Motion-Compensation in CT
and in other Modalities

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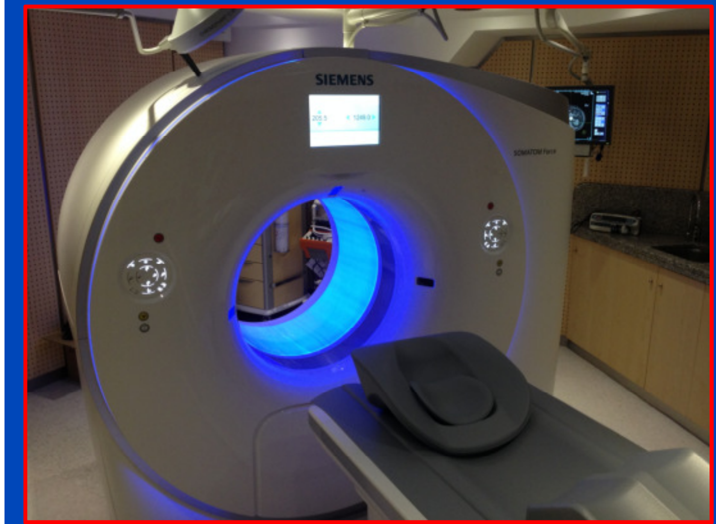
DEUTSCHES
KREBSFORSCHUNGSZENTRUM
IN DER HELMHOLTZ-GEMEINSCHAFT

Definiton and Contents

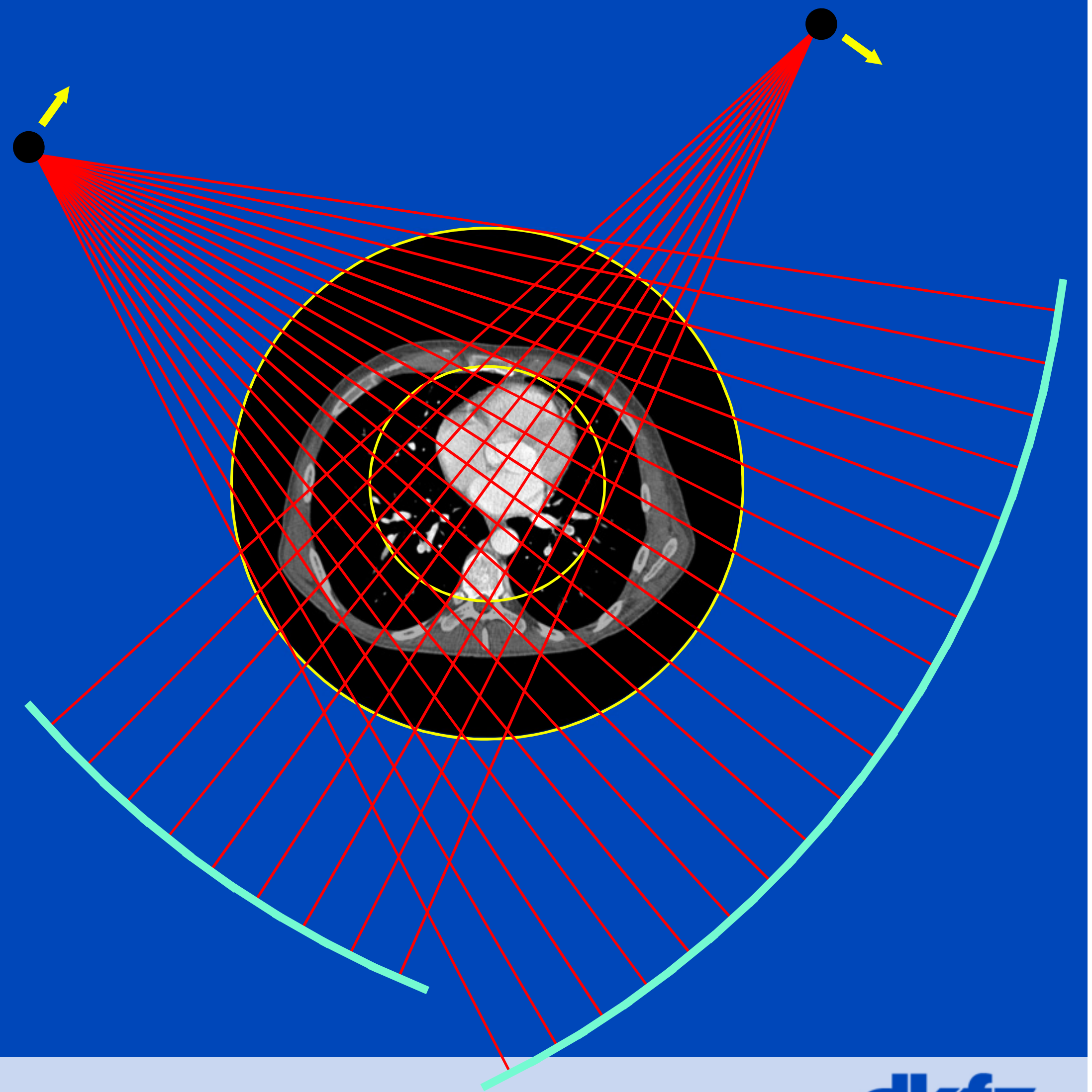
- Motion compensation (MoCo) is
  - a computational motion management approach
  - that estimates motion and
  - that compensates (corrects) for the estimated motion

- Scans much faster than one motion cycle
  - MoCo for cardiac CT
- Scans much slower than one motion cycle
  - MoCo for respiratory CBCT
  - MoCo for cardiac CBCT
  - MoCo for MR and PET/MR

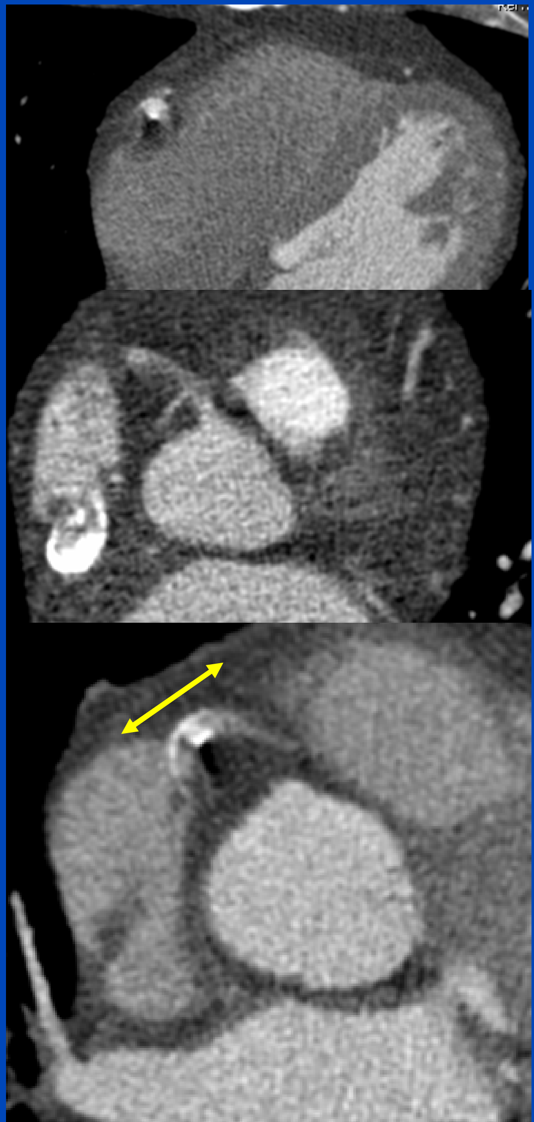
Cardiac CT



Siemens SOMATOM Force
dual source cone-beam spiral CT



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 $d \approx \frac{t_{rot}}{2}\,\bar{v} \approx \frac{250}{2}\,\mathrm{ms}\; 50\,\frac{\mathrm{mm}}{\mathrm{s}} = 6.25\ \mathrm{mm}$ are possible (RCA mean velocity measurements[1,2,3,4]).
- 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.

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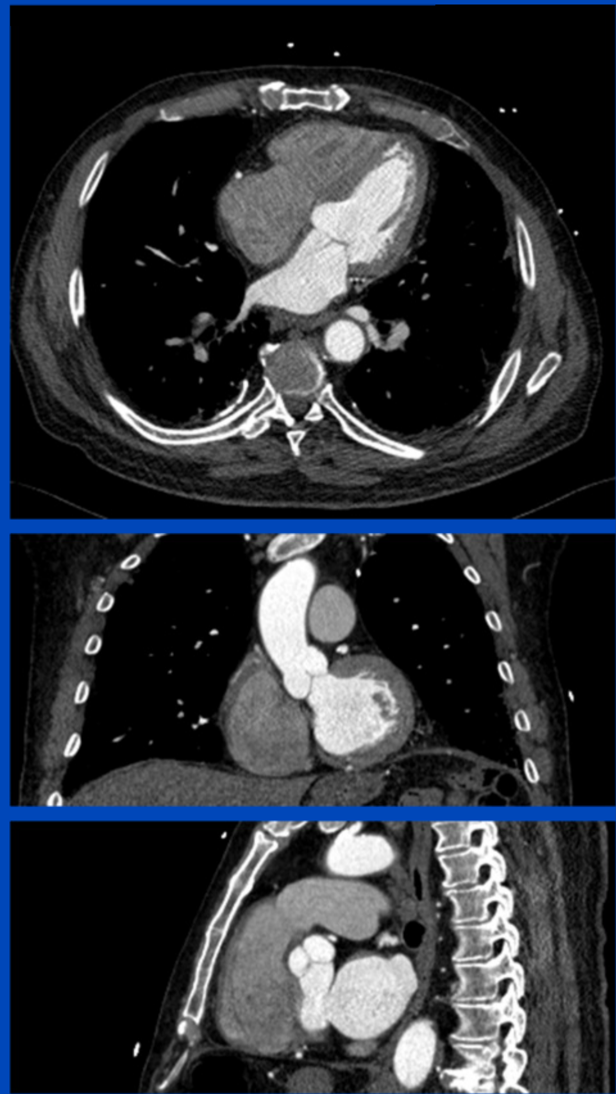
Motion Compensation is the Future!

Cardiac CT MoCo Strategies

- Acquire and reconstruct
  all phases
  - determine the MVFs (difficult)
  - either map all phases into a target phase
  - or improve on each phase separately
- Acquire and reconstruct
  some phases
  - determine the MVFs (very difficult)
  - either map all phases into a target phase
  - or improve on each phase separately
- Acquire and reconstruct
  a single phase
  - determine the MVFs (extremely difficult because there are no redundancies)
  - improve on the single phase image



Algorithms to Improve Temporal Resolution in Cardiac CT

	Data Range	Anatomical Landmarks	Dose Usage	MoCo (MVFs)
Taguchi et al. (Johns Hopkins)	1 heart cycle	no	100%	yes
SSF, Bhaglia et al. (GE)	>> 180°	arteries	<< 100%	yes
SSF+MEAD, Nett et al. (GE)	>> 180°	arteries	<< 100%	yes
Tang et al. (Toshiba)	>> 180°	arteries	<< 100%	yes
Kim et al. (KAIST)	> 180°	no	< 100%	yes
TRI-PICCS, Chen et al. (UW)	180°	no	< 100%	no
TRIM, Schöndube et al. (Siemens)	180°	arteries	< 100%	no
MAM, Rohkohl et al. (Siemens)	180°	arteries	100%	yes
PAMoCo, Hahn et al. (DKFZ)	180°	arteries	100%	yes

All algorithms can potentially also be applied to DSCT. However, this has not been done, yet.

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

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

Juliane Hahn[a,b], Herbert Bruder[b], Thomas Allmendinger[b], Karl Stierstorfer[b], Thomas Flohr[b], and Marc Kachelrieß[a]

[a]Medical Physics in Radiology, German Cancer Research Center (DKFZ), Im Neuenheimer Feld 280, 69120 Heidelberg, Germany

[b]Computed Tomography, Siemens Healthcare GmbH, Siemensstraße 1, 91301 Forchheim, Germany

ABSTRACT

Until today, several software-based approaches to increase the temporal resolution in cardiac computed tomography by estimating motion vector fields (MVFs) have been developed. Thereunder, the majority are motion compensation algorithms, which estimate the MVFs employing a three-dimensional registration routine working on reconstructions of multiple cardiac phases.[2,6,7,12]

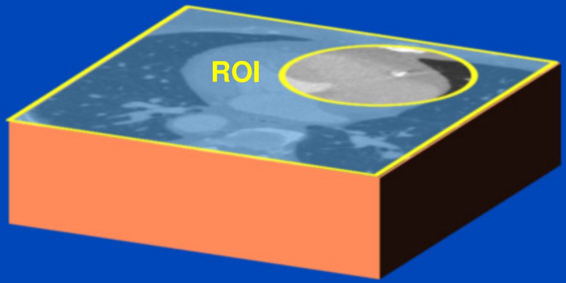
We present an algorithm that requires nothing more than the data needed for a short scan reconstruction for motion estimation and motion-compensated reconstruction, which both are based on the reconstruction of volumes from a limited angular range.[2,3,7,8] Those partial angle reconstructions are centered at different time

J. Hahn, M. Kachelrieß et al. Reduction of motion artifacts in cardiac CT based on partial angle reconstruc-tions from short scan data. SPIE Medical Imaging Conference Record 97831A:1-9, March 2016.

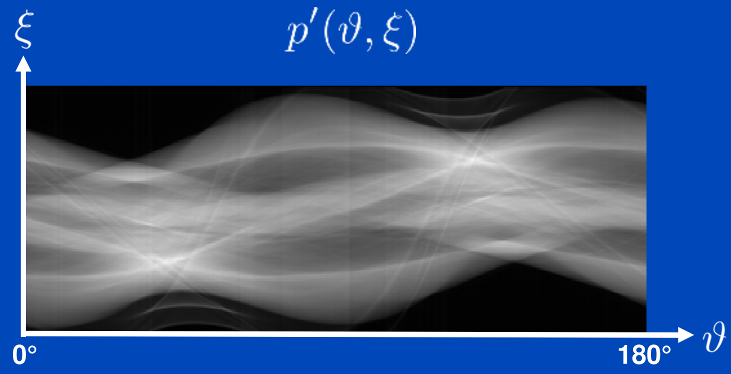
PAMoCo

Generate 2K+1 Partial Angle Reconstructions

Initial segmented stack volume



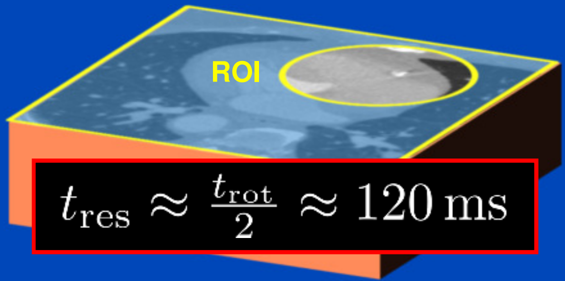
Subdivide the projection data $p'(\vartheta, \xi)$
into 2K + 1 overlapping sectors



PAMoCo

Generate 2K+1 Partial Angle Reconstructions

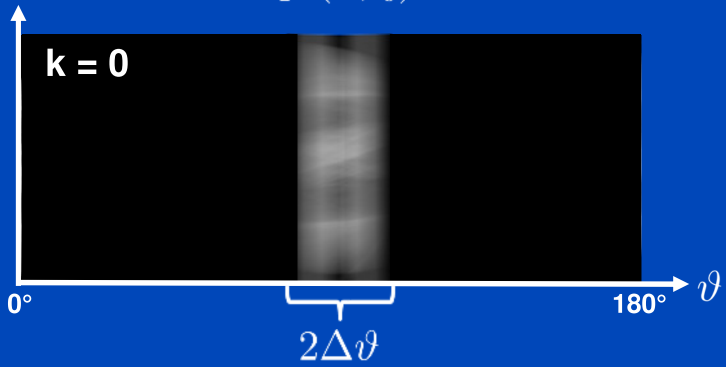
Initial segmented stack volume



$$t_{\text{res}} \approx \frac{t_{\text{rot}}}{2} \approx 120\,\text{ms}$$

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

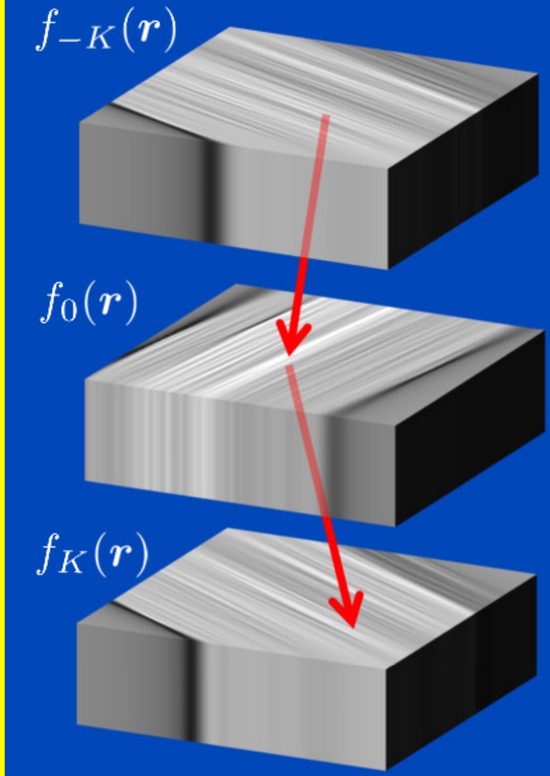
$p'(\vartheta,\xi)$



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

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

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 = 12**

J. Hahn, M. Kachelrieß et al. Reduction of motion artifacts in cardiac CT based on partial angle reconstructions from short scan data. SPIE Medical Imaging Conference Record 97831A:1-9, March 2016.

PAMoCo Motion Model

- Control points along coronary arteries

$$\boldsymbol{r} = \boldsymbol{r}(\lambda_n)$$

- Polynomial around each control point

$$\boldsymbol{d}(\boldsymbol{s}, \lambda, t) = \sum_{p,l} \boldsymbol{s}_{lp}(\lambda - \lambda_0)^l (t - t_0)^p$$

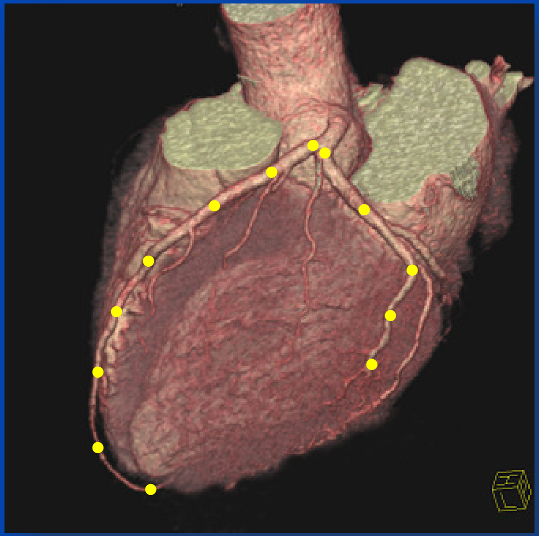
- DVFs continued onto all voxels

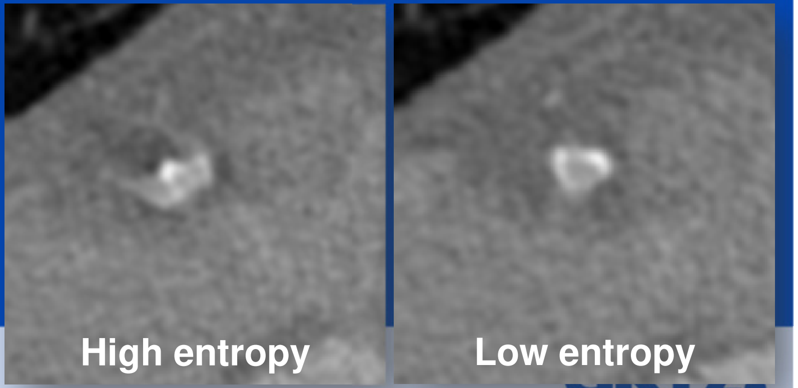
$$\boldsymbol{d} = \boldsymbol{d}(\boldsymbol{s}, \boldsymbol{r}, t)$$

- Sum up partial angle images

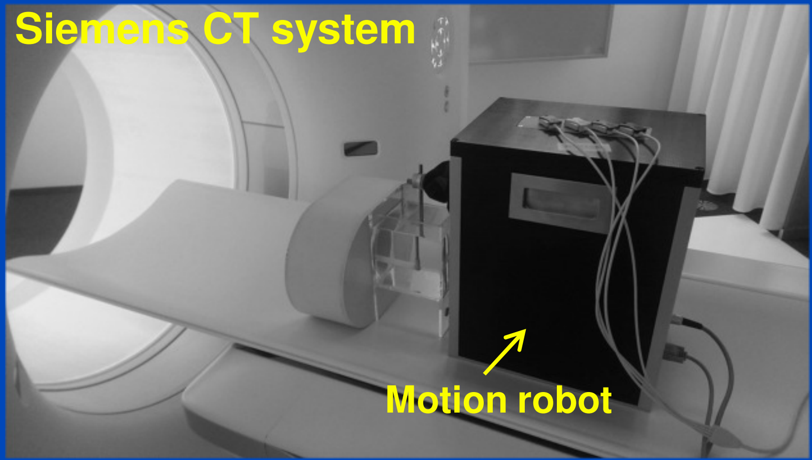
$$f_{\mathrm{MoCo}}(\boldsymbol{r}) = \sum_{k=-K}^{K} f_k\big(\boldsymbol{r} + \boldsymbol{d}(\boldsymbol{s}, \boldsymbol{r}, t_k)\big)$$

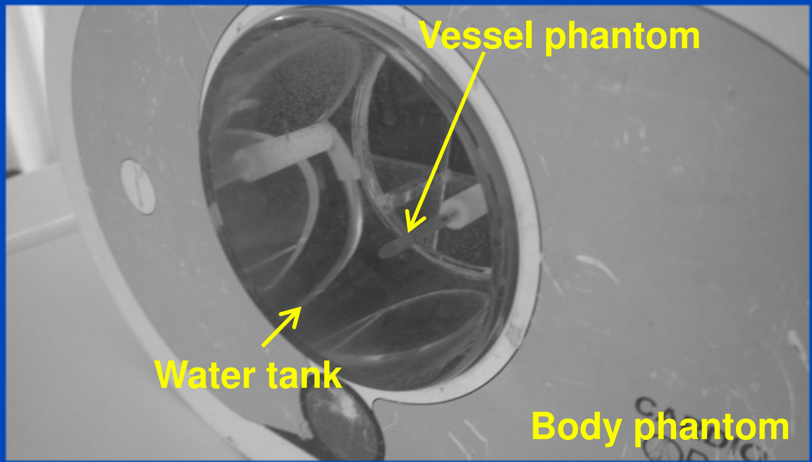
- Open DVF parameters chosen to minimize the image entropy



