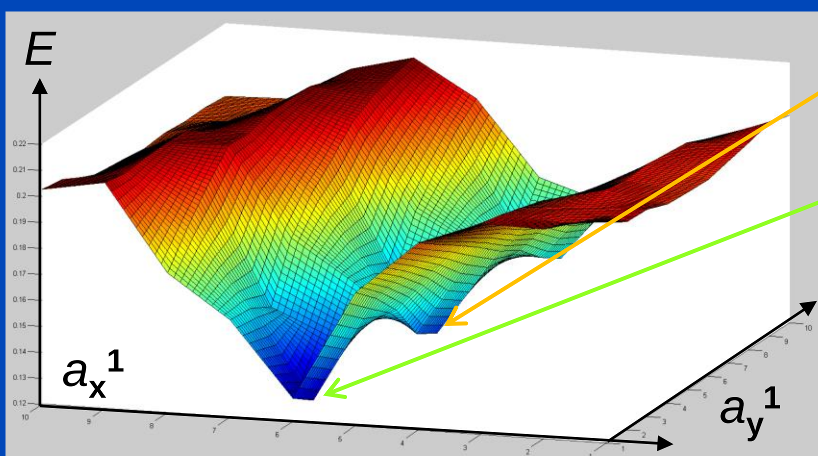


Optimization with Powell‘s algorithm



Optimization with re-initialization of Powell‘s algorithm





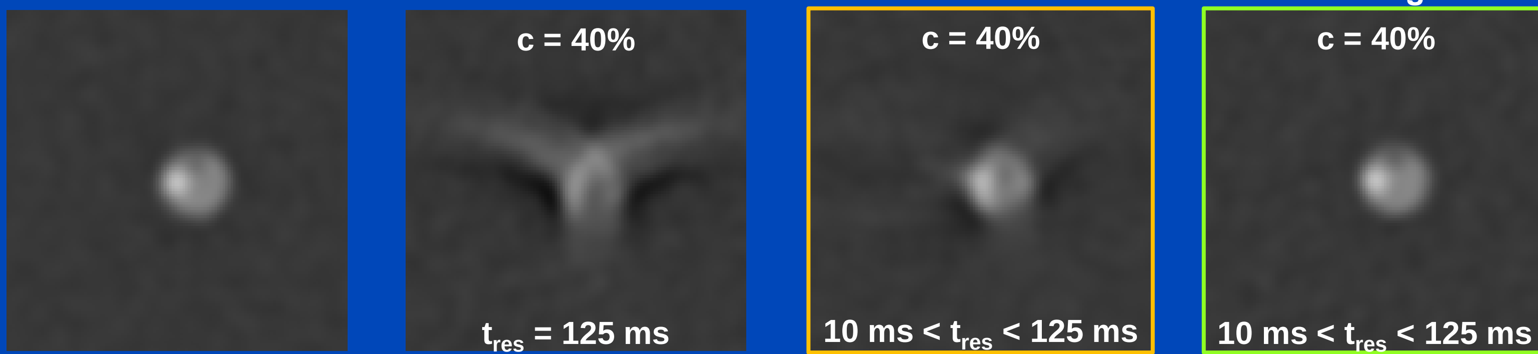
Optimization might be trapped in local minimum. Re-initialization of the optimization helps to escape from local minima.

Results

Simulation Study: Entropy Improvement

Vessel phantom static reference | Standard FBP reconstruction | Optimization with Powells's algorithm | Optimization with re-initialization of Powells's algorithm

c = 40%



Relative improvement in entropy: $$\frac{E(f_{\mathrm{FBP}}) - E(f_{\mathrm{MoCo}})}{E(f_{\mathrm{FBP}})}$$

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C = 400 HU; W = 1500 HU

dkfz.