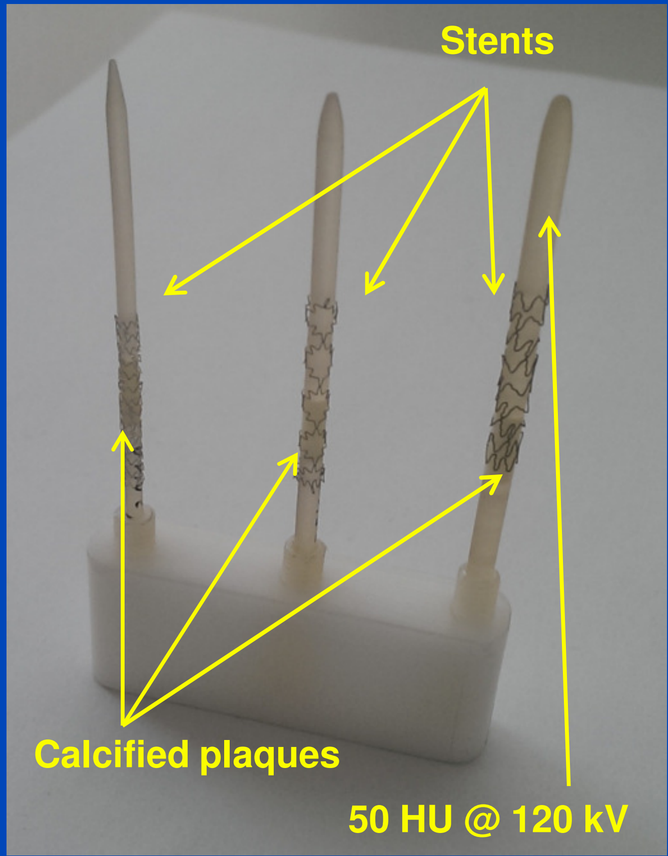


Phantom Measurement

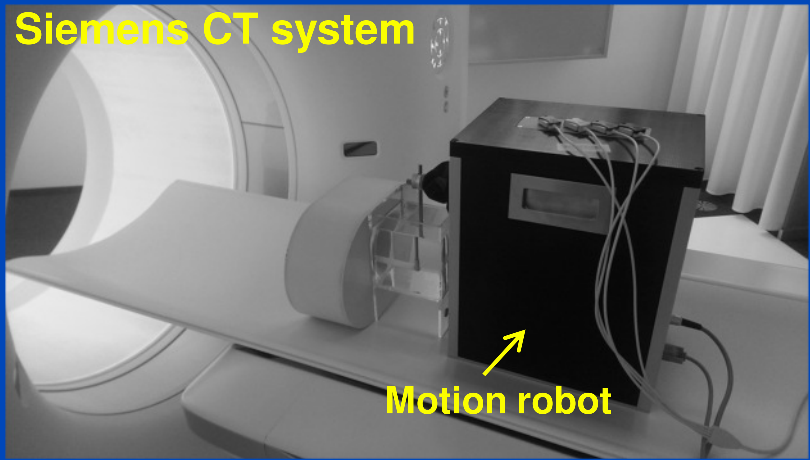


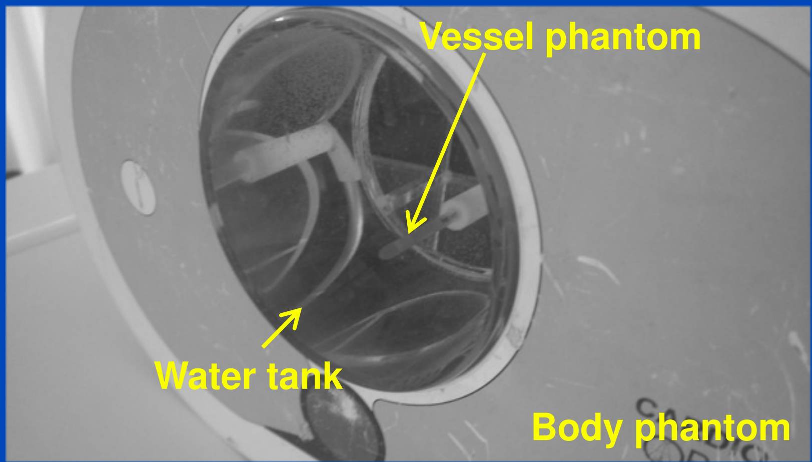


d = 1.5 mm 2.5 mm 3 mm



Phantom Measurement

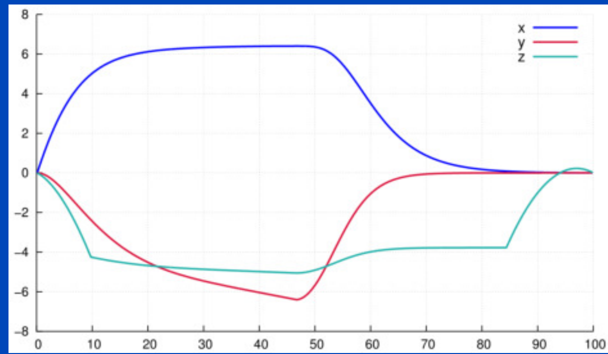




Applied motion

70 bpm

position in mm

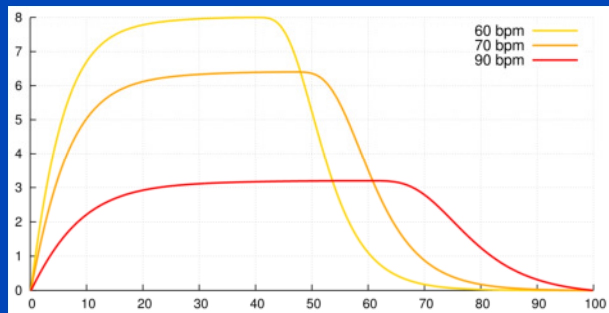


c in %

systole

diastole

x in mm

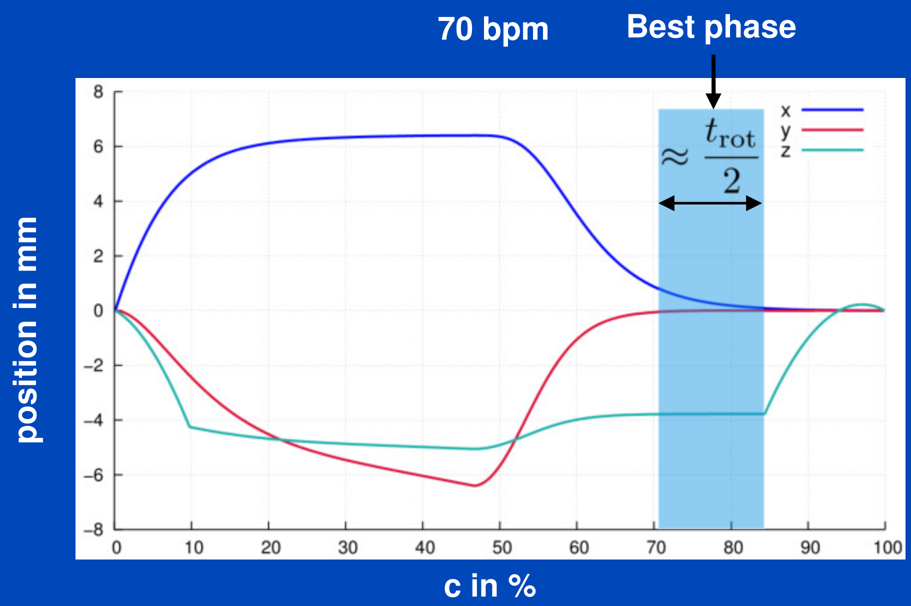


c in %

→ Non-proportional shortening of systolic and diastolic phase considered

Phantom Measurement

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

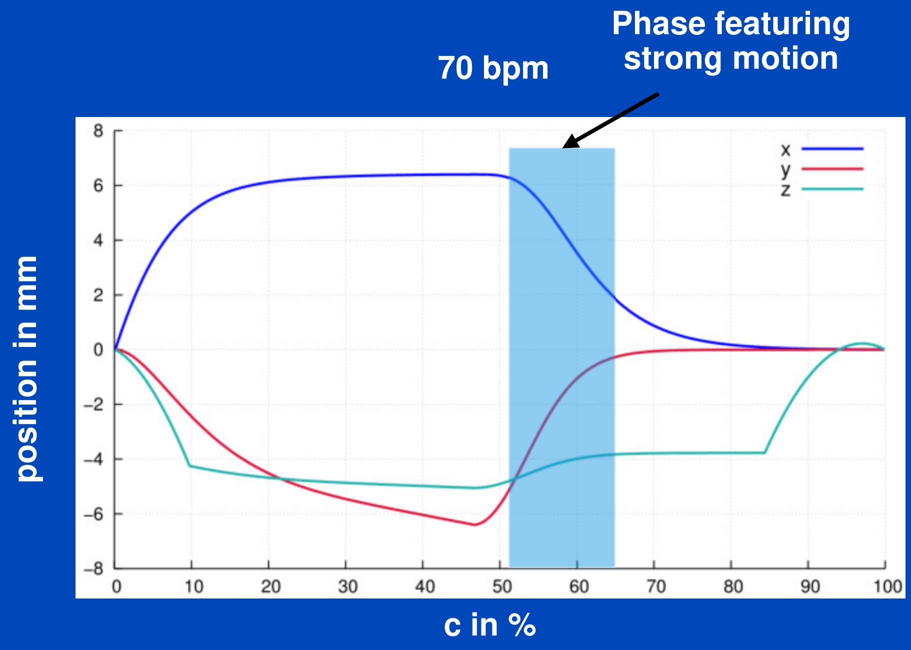


- MAM and TRIM aim at increasing the image quality close to the best phase.

- → 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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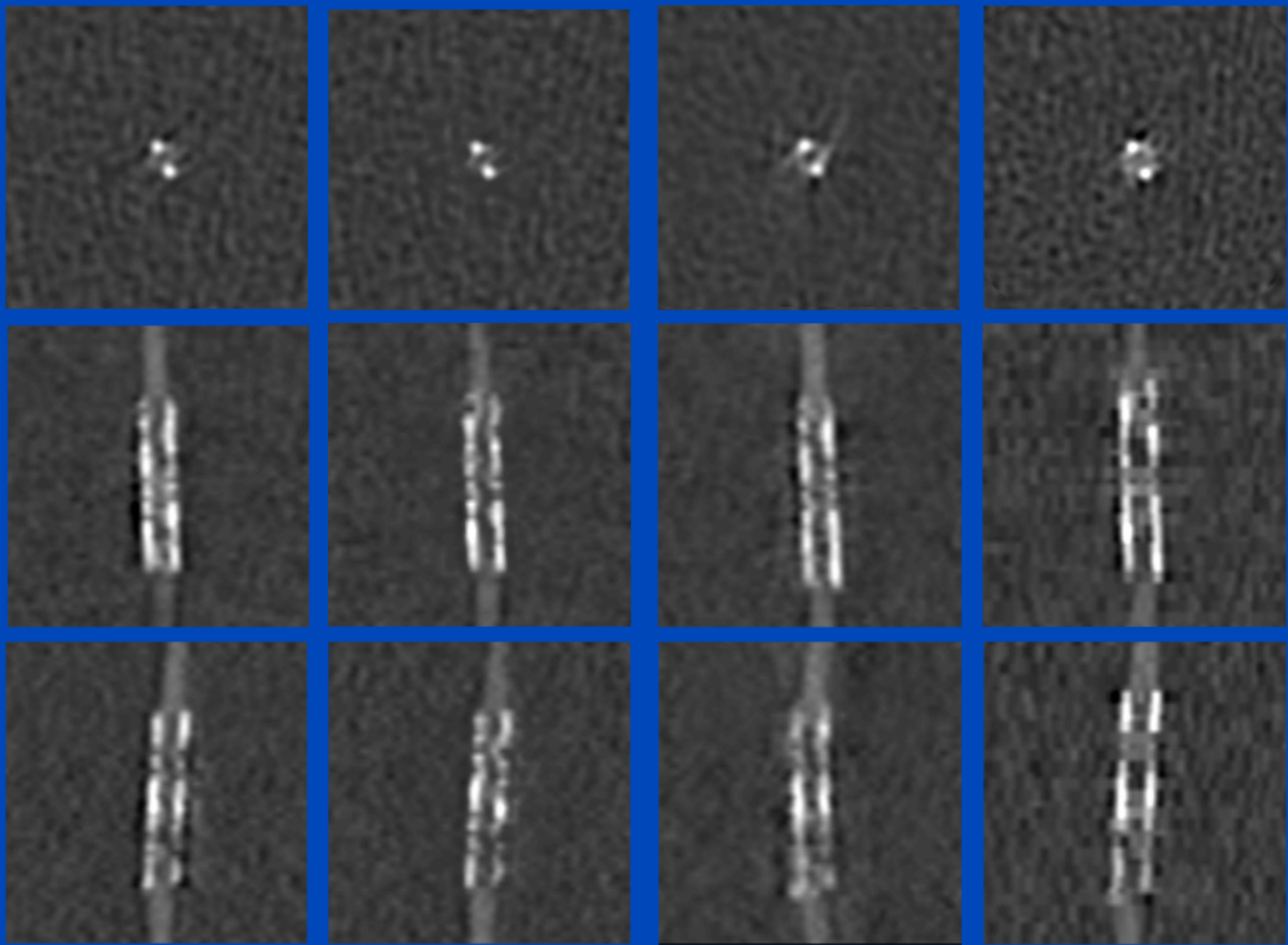
Phantom

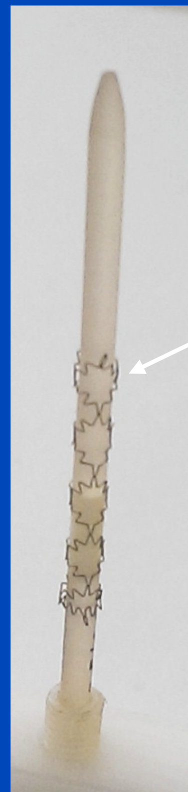
Best Phase

FBP | TRIM | MAM | PAMoCo

axial / sagittal / coronal



Vessel phantom



Stent

d = 2.5 mm

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

PAMoCo with $N_t \times N_\lambda \times 3 = 3 \times 3 \times 3 = 27$
parameter each stack