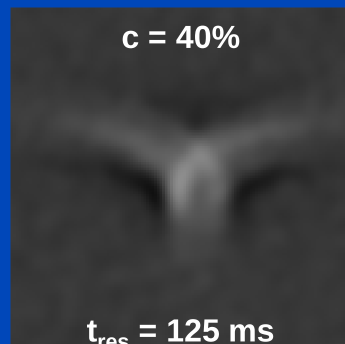
Results

Simulation Study: Entropy Improvement

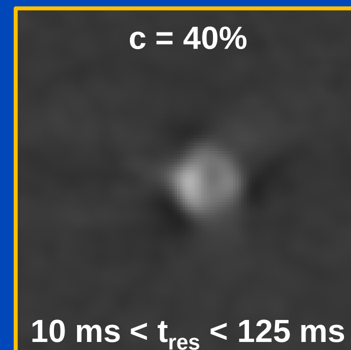
c = 40%

Vessel phantom static reference

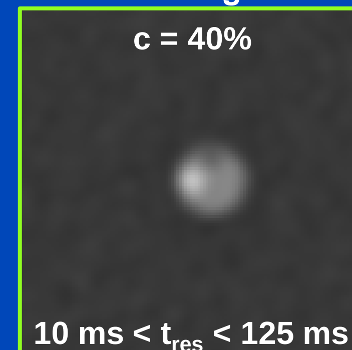
Standard FBP reconstruction

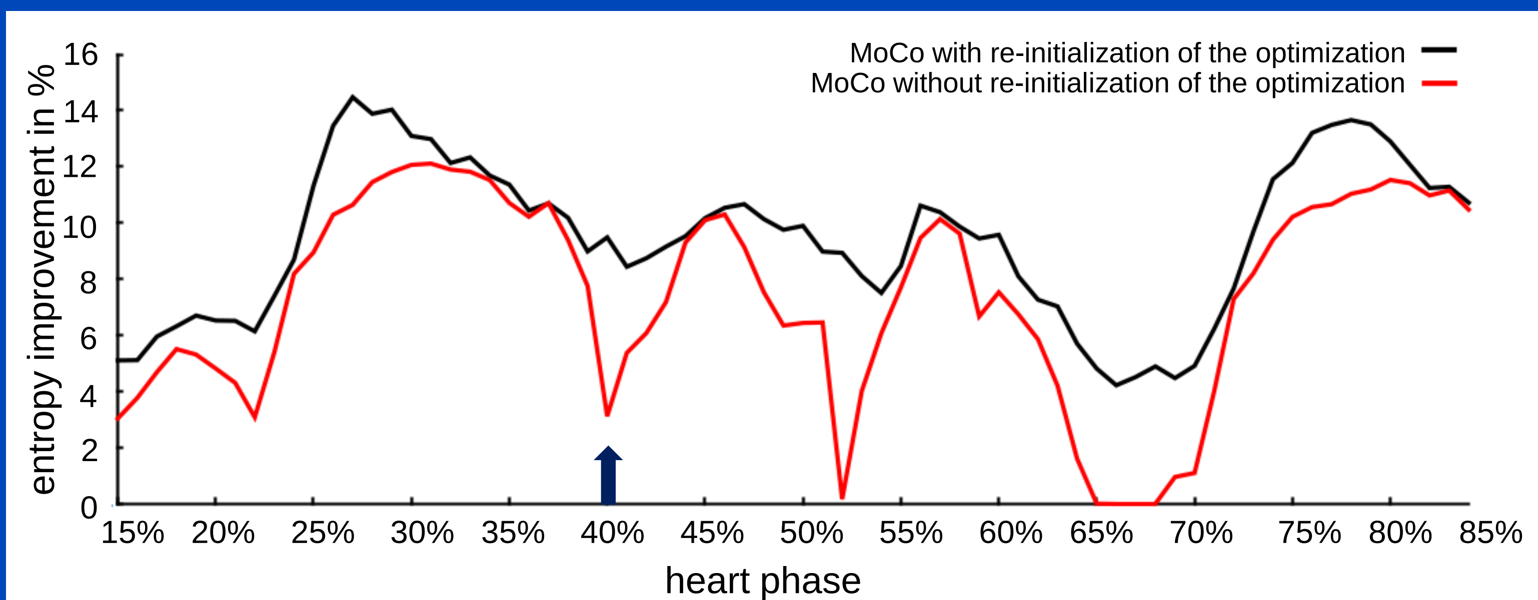


Optimization with Powells's algorithm



Optimization with re-initialization of Powells's algorithm





Results

Simulation Study: Entropy Improvement

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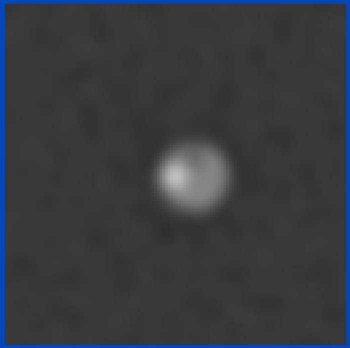
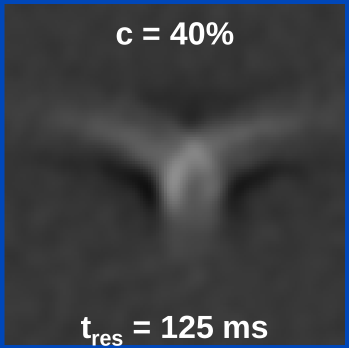
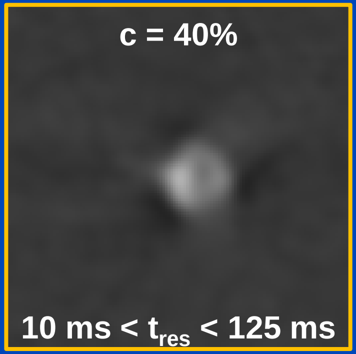
Vessel phantom static reference



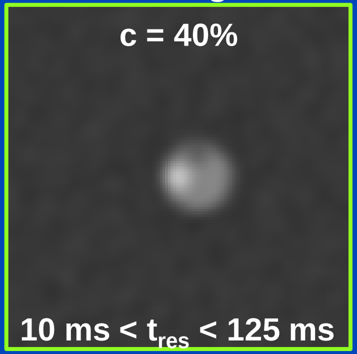
Standard FBP reconstruction

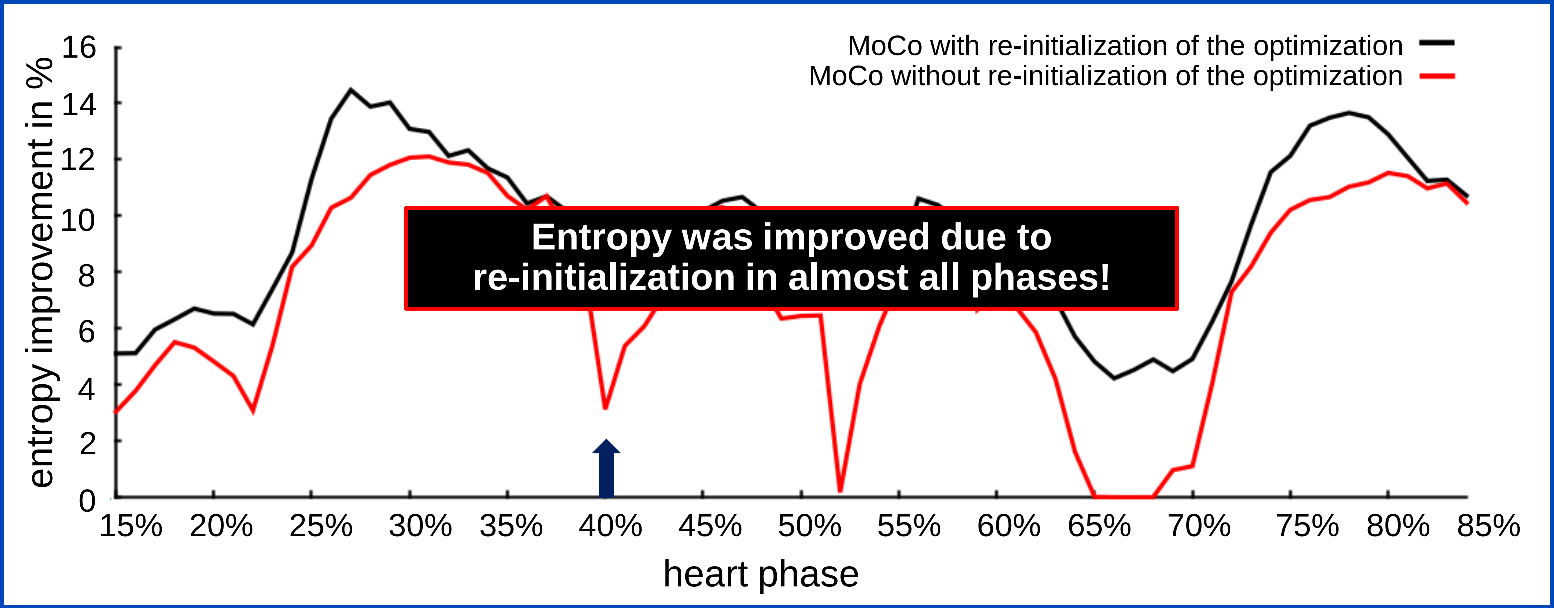


Optimization with Powells's algorithm



Optimization with re-initialization of Powells's algorithm





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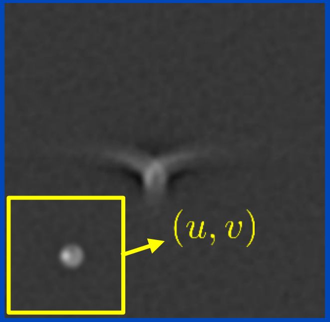
C = 400 HU; *W* = 1500 HU

dkfz.