Phantom

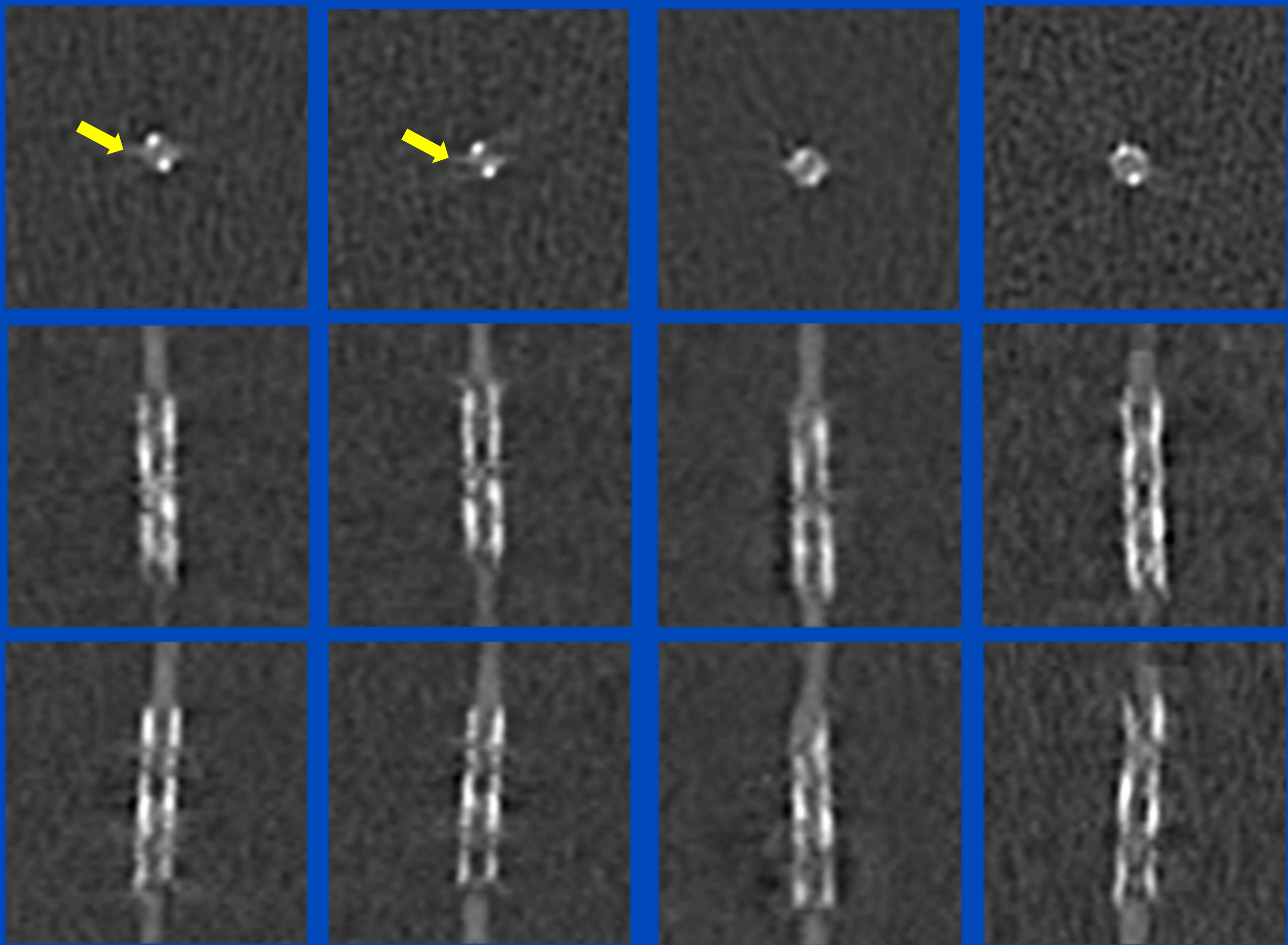
5% off Best Phase

FBP TRIM MAM PAMoCo

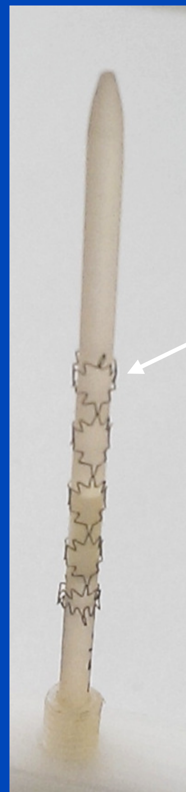
axial

sagittal

coronal



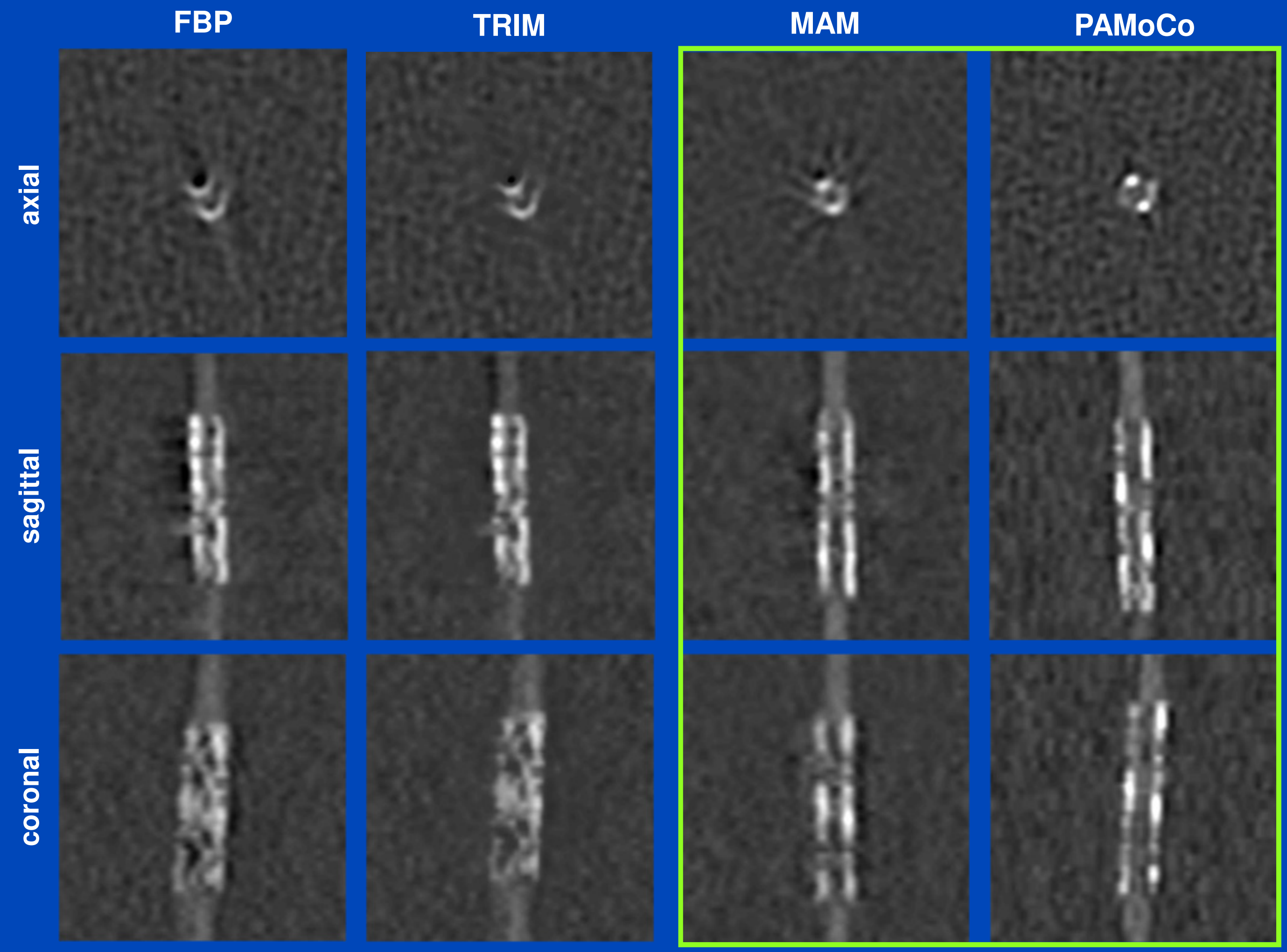
Vessel phantom

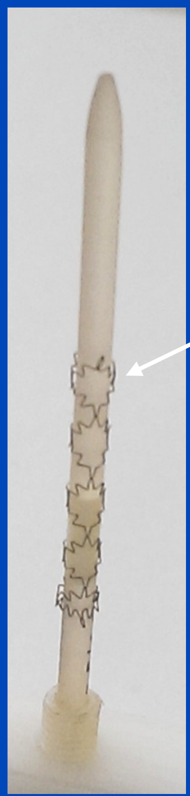


d = 2.5 mm

Phantom

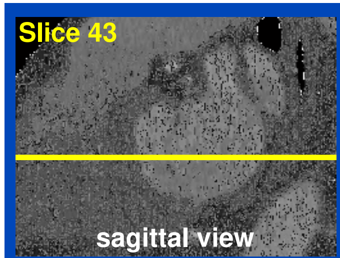
10% off Best Phase



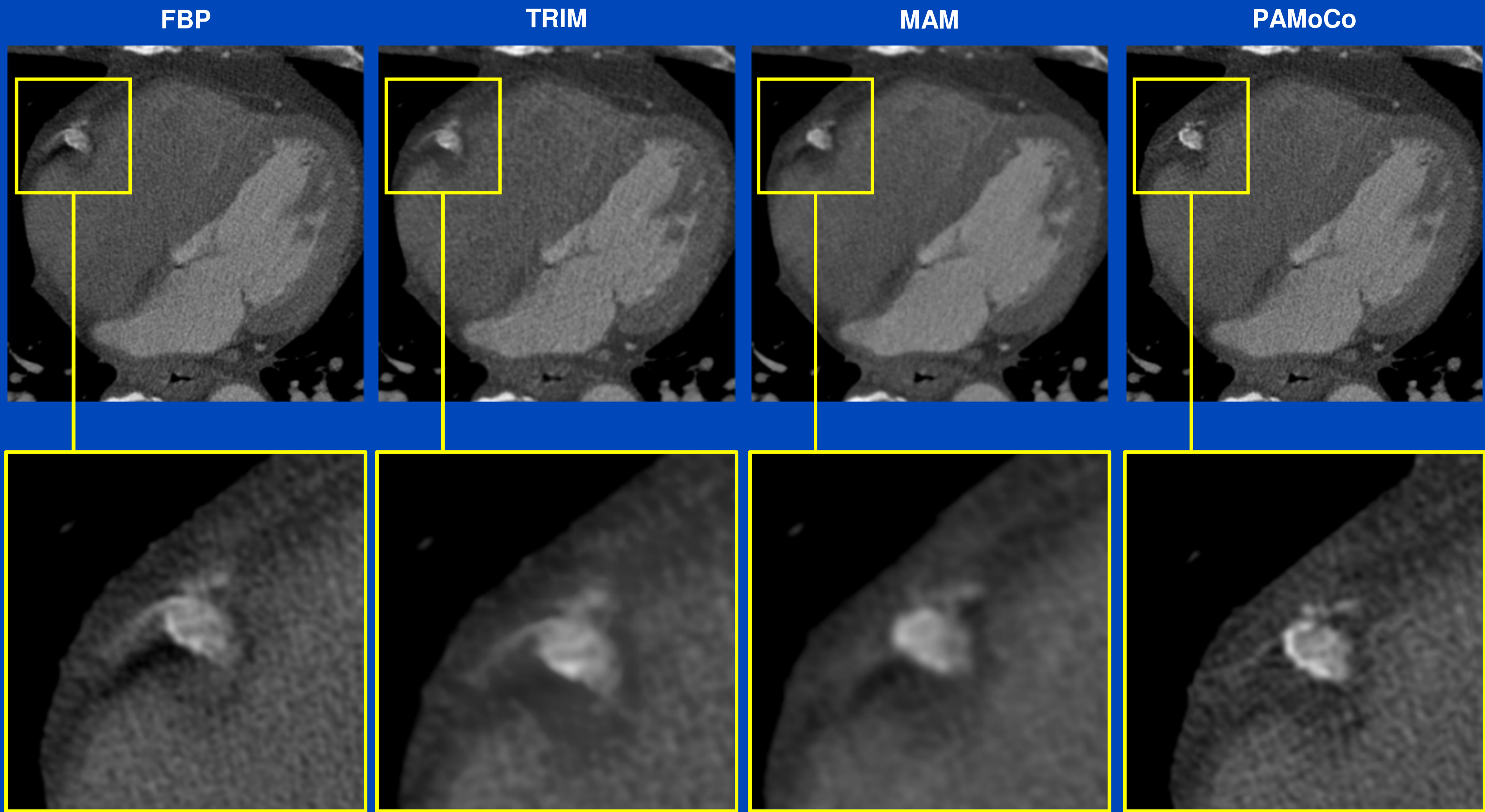


$\overline{HR}$ = 74 bpm, c = 74%,
C = 400 HU, W = 1500 HU

PAMoCo with $N_t \times N_\lambda \times 3 = 3 \times 3 \times 3 = 27$ parameter each stack

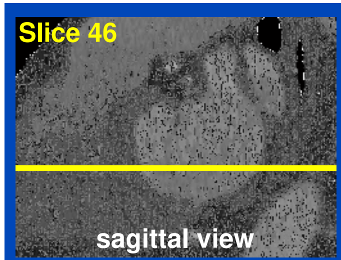


Patient 1

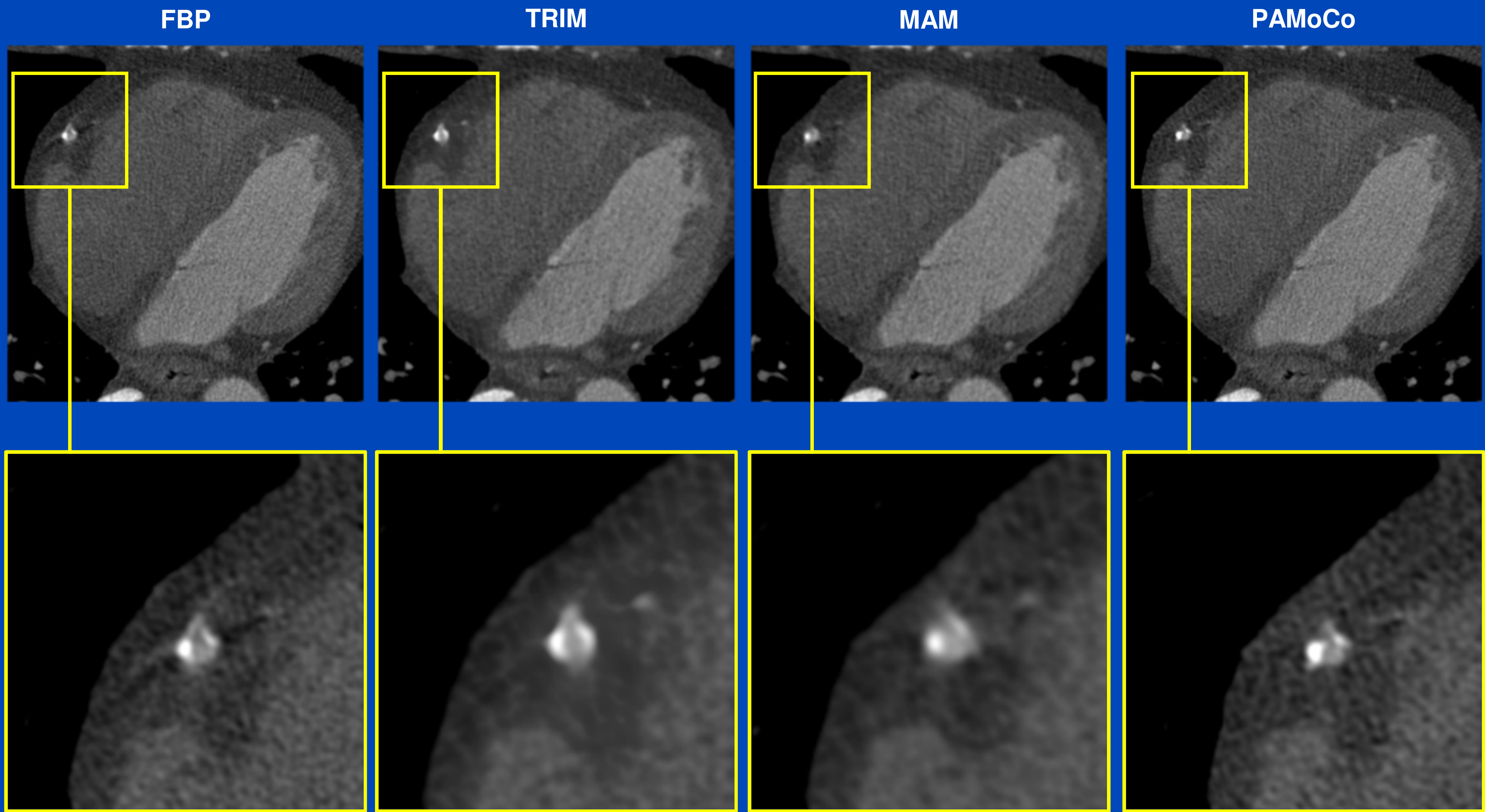


SIEMENS $\overline{\mathrm{HR}}$ = 74 bpm, c = 74%,
C = 400 HU, W = 1500 HU

PAMoCo with $N_t \times N_\lambda \times 3 = 3 \times 3 \times 3 = 27$
parameter each stack dkfz.



Patient 1



$\overline{HR}$ = 74 bpm, c = 74%,
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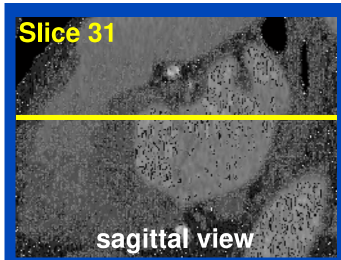
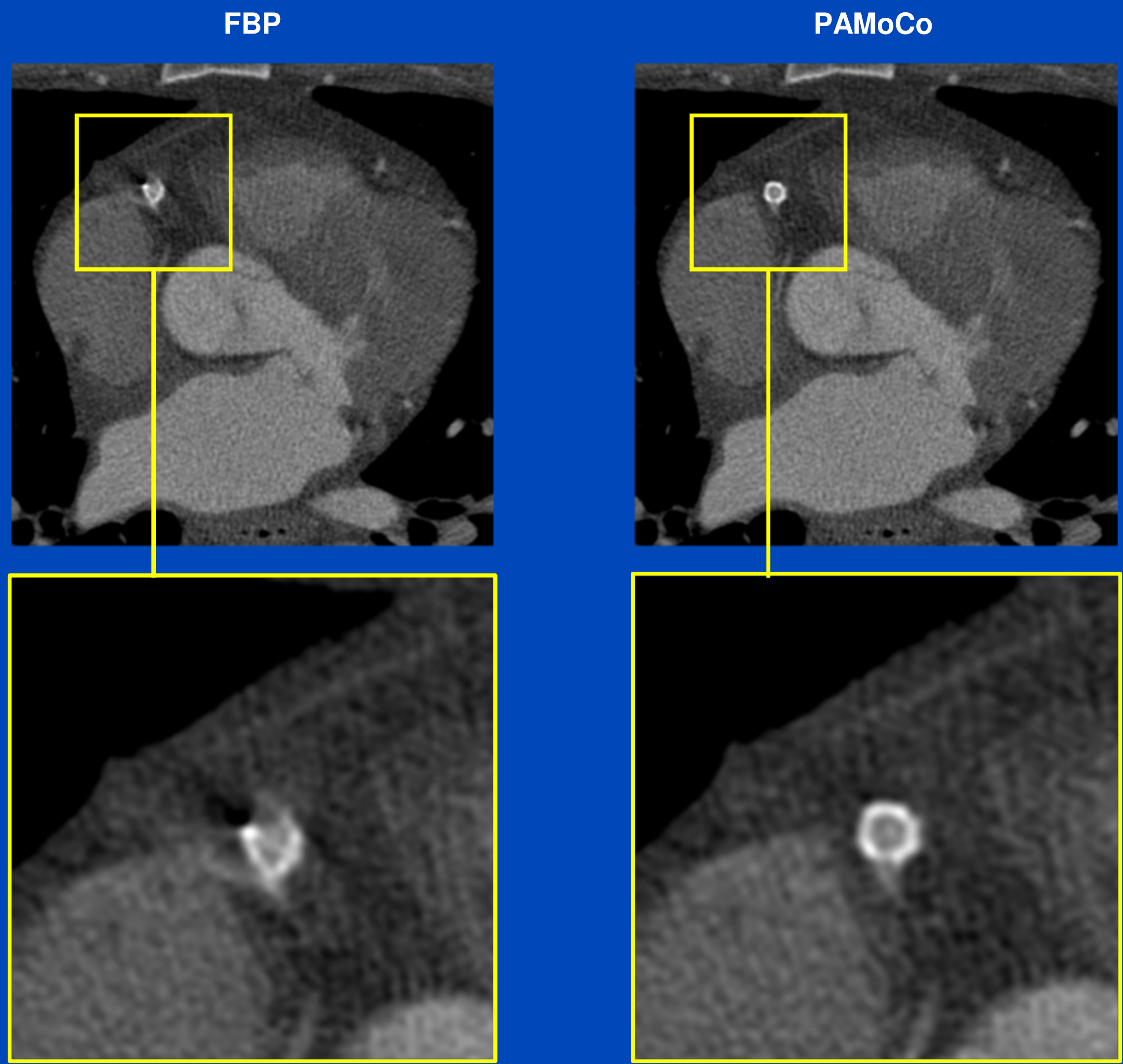
PAMoCo with $N_t \times N_\lambda \times 3 = 3 \times 3 \times 3 = 27$
parameter each stack



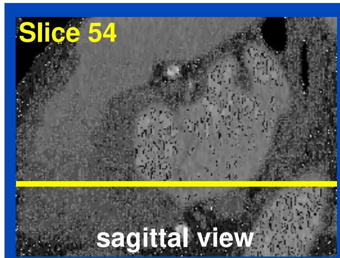
Patient 1



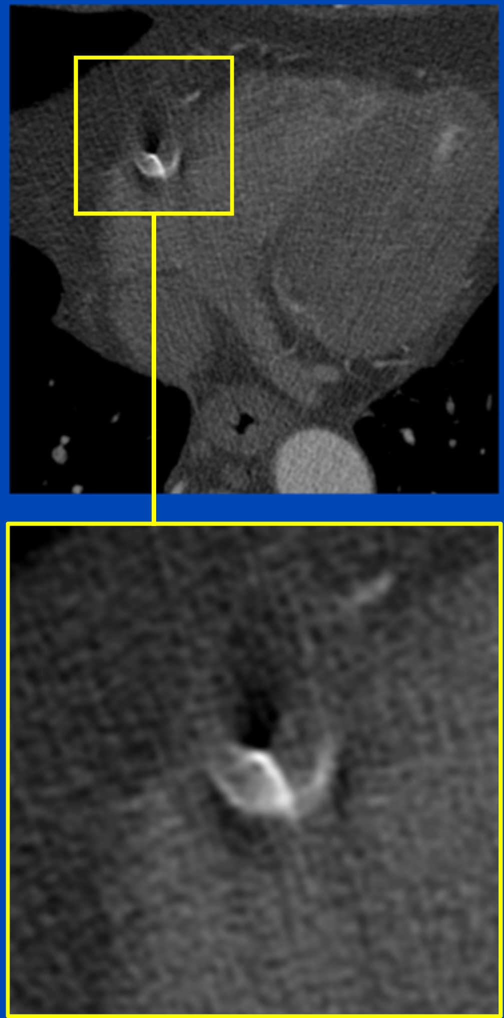
$\overline{\mathrm{HR}}$ = 74 bpm, c = 30%,
C = 400 HU, W = 1500 HU

PAMoCo with $N_t \times N_\lambda \times 3 = 3 \times 3 \times 3 = 27$ parameter each stack

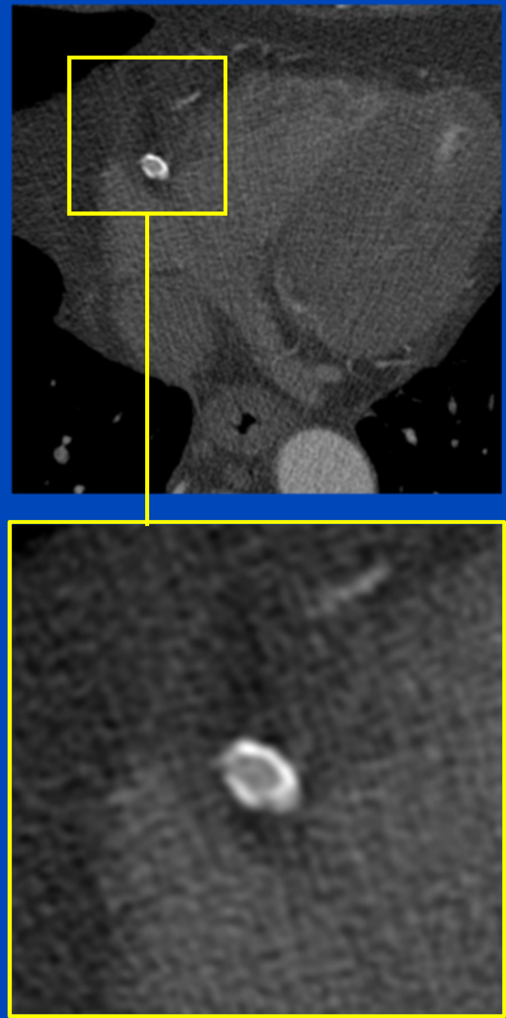
Patient 1



FBP



PAMoCo



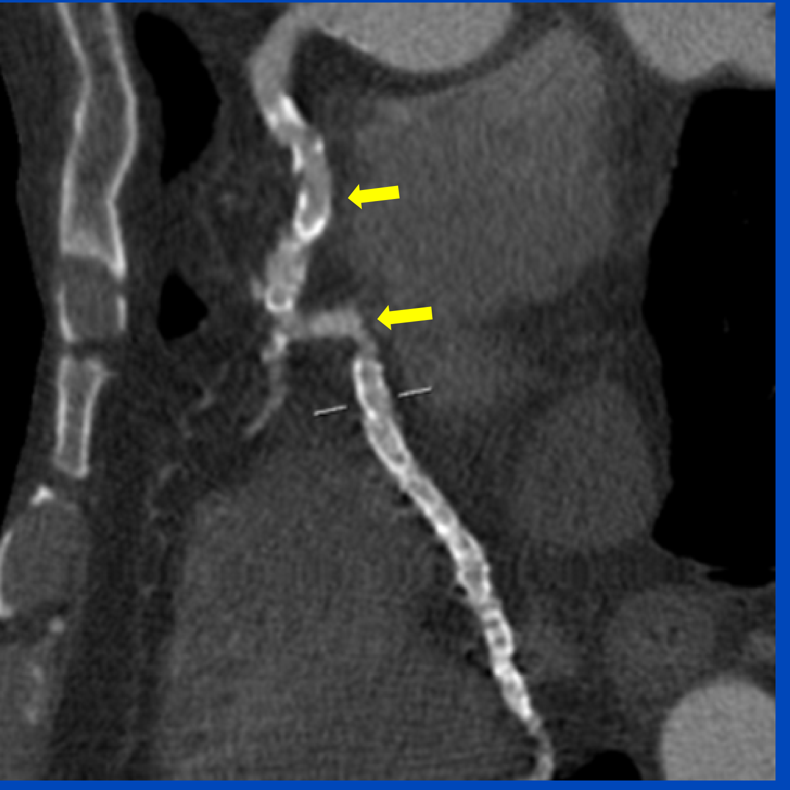
$\overline{HR}$ = 74 bpm, c = 30%,
C = 400 HU, W = 1500 HU

PAMoCo with $N_t \times N_\lambda \times 3 = 3 \times 3 \times 3 = 27$
parameter each stack

Patient 1

FBP

PAMoCo



curved MPRs of the RCA

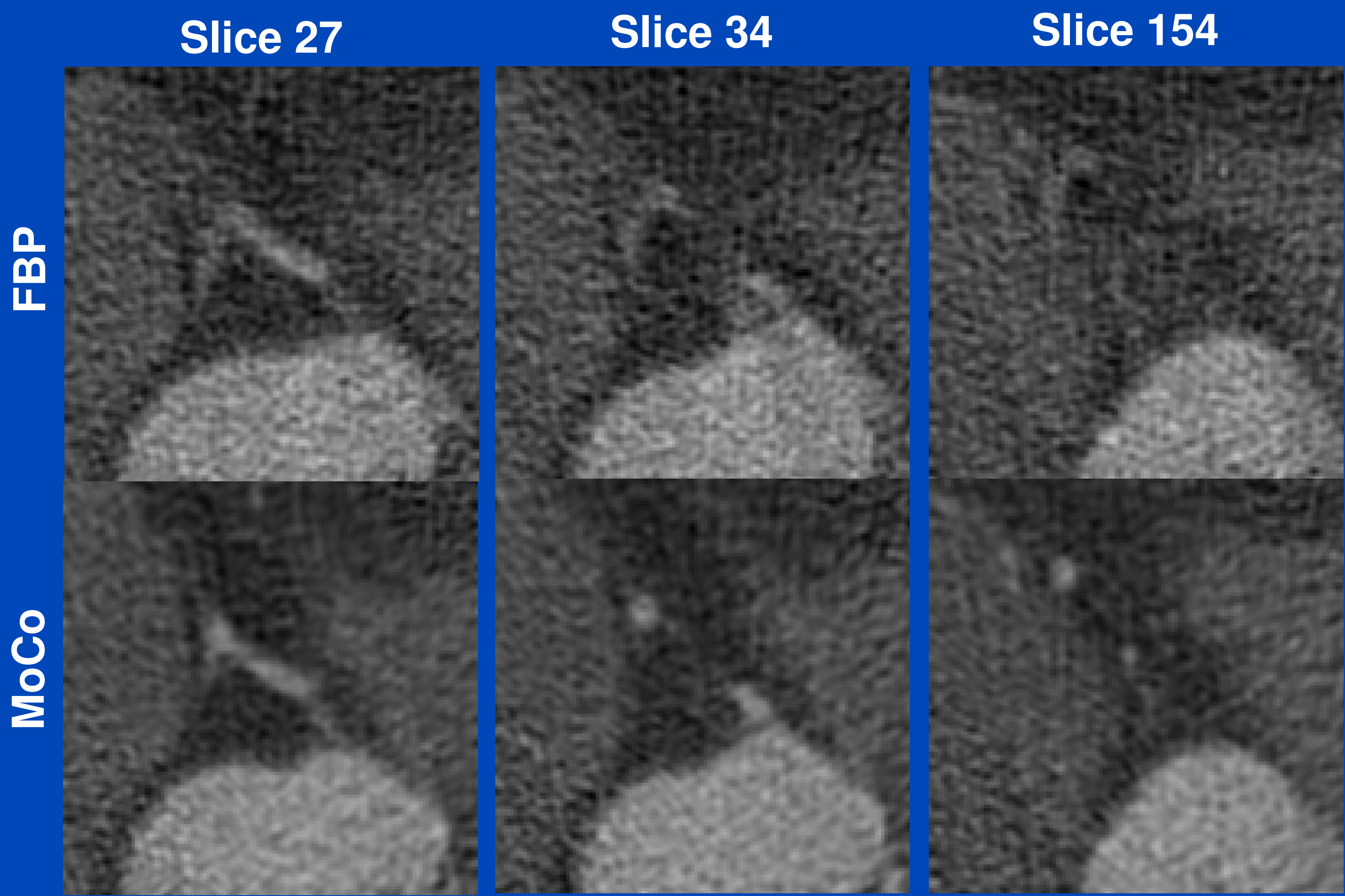
SIEMENS $\overline{HR}$ = 74 bpm, c = 30%,
C = 400 HU, W = 1500 HU

PAMoCo with $N_t \times N_\lambda \times 3 = 3 \times 3 \times 3 = 27$
parameter each stack

dkfz.