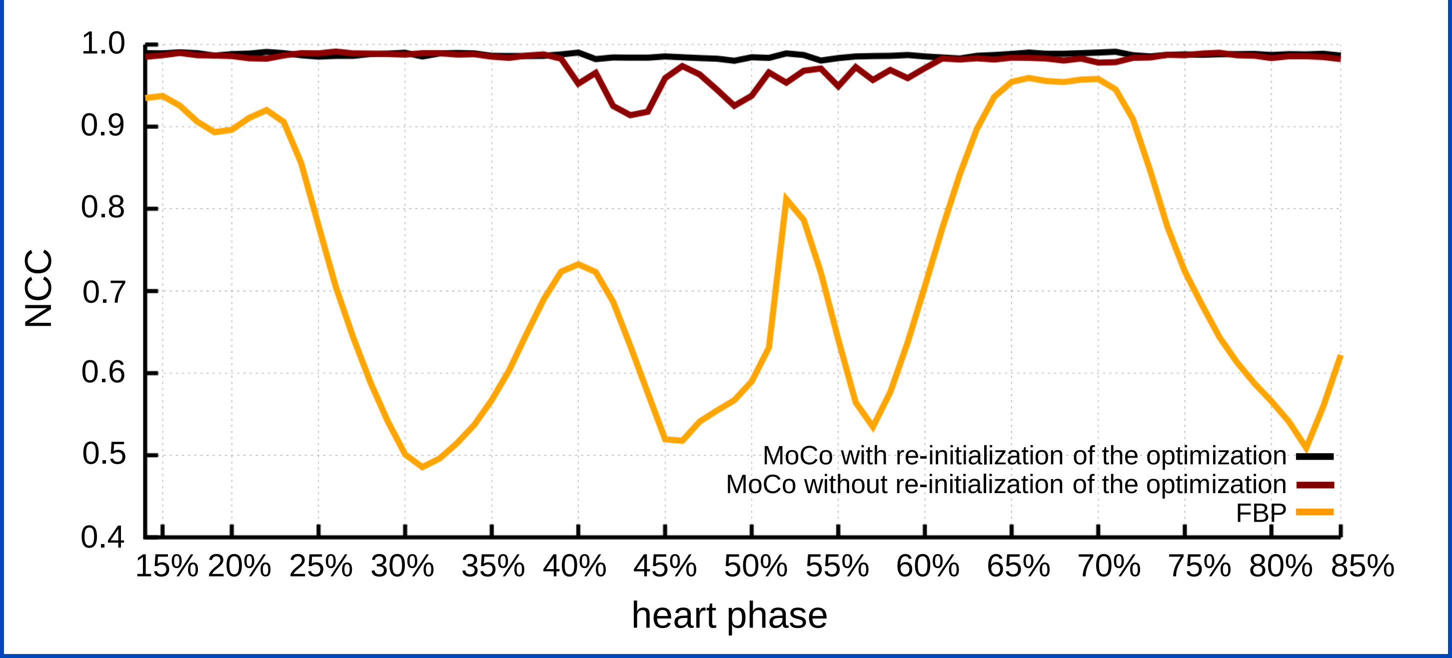


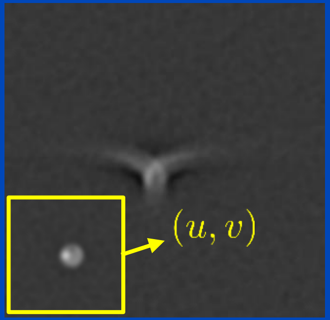
Results

Simulation Study: Image Quality

NCC between static reference image and a reconstruction.

$$\mathrm{NCC}(u,v) = \frac{\sum_{i,j}(f_t(i,j)-\bar{f}_{t,u,v})(f_r(i-u,j-v)-\bar{f}_r)}{\left(\sum_{i,j}(f_t(i,j)-\bar{f}_{t,u,v})^2\sum_{i,j}(f_r(i-u,j-v)-\bar{f}_r)^2\right)^{\frac{1}{2}}}$$





Results

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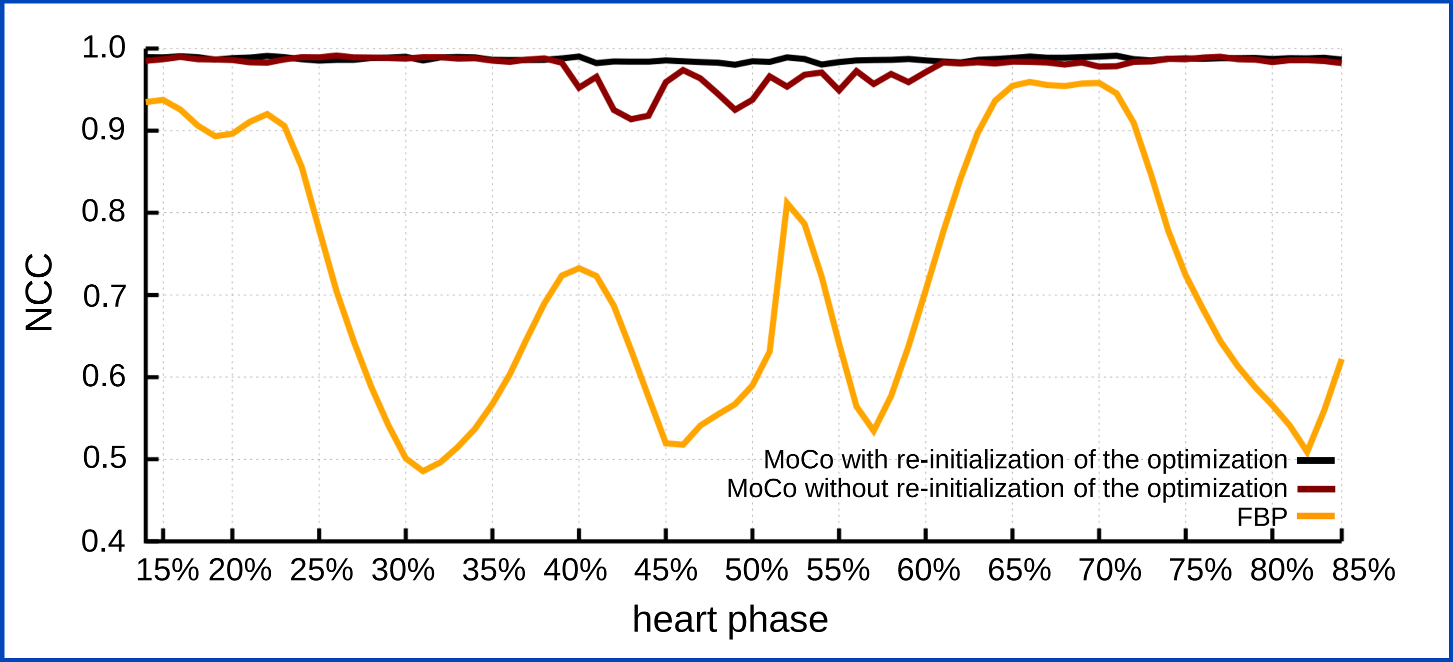
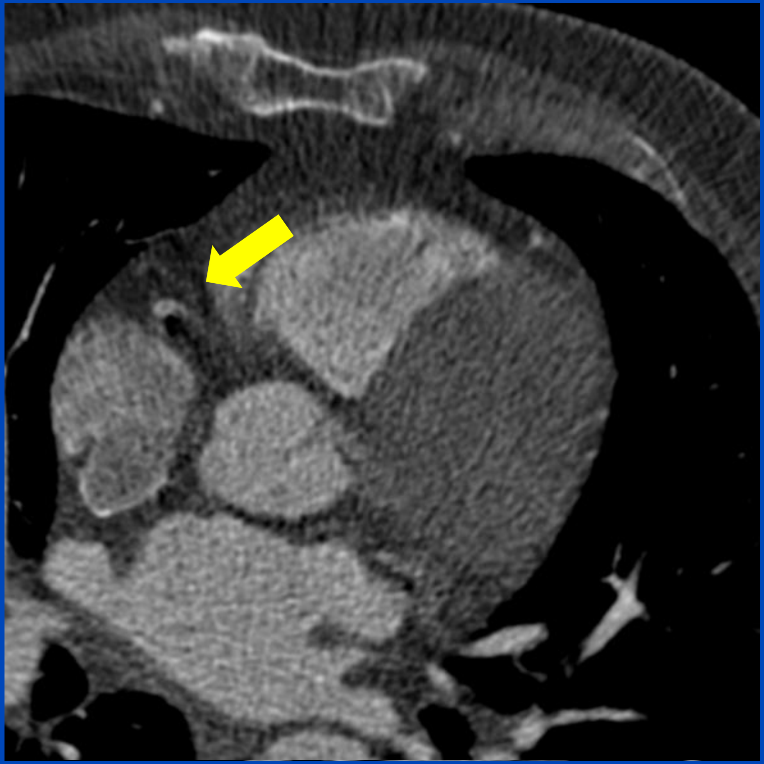


An almost constant image quality is obtained!