Stack 1

Patient 2



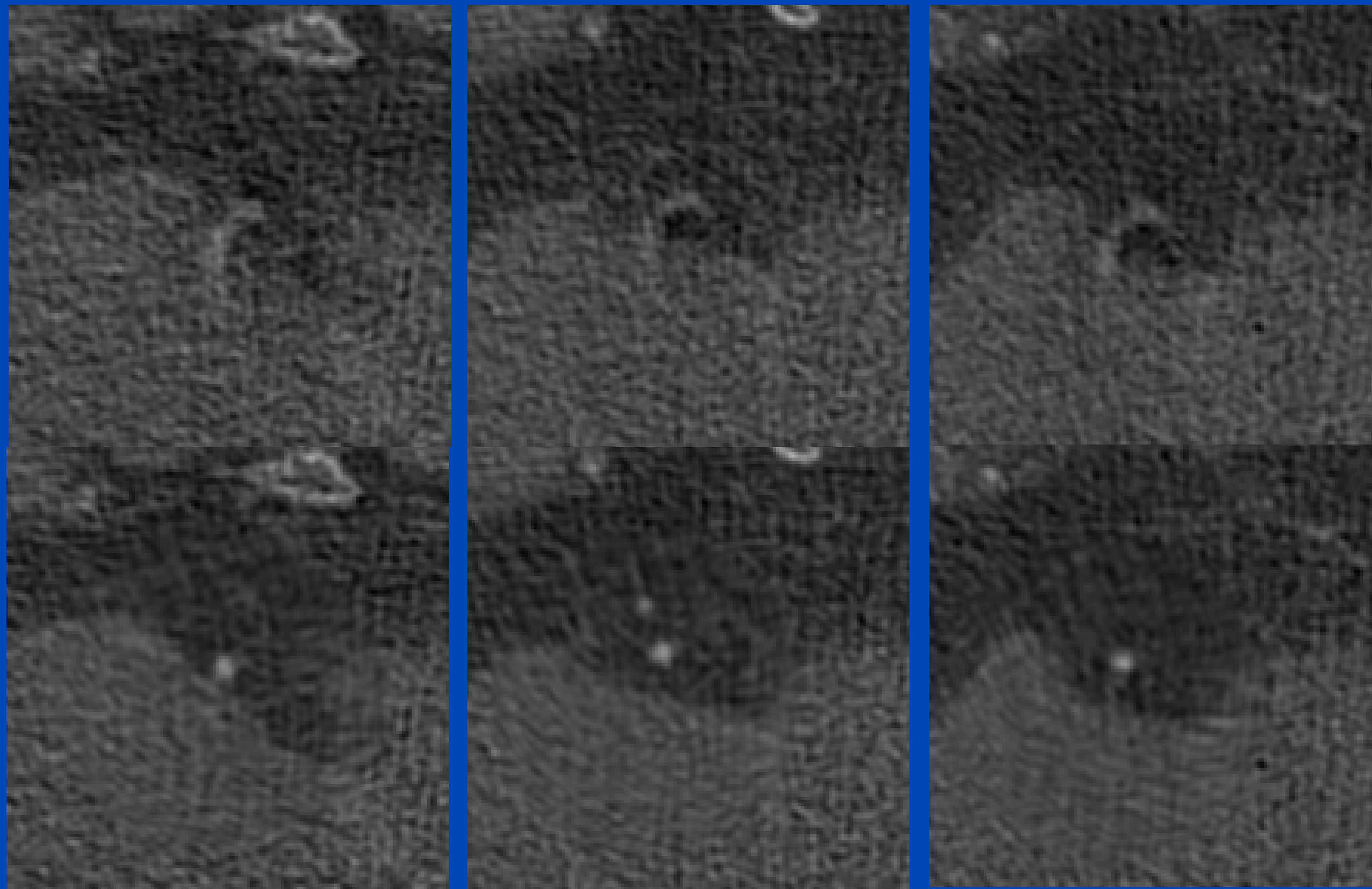
Stack 2

Patient 2

Slice 6 | Slice 37 | Slice 154

FBP

MoCo



$\overline{HR}$ = 70 bpm, c = 50%,
C = 400 HU, W = 1500 HU

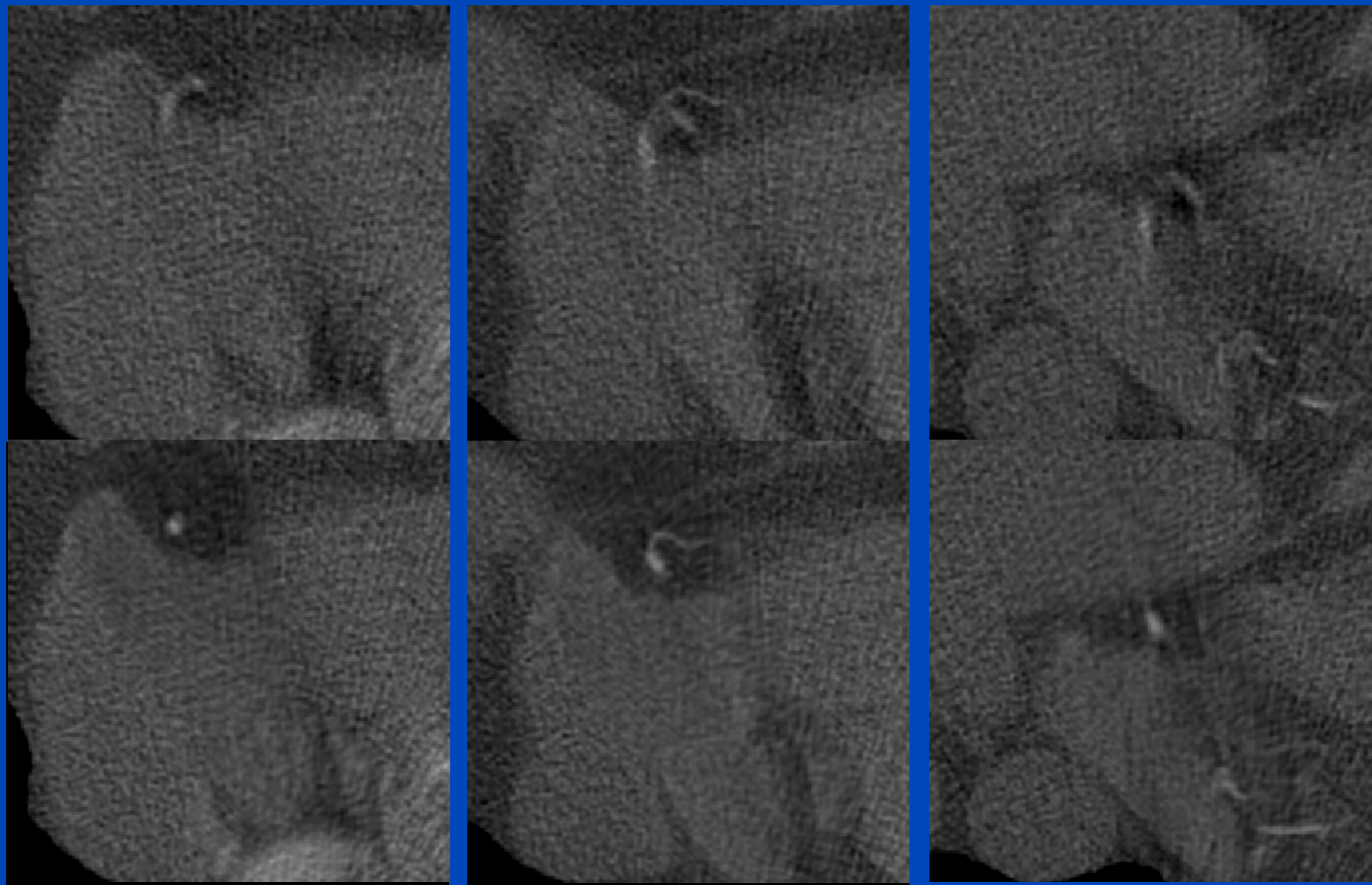
Stack 3

Patient 2

Slice 6 | Slice 37 | Slice 154

FBP

MoCo

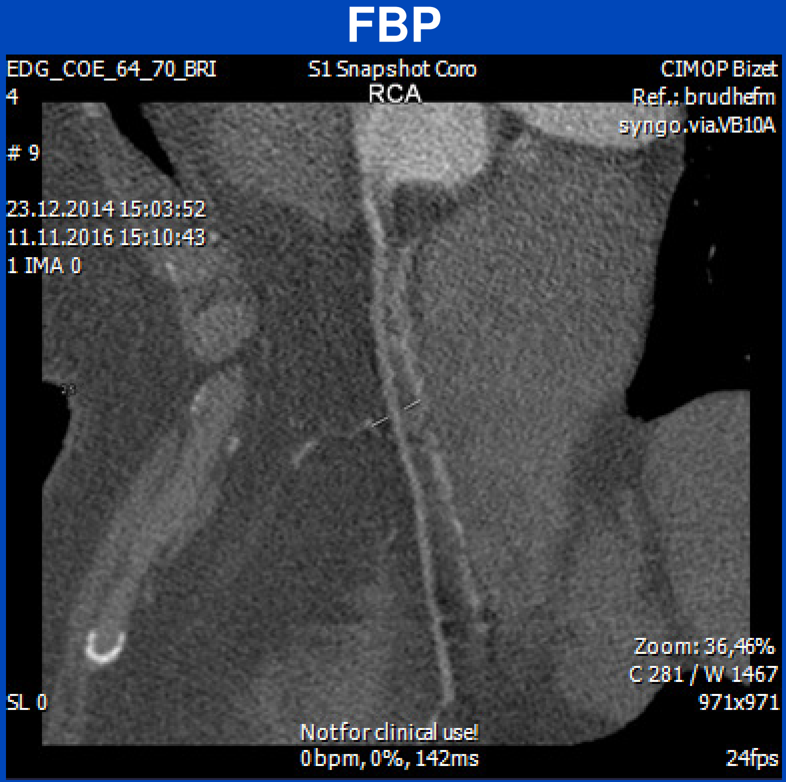


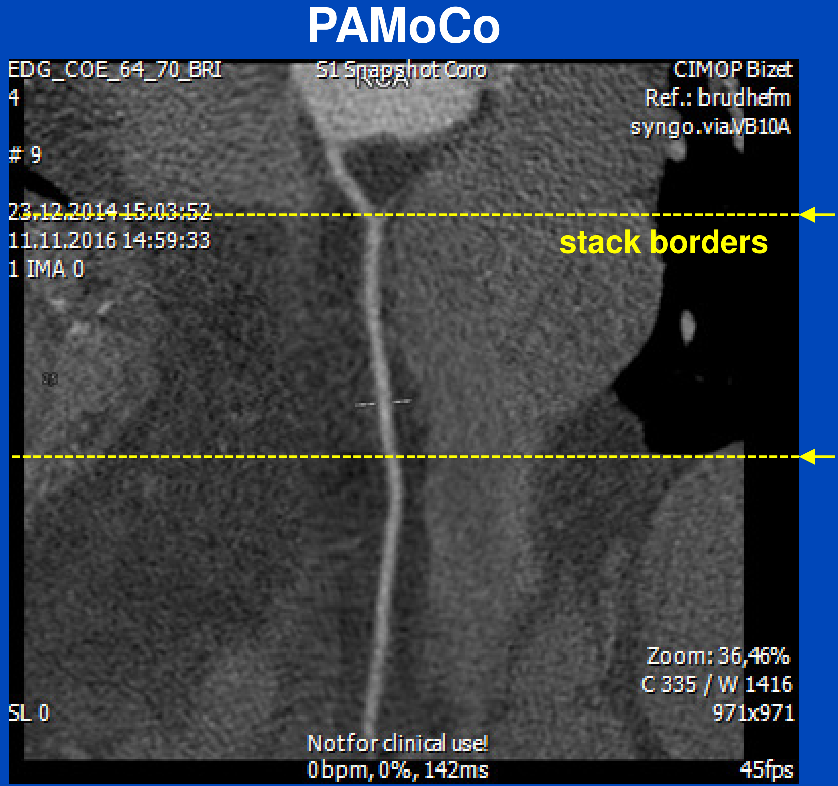
SIEMENS

$\overline{\mathrm{HR}}$ = 70 bpm, c = 50%,
C = 400 HU, W = 1500 HU

dkfz.

Patient 2

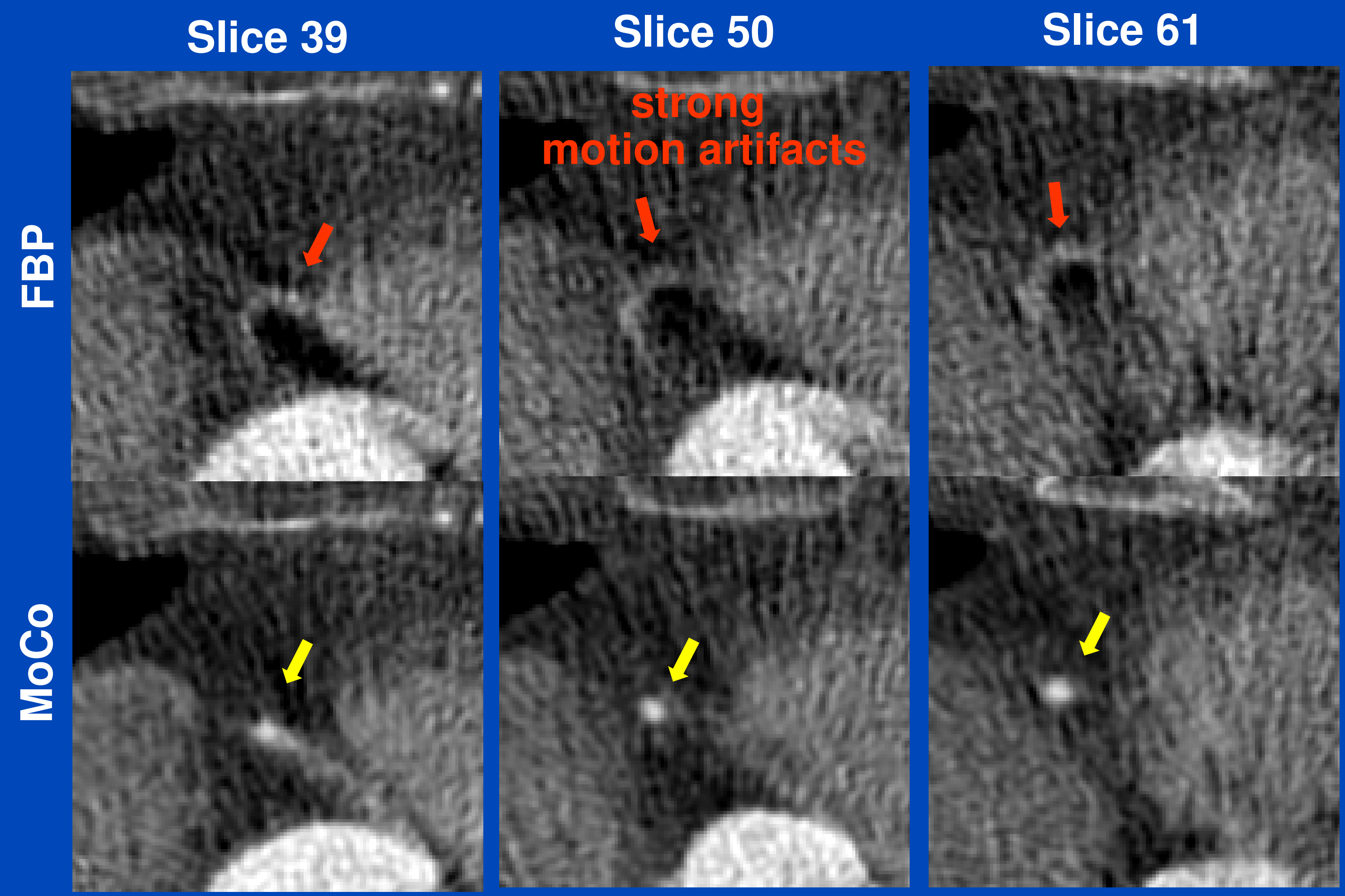




curved MPRs created with syngo.via

Stack 1

Patient 3



SIEMENS

$\overline{HR}$ = 69 bpm, c = 50%,
C = 400 HU, W = 1500 HU

dkfz.

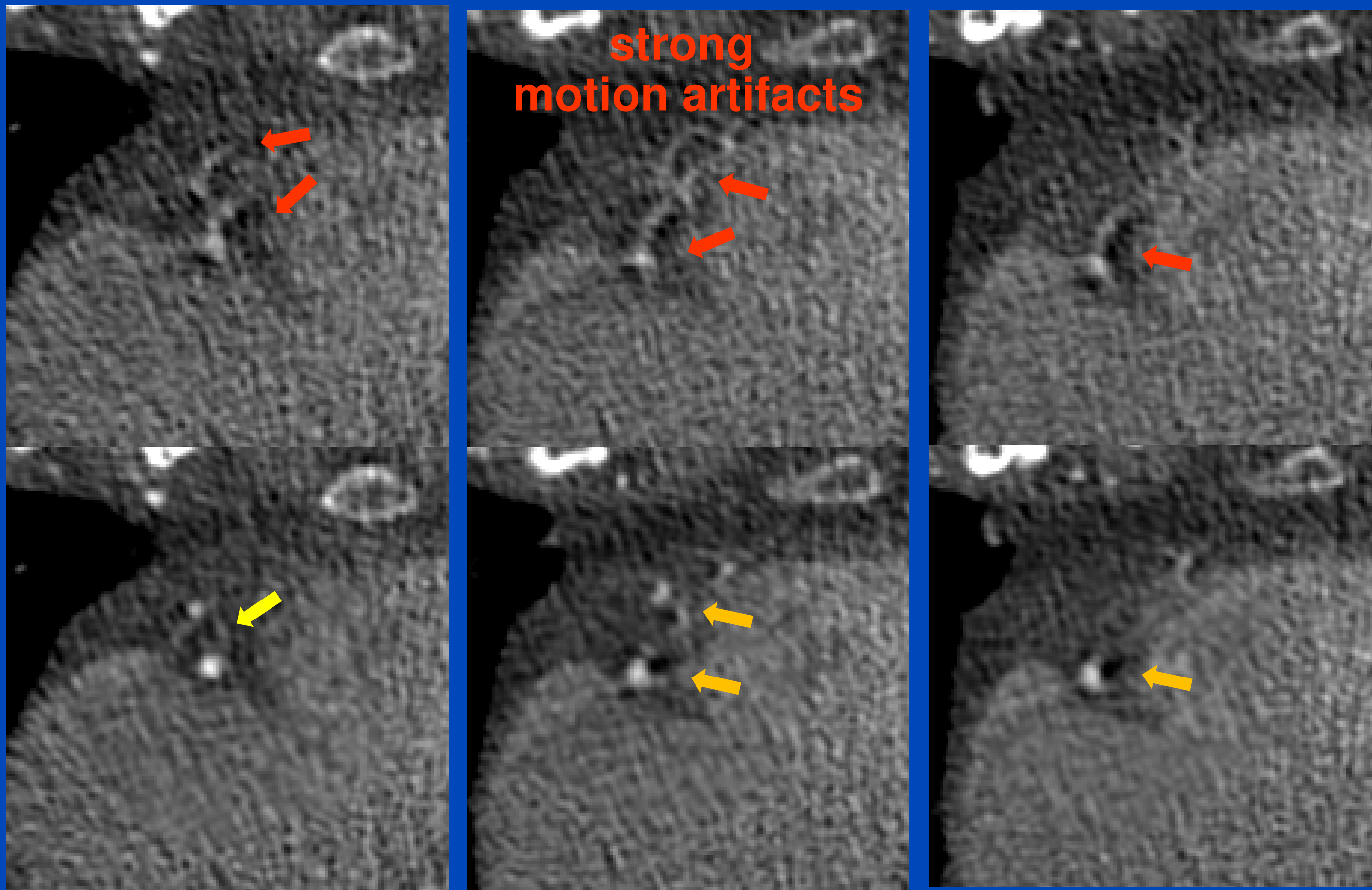
Stack 2

Patient 3

Slice 88 | Slice 107 | Slice 114

FBP

MoCo

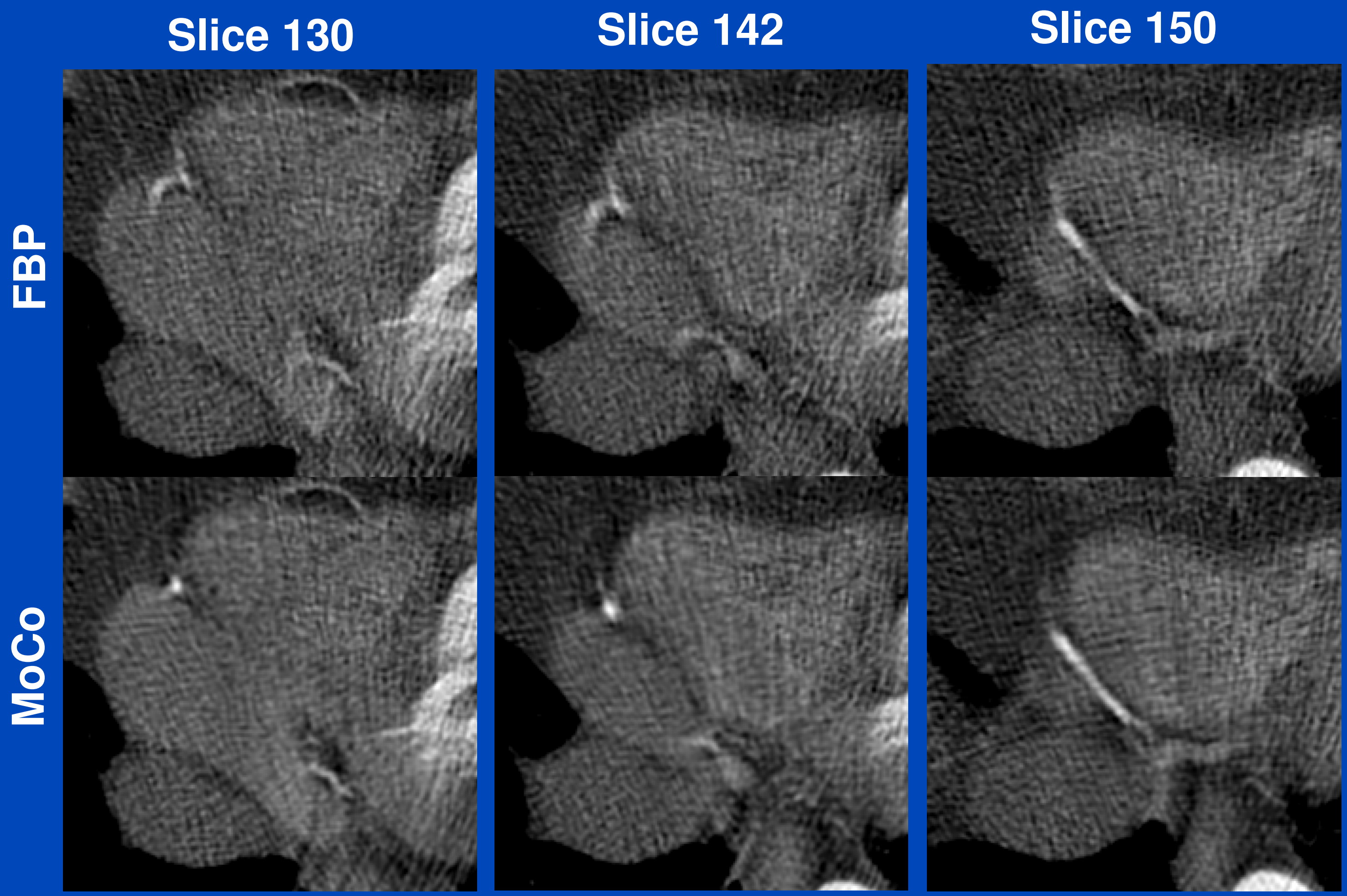


slight motion artifacts remain

$\overline{HR}$ = 69 bpm, c = 50%,
C = 400 HU, W = 1500 HU

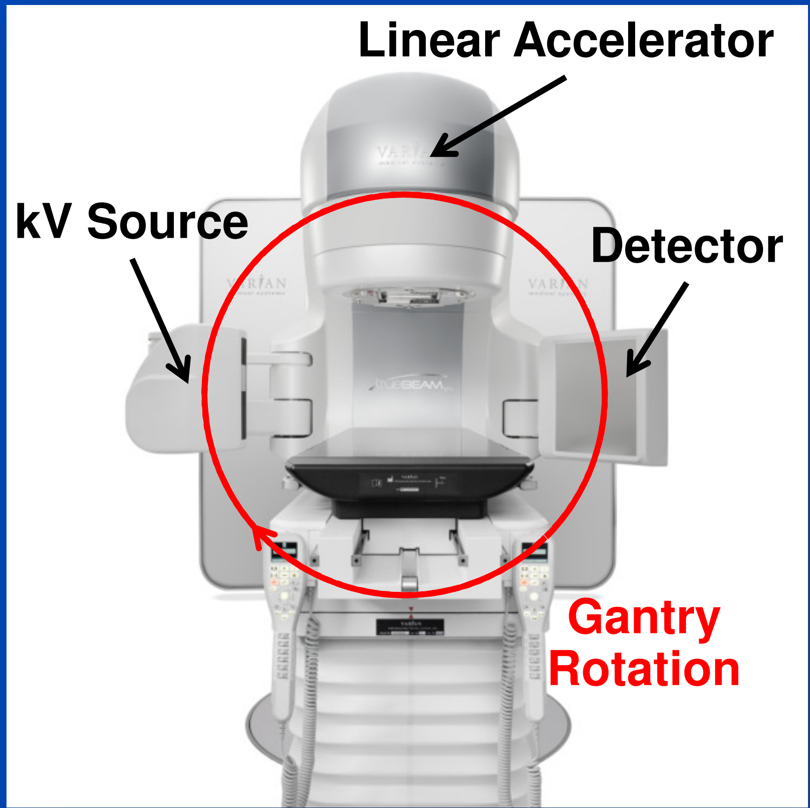
Stack 3

Patient 3

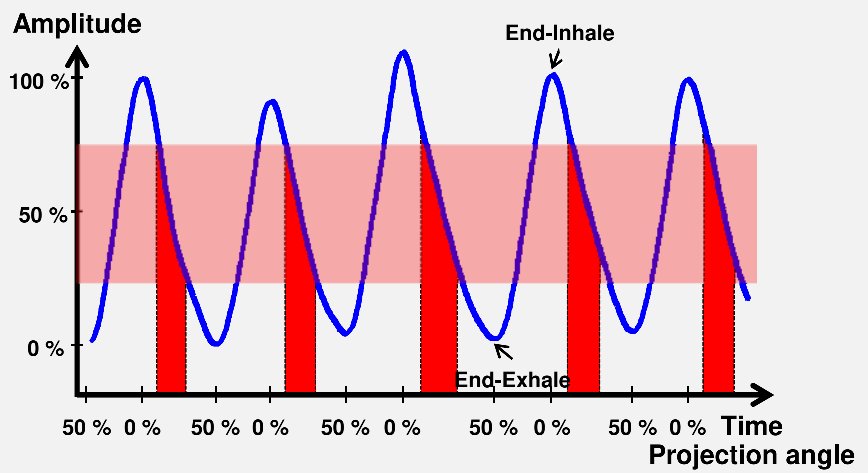


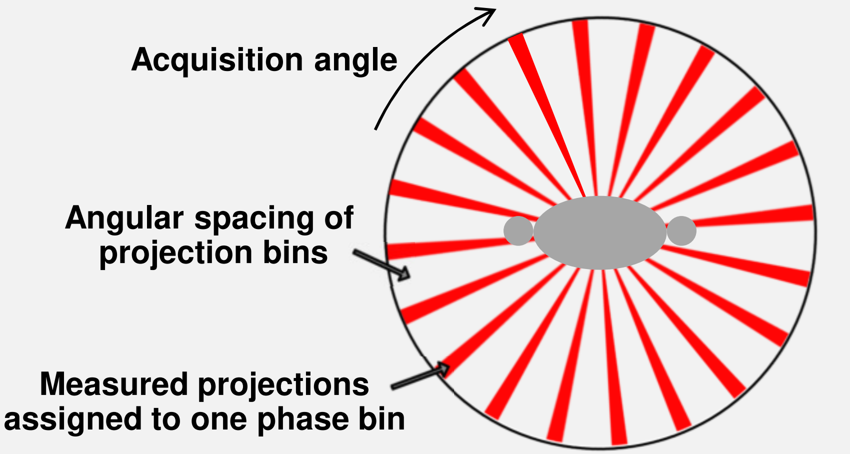
$\overline{HR}$ = 69 bpm, c = 50%,
C = 400 HU, W = 1500 HU

Motion Management for CBCT in IGRT



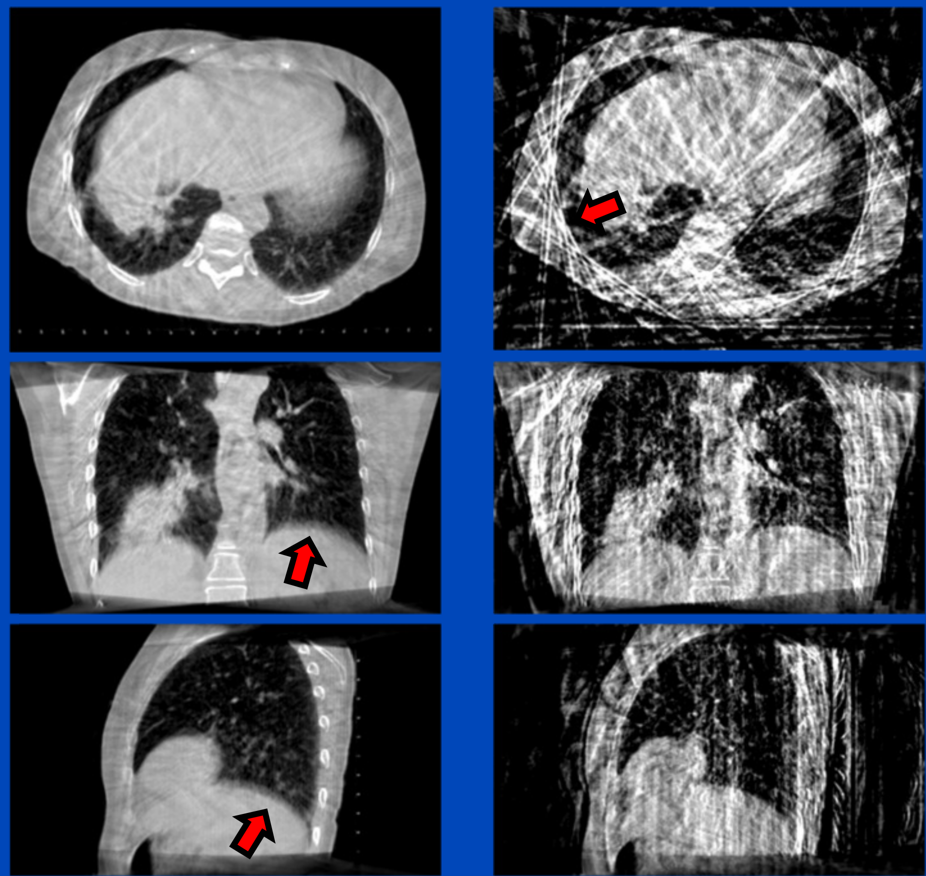
4D CBCT Scan with Retrospective Gating





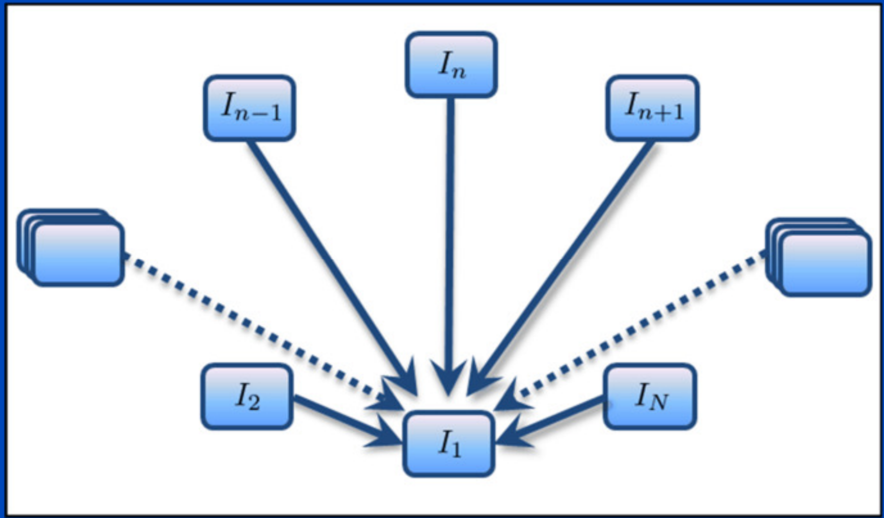
Without gating (3D): Motion artifacts

With gating (4D): Sparse-view artifacts

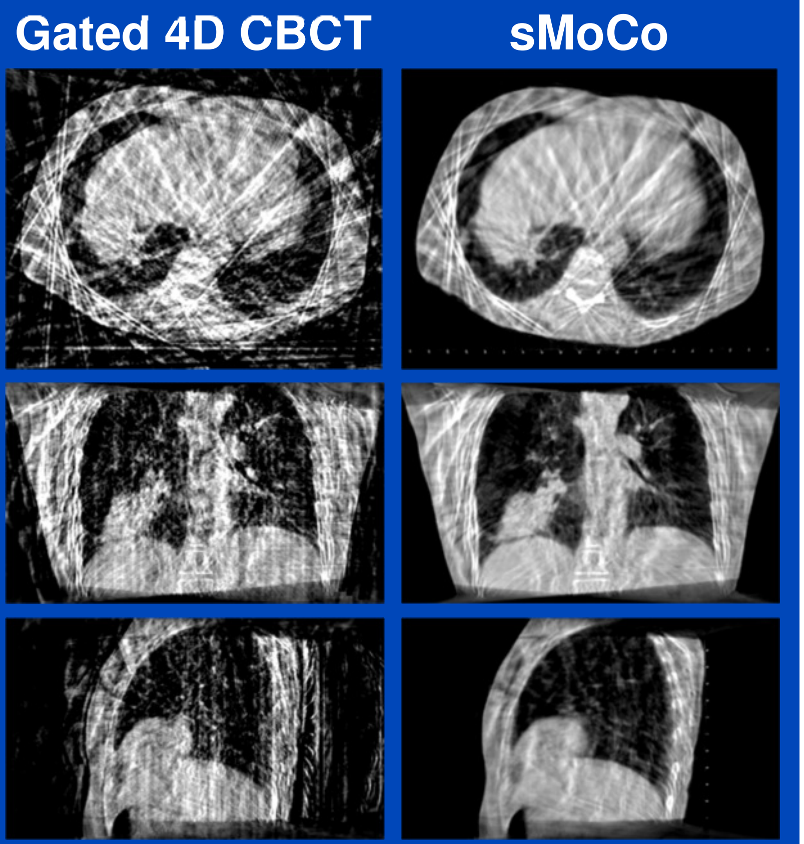


A Standard Motion Estimation and Compensation Approach (sMoCo)

- Motion estimation via standard 3D-3D registration



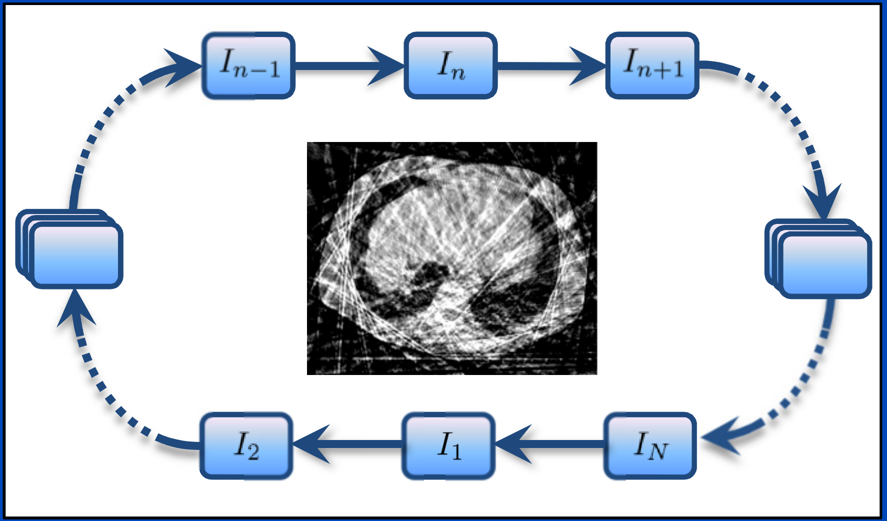
- Has to be repeated for each reconstructed phase

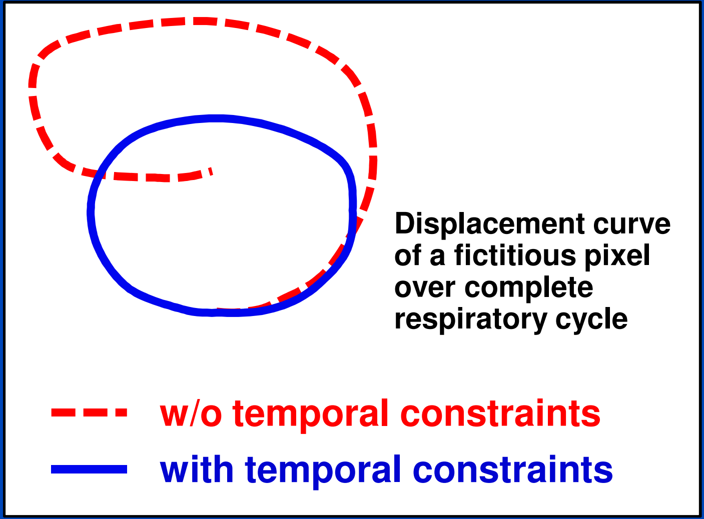


- Streak artifacts from gated reconstructions propagate into sMoCo results

The Cyclic Motion Estimation and Compensation Approach (cMoCo)

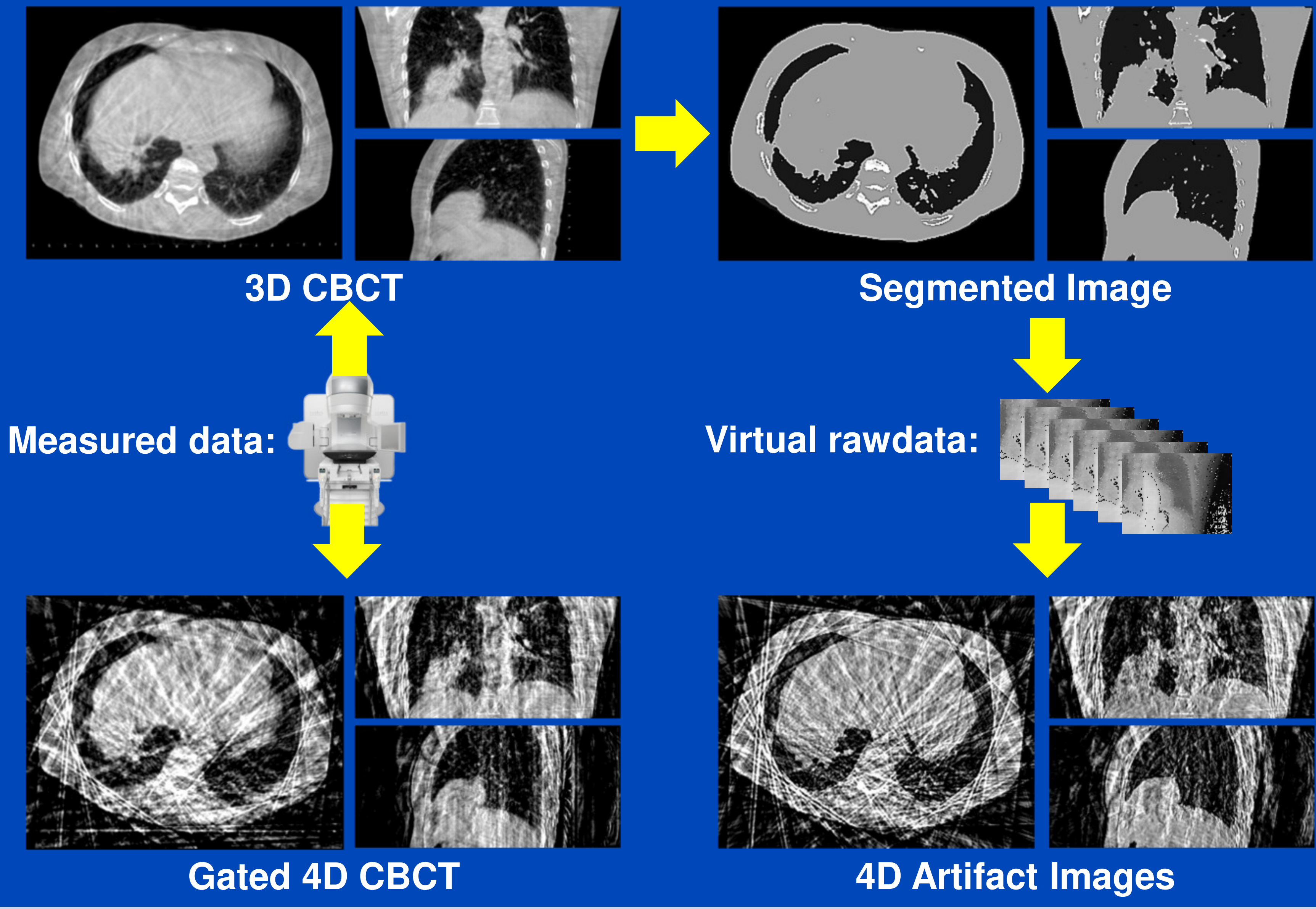
- Motion estimation only between adjacent phases
- Incorporate additional knowledge
  - A priori knowledge of quasi periodic breathing pattern
  - Non-cyclic motion is penalized
  - Error propagation due to concatenation is reduced





Brehm, Paysan, Oelhafen, Kunz, and Kachelrieß, "Self-adapting cyclic registration for motion-compensated cone-beam CT in image-guided radiation therapy," Med. Phys. 39(12):7603-7618, 2012.

Artifact Model-Based MoCo (aMoCo)



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Brehm, Paysan, Oelhafen, and Kachelrieß, "Artifact-resistant motion estimation with a patient-specific artifact model for motion-compensated cone-beam CT" Med. Phys. 40(10):101913, 2013.