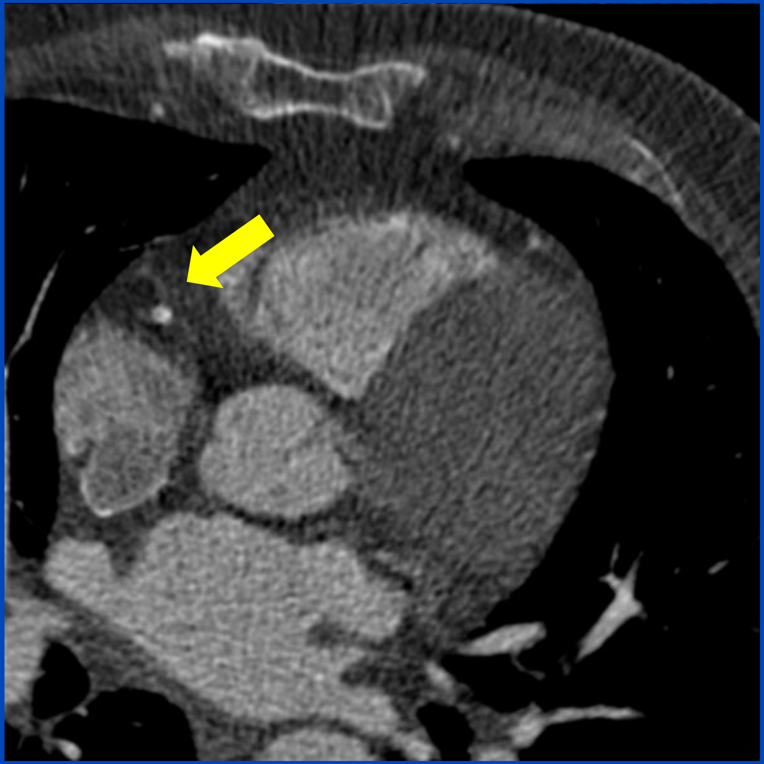
Results
Clinical Case 1

t_{res} = 143 ms, HR = 72 bpm, c = 70% RR

Standard reconstruction



MoCo reconstruction

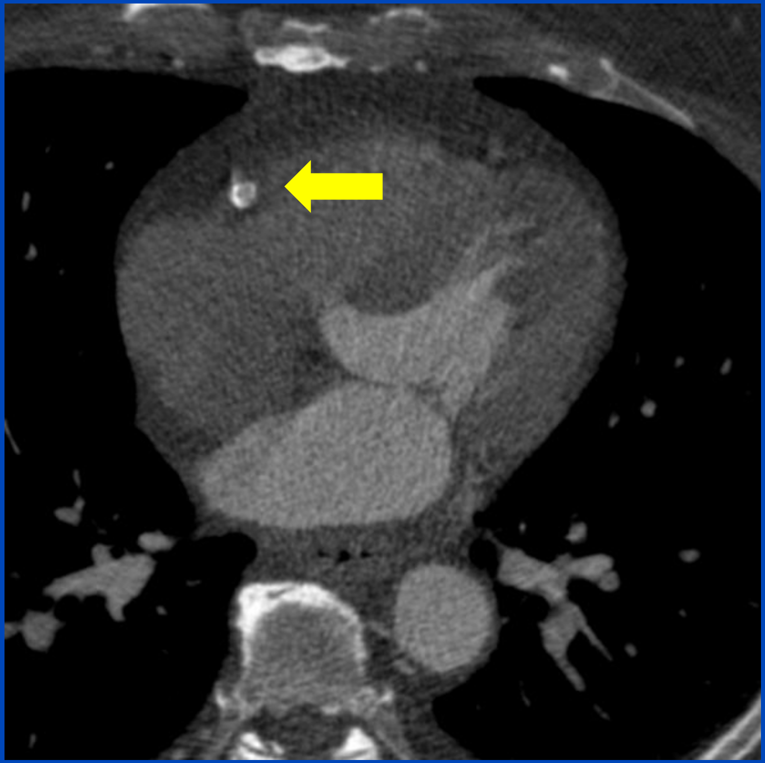


Phase shifted by 5% from the best phase to obtain an image with motion artifacts

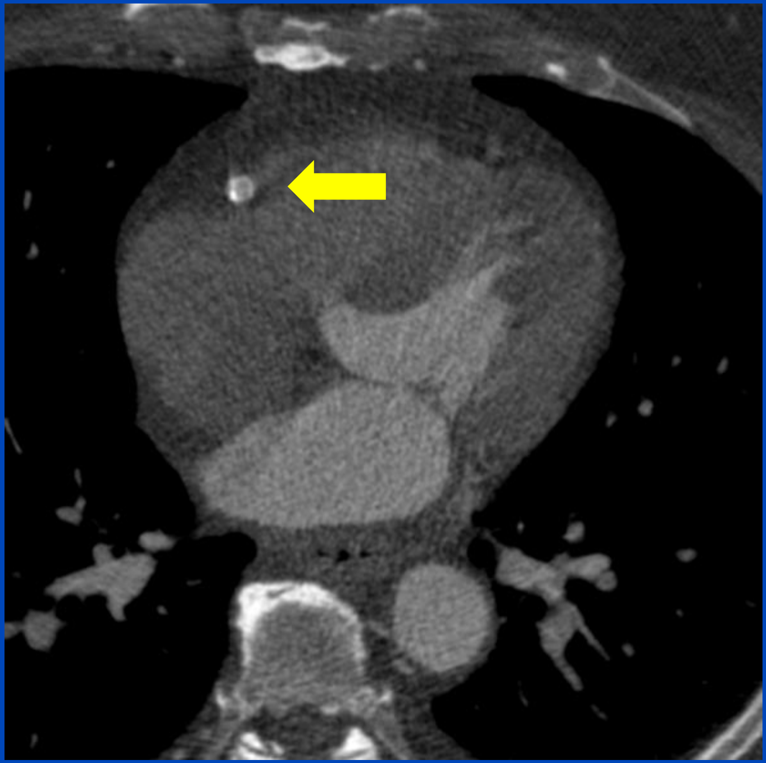
Results
Clinical Case 2

t_{res} = 143 ms, HR = 70 bpm, c = 40% RR

Standard reconstruction



MoCo reconstruction

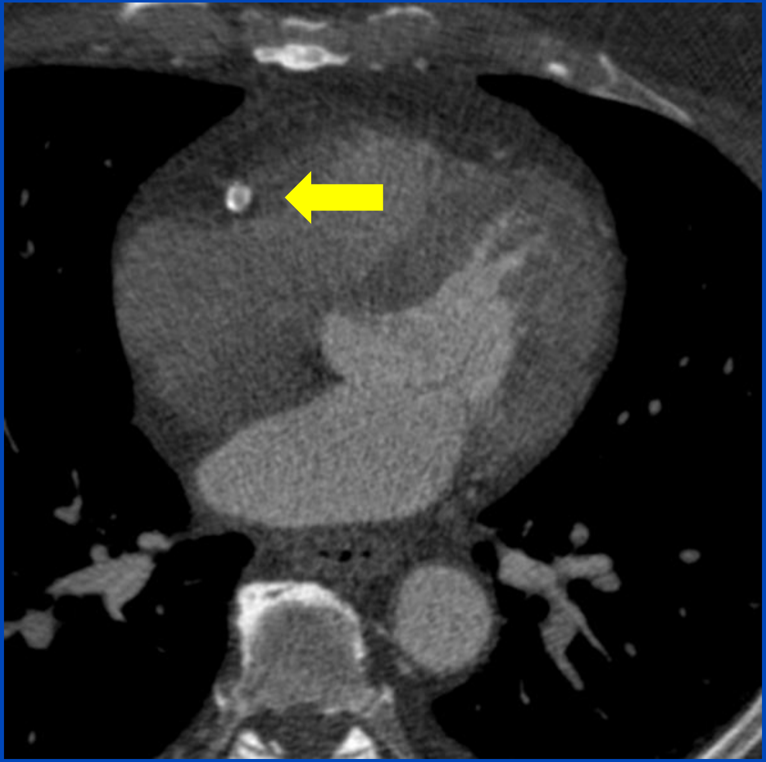


Results
Clinical Case 2