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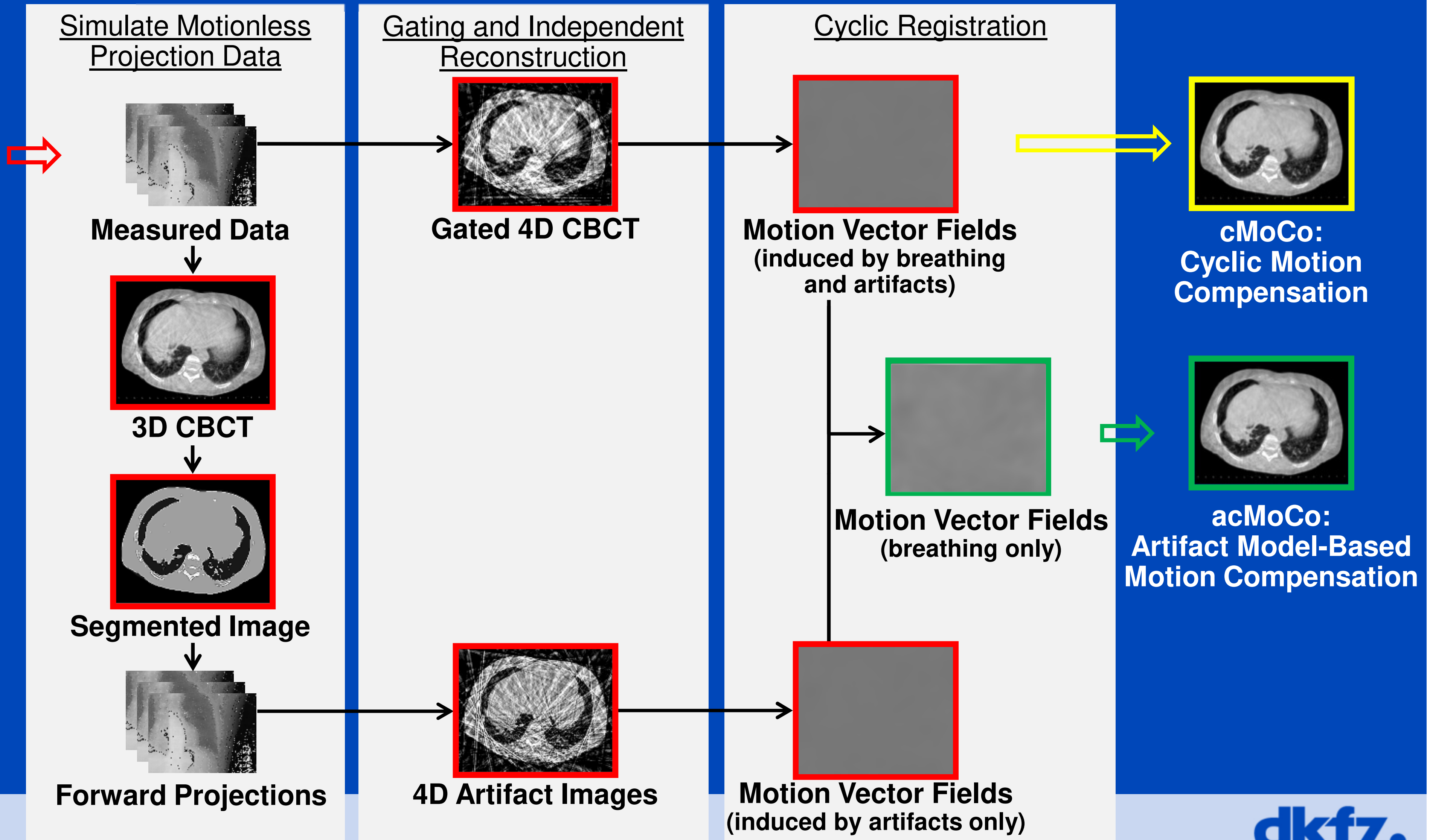
Motion Estimation using a Patient-Specific Artifact Model



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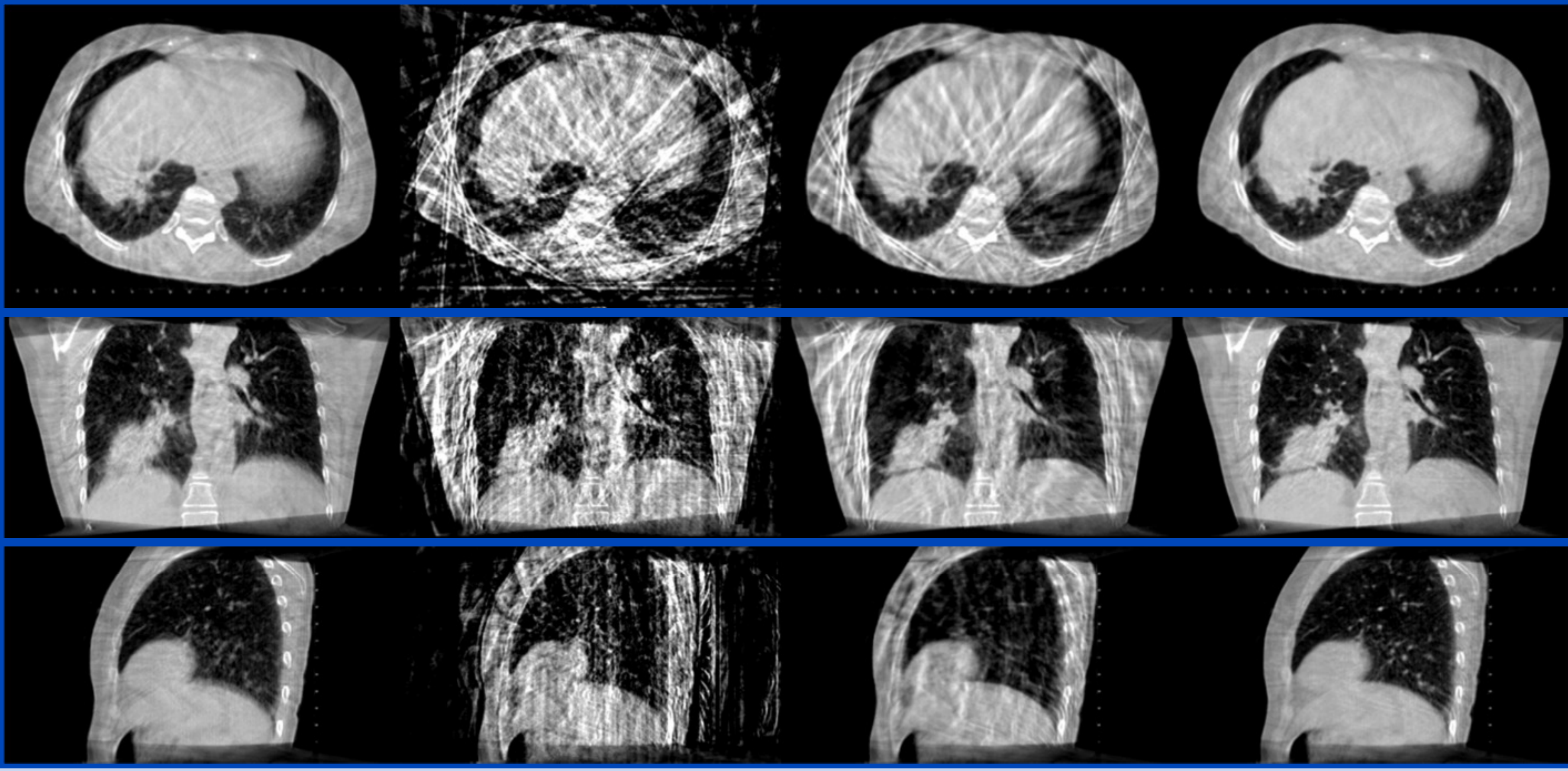
Patient Data – Results

3D CBCT	Gated 4D CBCT	sMoCo	acMoCo
Standard	Conventional Phase-Correlated	Standard Motion Compensation	Artifact Model-Based Motion Compensation



C = -200 HU, *W* = 1400 HU, displayed with 30 rpm.
Patient data provided by Memorial Sloan–Kettering Cancer Center, New York, NY.

Spin-Off Effects?

4D PET/MR Motion Compensation

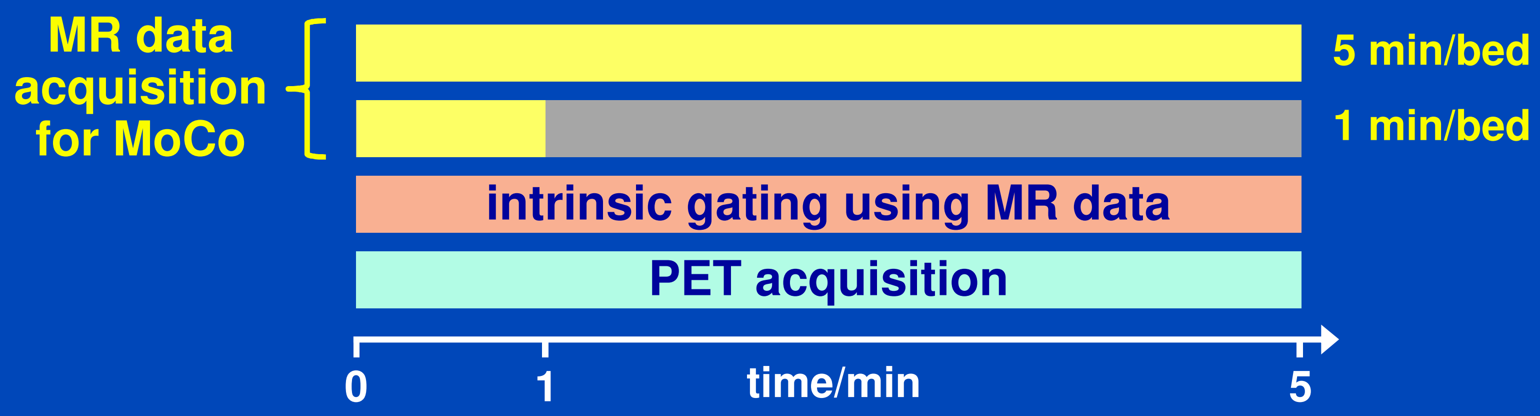
Data Acquisition and Processing

- Simultaneous PET/MR acquisition at Siemens Biograph mMR at DKFZ
  - number of subjects: 5 (thorax), 2 (abdomen)
  - tracer: fluorodeoxyglucose (^{18}F-FDG)
  - acquisition time per bed: 5 min
  - MR sequence: 3D-encoded gradient echo sequence with radial stack-of-stars sampling scheme and golden angle radial spacing
  - pre-processing of PET list-mode data
    - » sorting of list-mode data into sinograms for different motion phases with binning tools
    - » scatter estimation with e7-tools
  - in-house cMoCo OSEM algorithm for reconstruction

4D PET/MR Motion Compensation

Generation of Highly Undersampled MR Data Set

- Retrospective generation of a sparse MR rawdata set reproducing an interlaced MR acquisition



- Intrinsic gating: motion amplitudes were estimated from measured MR data
- MR and PET data were sorted retrospectively into 20 overlapping motion phase bins (10% width)

MoCo PET Image Reconstruction[1]

- **MoCo MLEM update equation of motion phase *i*:**

$$\lambda_i^{(n+1)} = \lambda_i^{(n)} \frac{1}{\sum_{i'} T_{i'}^{i} M^{\mathrm{T}} \frac{1}{a_{i'}}} \sum_{i'} T_{i'}^{i} M^{\mathrm{T}} \frac{p_{i'}}{M T_i^{i'} \lambda_i^{(n)} + a_{i'}(r_{i'} + s_{i'})}$$

n:	iteration index
M, M^{T}:	system matrix including forward-/backprojection
a:	attenuation correction factors
p:	measured rawdata (prompts)
r:	estimated randoms
s:	estimated scatter
$\lambda^{(n)}$:	image estimate at iteration n
i, i':	indices of motion phases
$T_i^{i'}$:	warping operation mapping motion phase i to i'

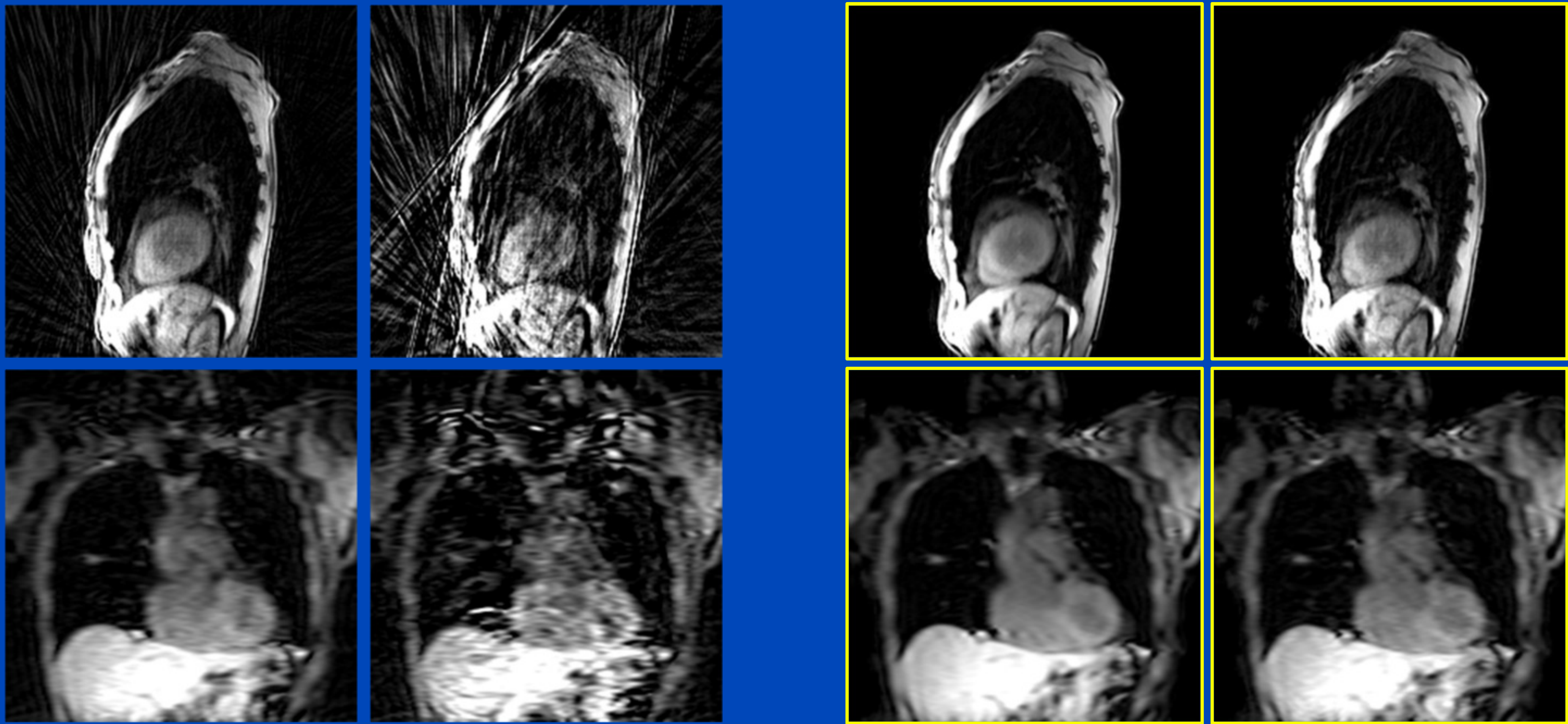
- **To reduce computation time, an ordered subset implementation (OSEM) was used**

[1] Qiao, Pan, Clark, Mawlawi. A motion-incorporated reconstruction method for gated PET studies. Phys. Med. Biol. 2006.

4D PET/MR Motion Compensation

MR Results Patient s04

4D gated gridding		4D cMoCo	
5 min / bed	1 min / bed	5 min / bed	1 min / bed

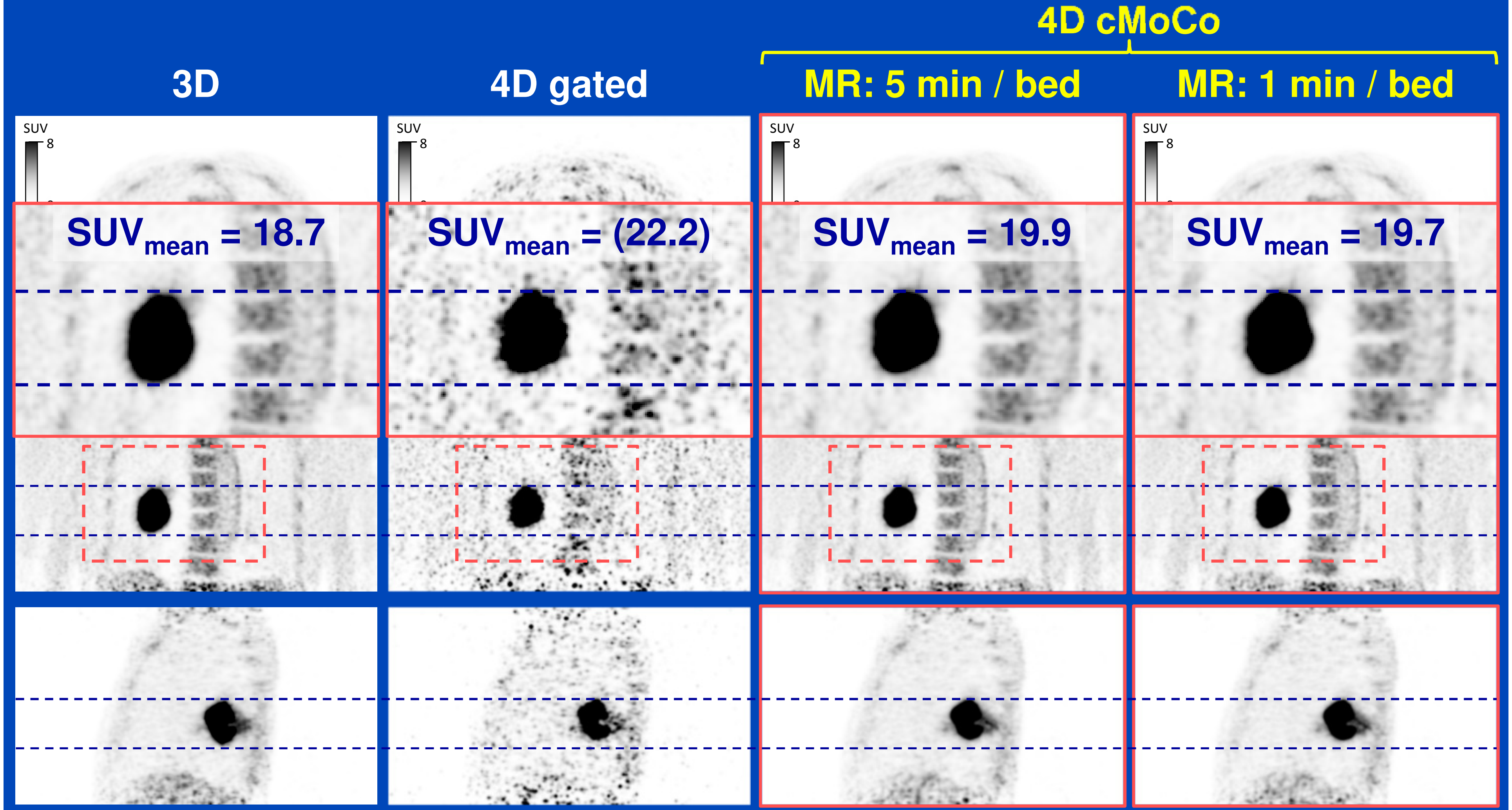


		MVFs	MVFs

Rank, Heußer, Buzan, Wetscherek, Freitag, Dinkel, Kachelrieß. 4D respiratory motion-compensated image reconstruction of free-breathing radial MR data with very high undersampling. Magn Reson Med 77(3):1170-1183, 2017.

4D PET/MR Motion Compensation

PET Results Patient s01

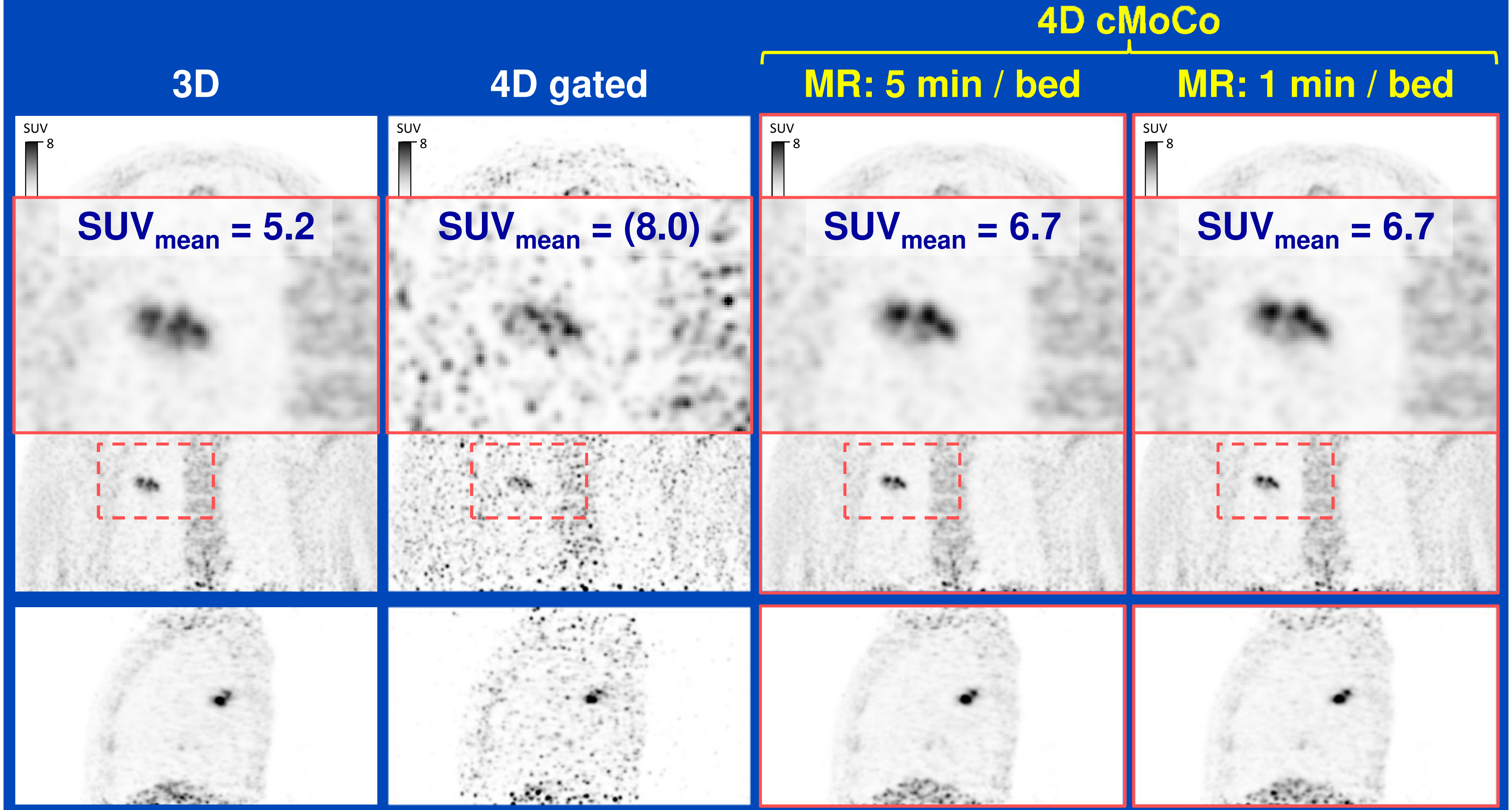


due to the high noise level of 4D gated PET, SUV_{mean} was systematically overestimated

C. Rank, T. Heußer, A. Wetscherek, M. Freitag, O. Sedlaczek, H.-P. Schlemmer, and M. Kachelrieß. Respiratory motion compensation for simultaneous PET/MR based on highly undersampled MR data. Med. Phys. 43(12):6234-6245, December 2016.