t_{res} = 143 ms, HR = 70 bpm, c = 50% RR

Standard reconstruction

MoCo reconstruction



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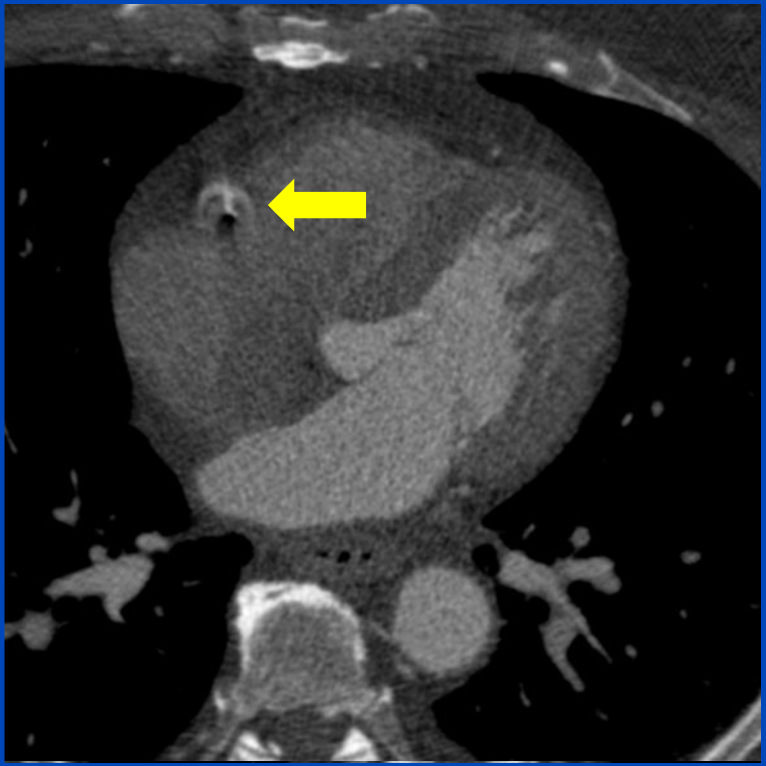
C = 400 HU; W = 1500 HU

dkfz.

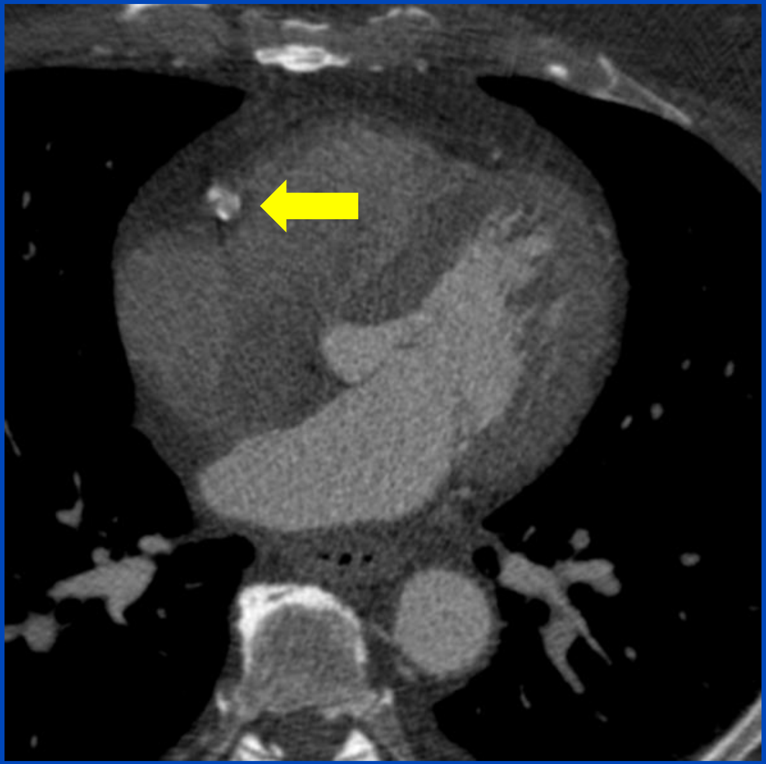
Results
Clinical Case 2

t_{res} = 143 ms, HR = 70 bpm, c = 60% RR

Standard reconstruction



MoCo reconstruction



Summary and Conclusion

- We see an increased sharpness of the coronary arteries in cardiac phases featuring motion artifacts of different severity.
- The computational effort is potentially low because of the simple way the MVFs are applied.
- Potential applications are:
  - Dual source high pitch scan protocols at high heart rates
  - Single source cardiac CT at high heart rates
- More on MoCo of our group:
  - Sauppe, Kachelrieß. 5D MoCo for respiratory and cardiac motion with CBCT of the thorax region.
    Sun, Feb 28

Thank You!

This study was supported by Siemens Healthcare GmbH.

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