4D PET/MR Motion Compensation

PET Results Patient s09



due to the high noise level of 4D gated PET,
SUV_{mean} was systematically overestimated

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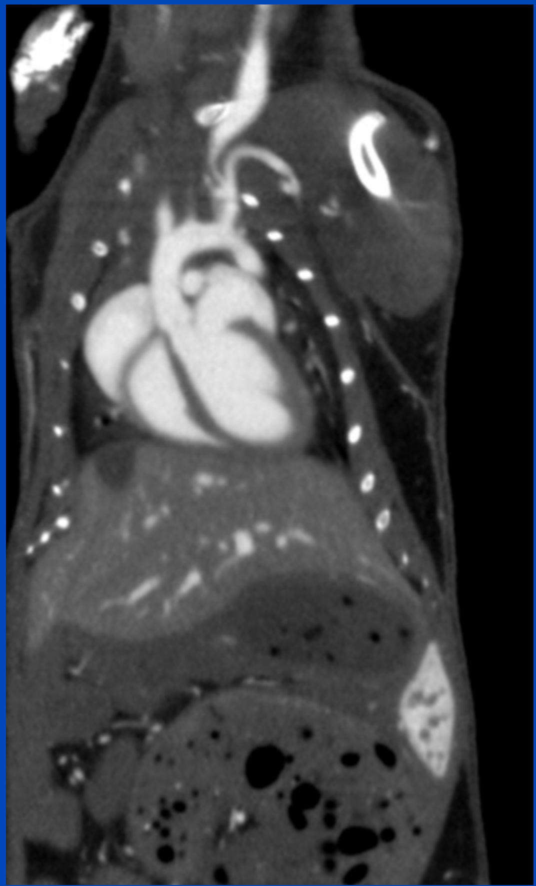
Is There More?

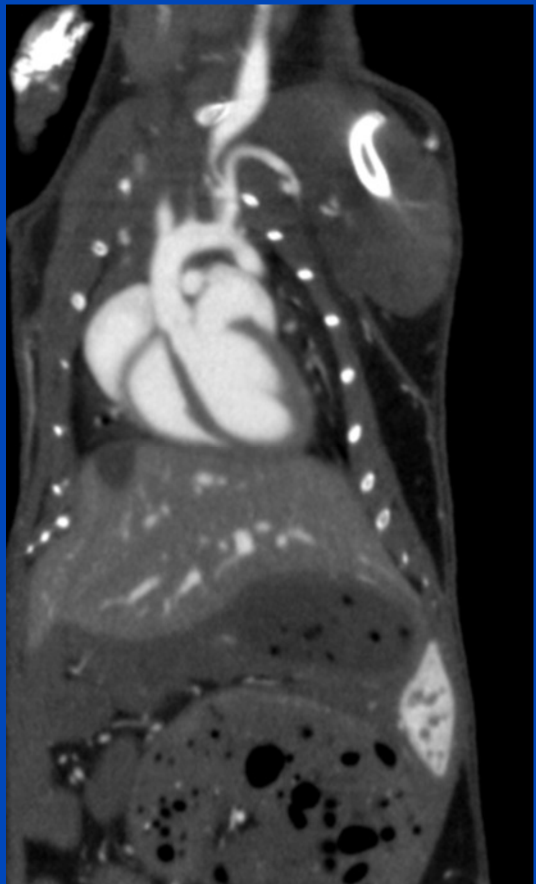


Data displayed as:

Heart: 280 bpm

Lung: 150 rpm

Mouse with 150 rpm and 280 bpm.

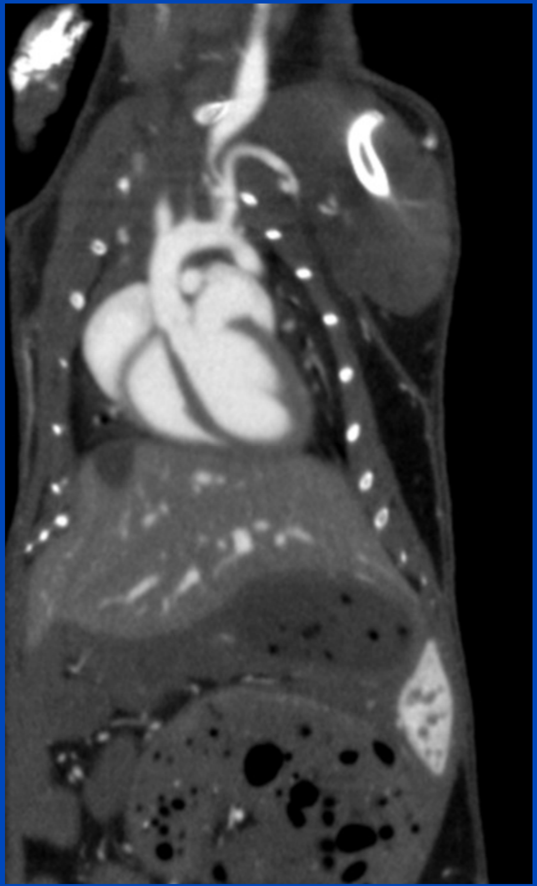


Data displayed as:

Heart: 180 bpm

Lung: 90 rpm

Mouse with 180 rpm and 240 bpm.



<u>**Data displayed as:**</u>

Heart: 90 bpm

Lung: 90 rpm

Mouse with 180 rpm and 240 bpm.

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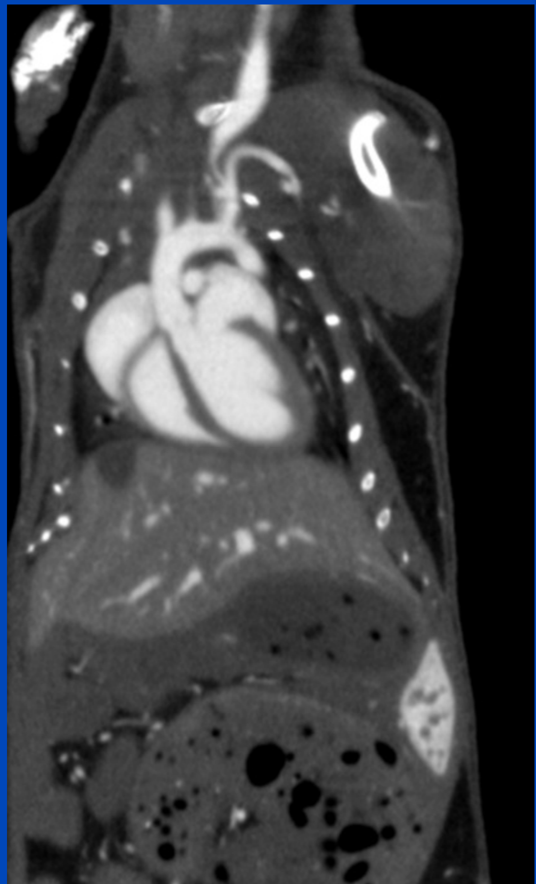
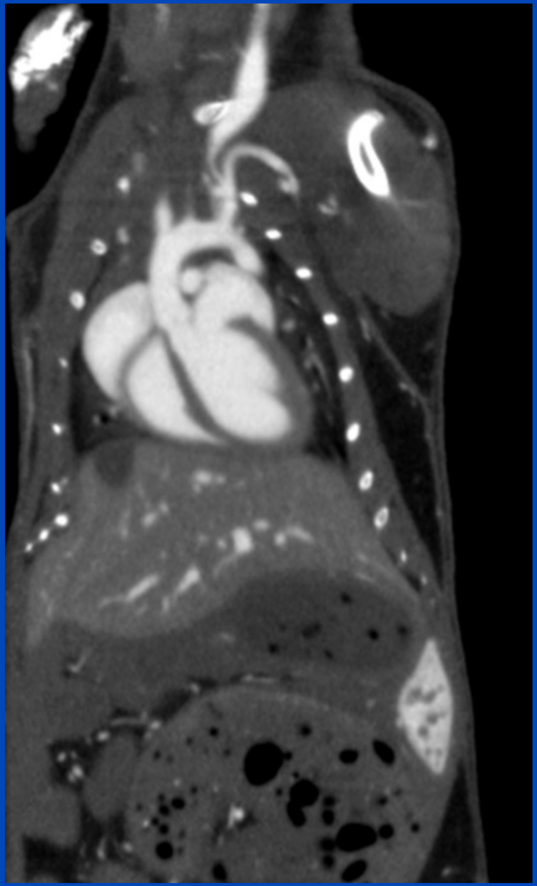


Data displayed as:

Heart: 0 bpm

Lung: 90 rpm

Mouse with 180 rpm and 240 bpm.



Data displayed as:

Heart: 90 bpm

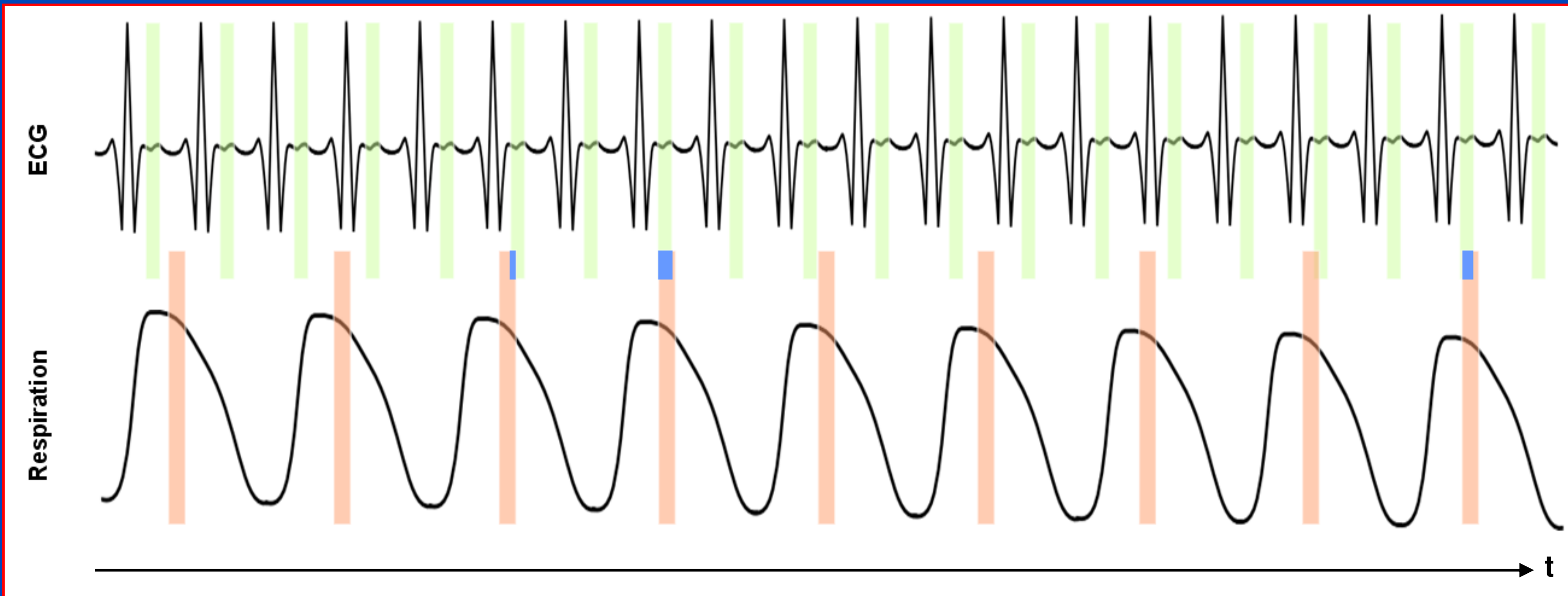
Lung: 0 rpm

Mouse with 180 rpm and 240 bpm.

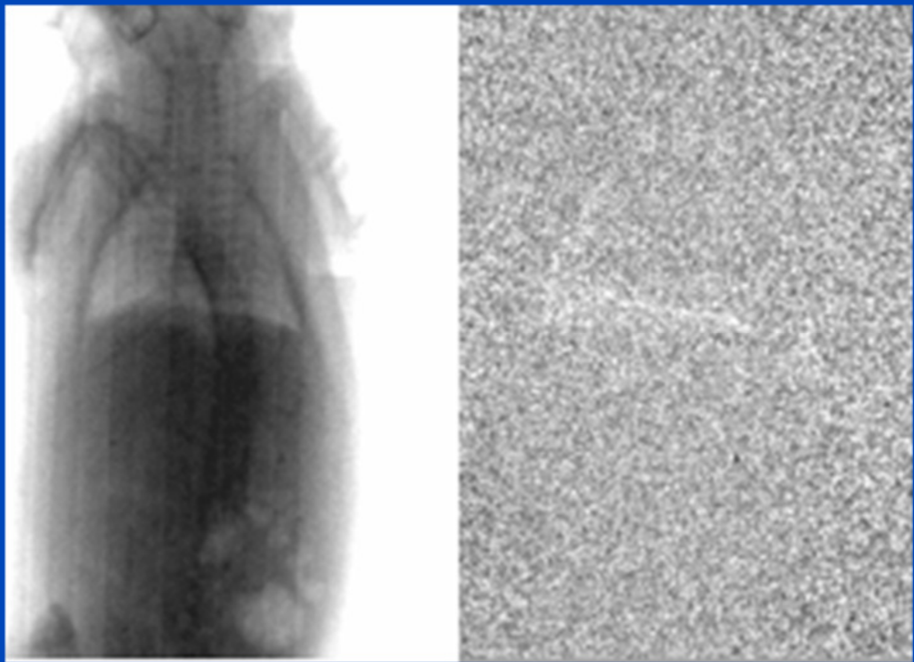
5D with Double Gating?

Double gating example:

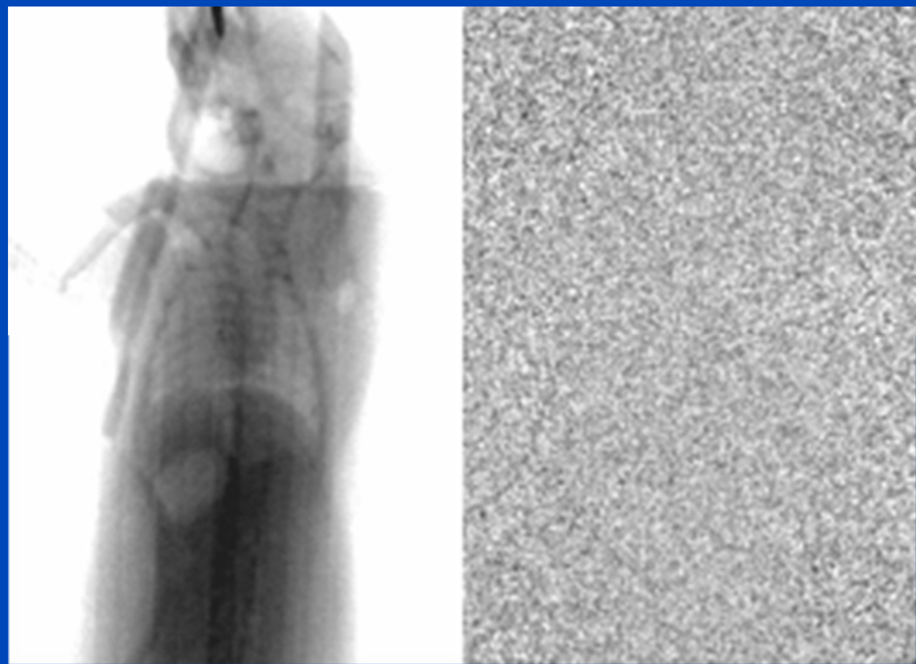
• Cardiac window width: 20%
• Respiratory window width: 10%
• Only 2% of all projections per reconstructed volume



Injection Techniques[1]



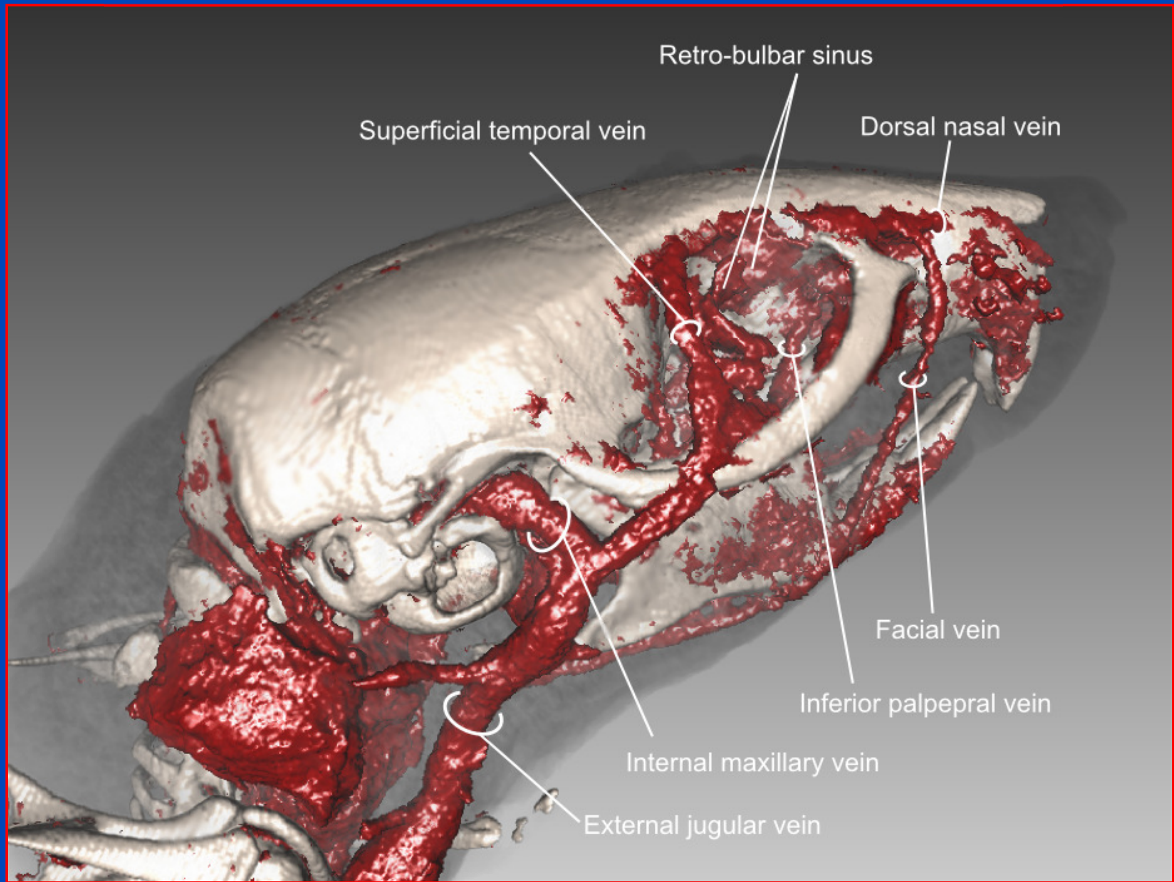
Tail Vein Injection



Retro Bulbar Injection

[1] M. Socher, J. Kuntz, S. Sawall, S. bartling, and M. Kachelrieß. The retrobulbar sinus is superior to the lateral tail vein for the injection of contrast media in small animal cardiac imaging. Lab. Anim. 48(2), pp. 105-113, February 2014.

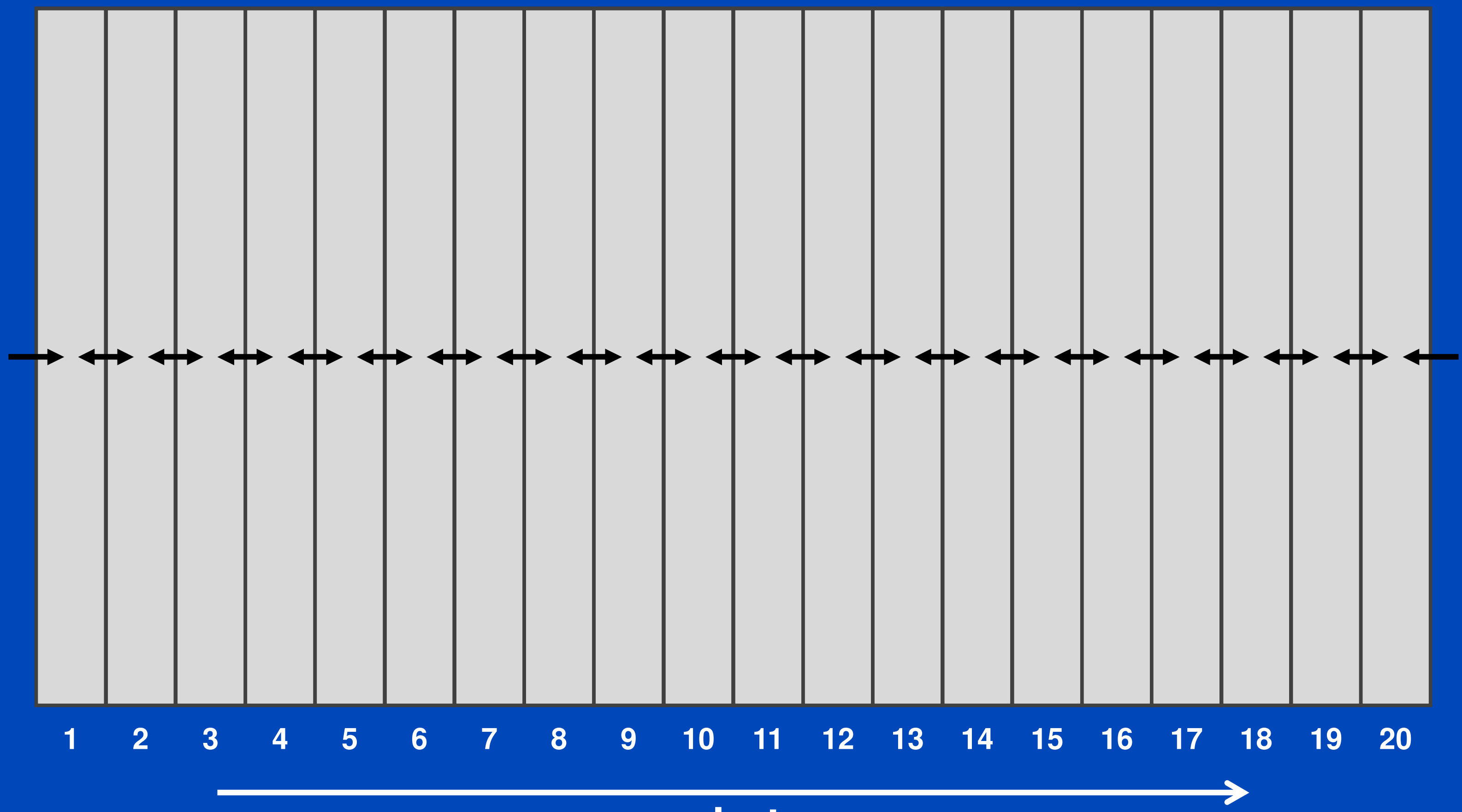
Contrast Injection[1]



Volume rendering of a high resolution micro-CT scan with a spatial resolution of about 40 µm.

[1] M. Socher, J. Kuntz, S. Sawall, S. bartling, and M. Kachelrieß. The retrobulbar sinus is superior to the lateral tail vein for the injection of contrast media in small animal cardiac imaging. Lab. Anim. 48(2), pp. 105-113, February 2014.

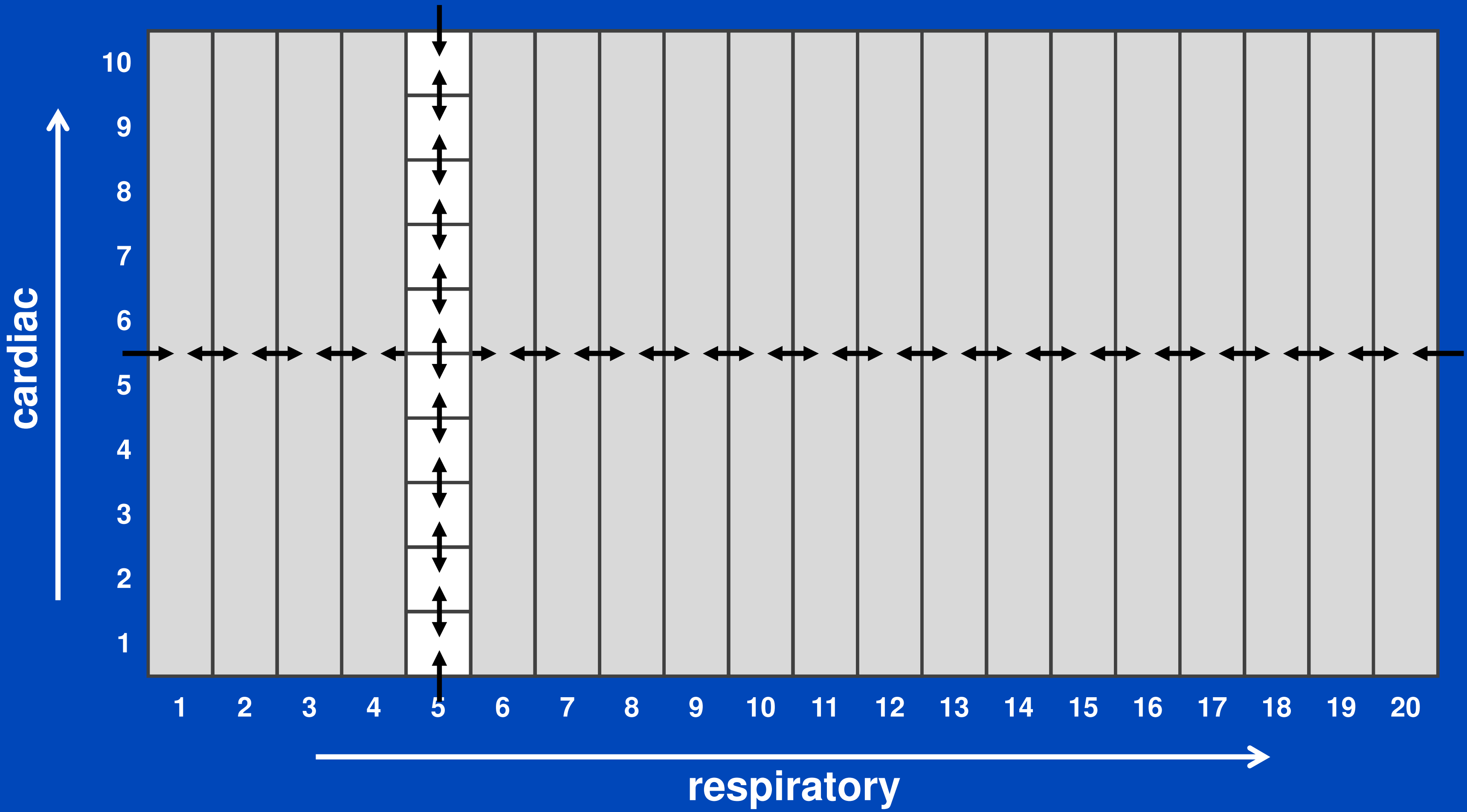
5D Motion Compensation



Brehm, Sawall, Maier, and Kachelrieß, “Cardio-respiratory motion-compensated micro-CT image reconstruction using an artifact model-based motion estimation” Med. Phys. 42(4):1948-1958, 2015.

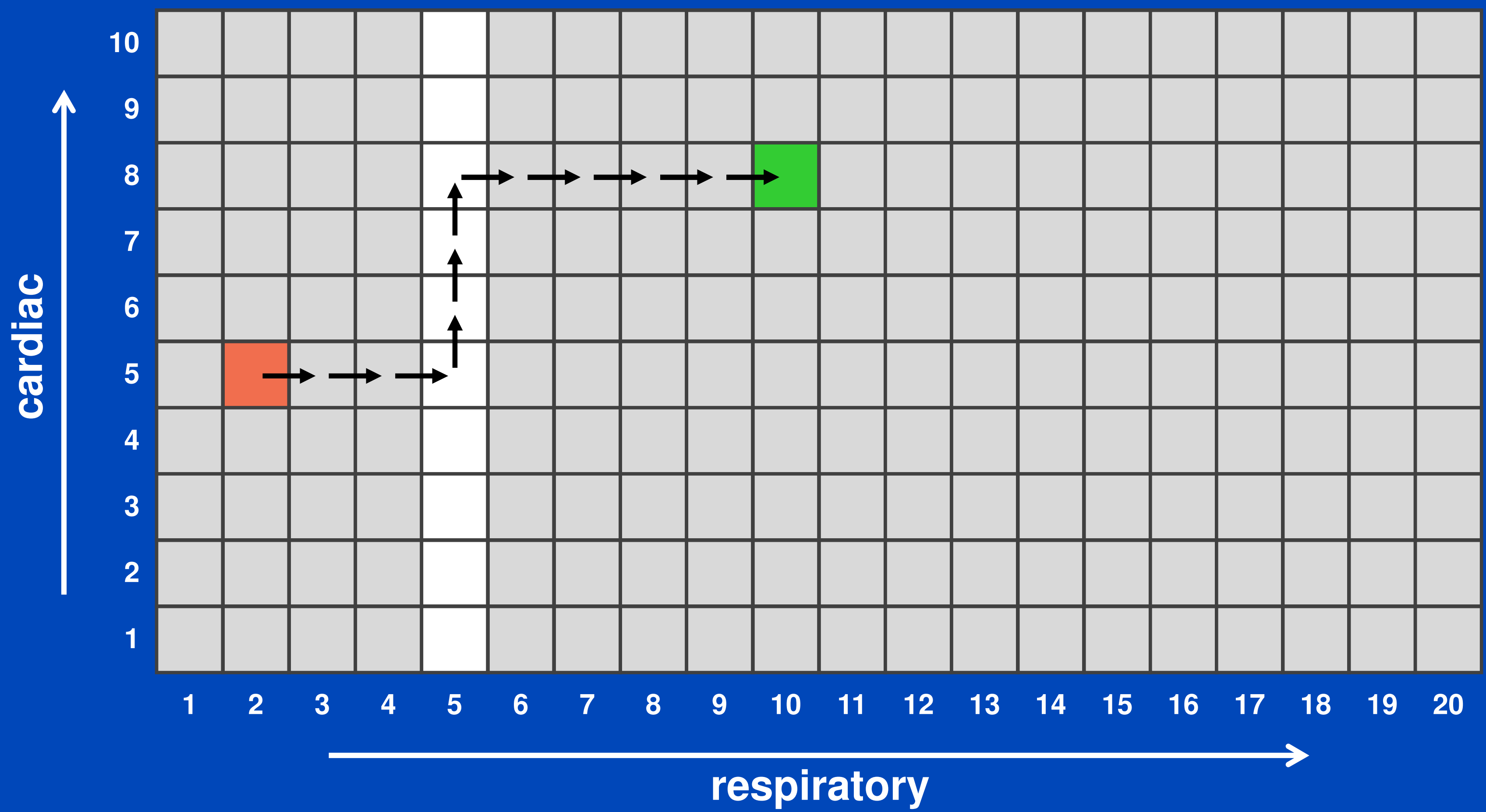
dkfz.

5D Motion Compensation



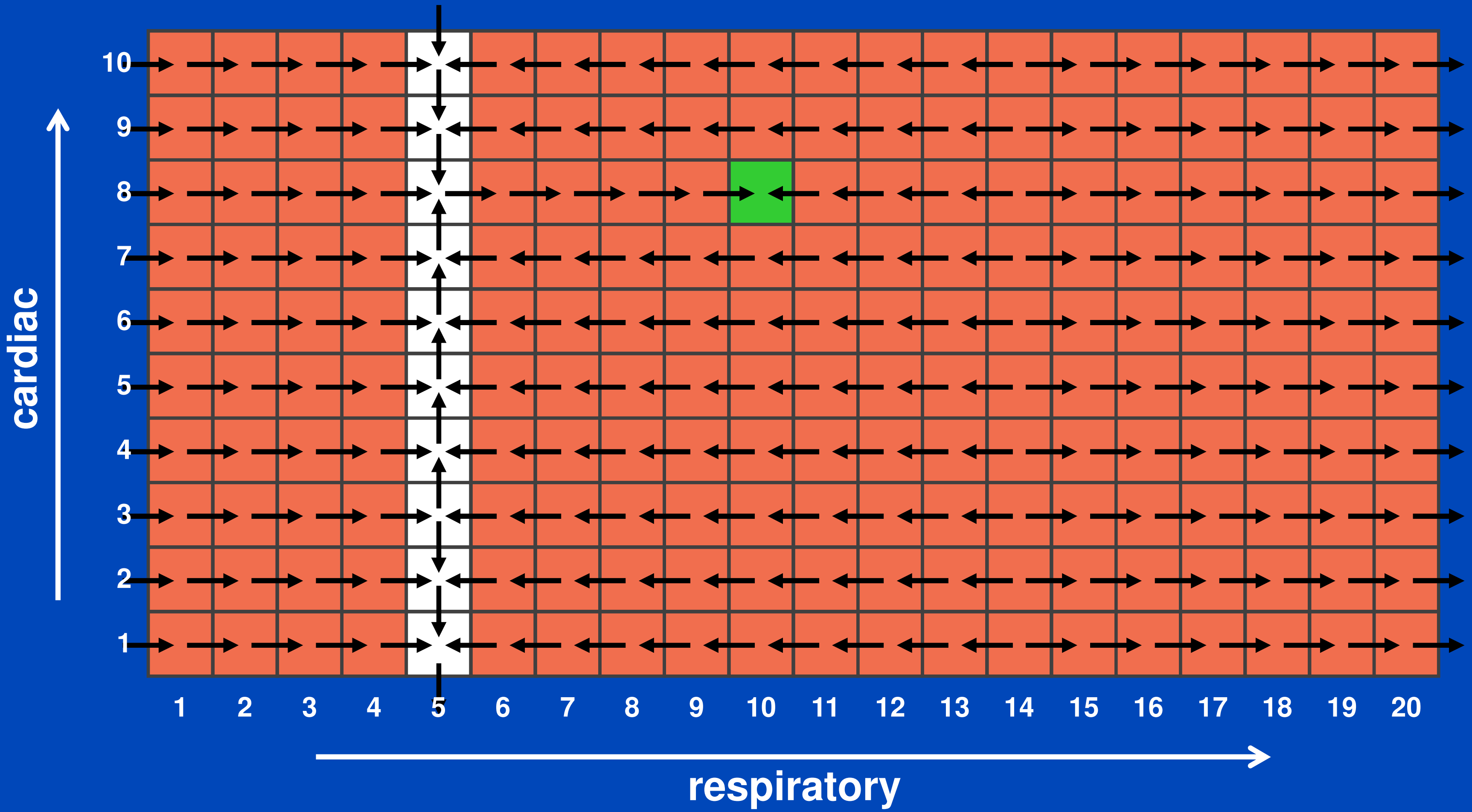
Brehm, Sawall, Maier, and Kachelrieß, "Cardio-respiratory motion-compensated micro-CT image reconstruction using an artifact model-based motion estimation" Med. Phys. 42(4):1948-1958, 2015.

5D Motion Compensation



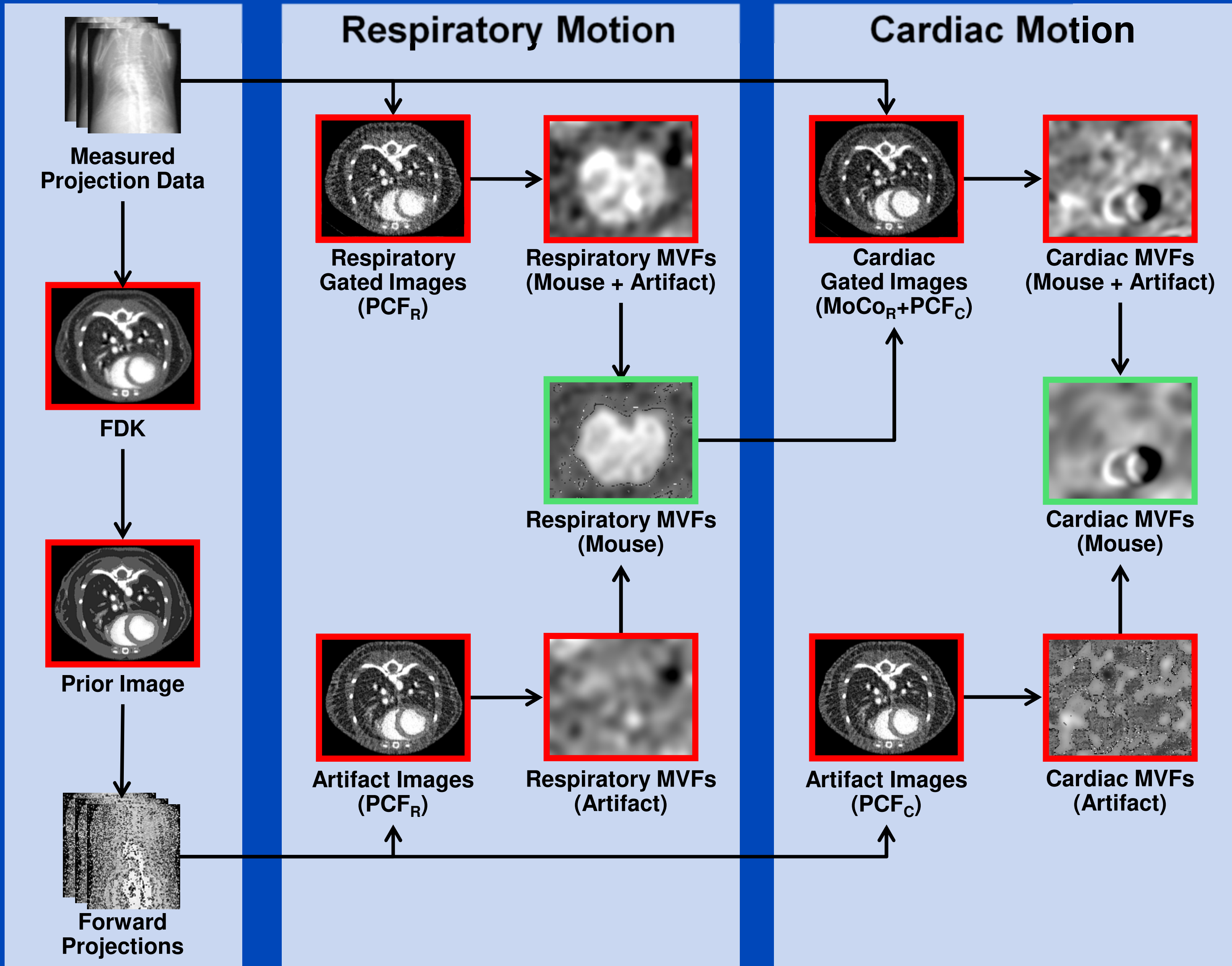
Brehm, Sawall, Maier, and Kachelrieß, “Cardio-respiratory motion-compensated micro-CT image reconstruction using an artifact model-based motion estimation” Med. Phys. 42(4):1948-1958, 2015.

5D Motion Compensation

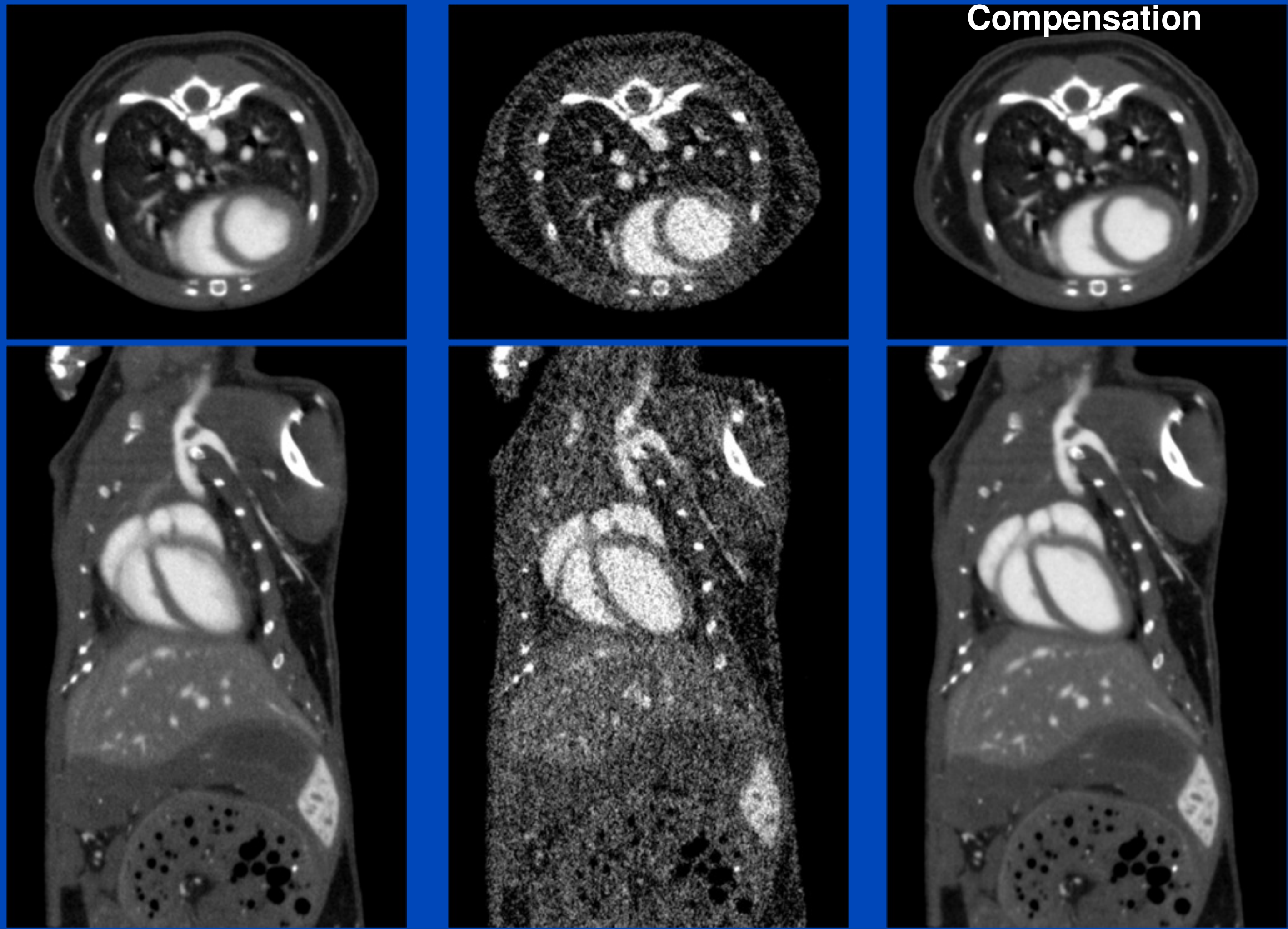


Brehm, Sawall, Maier, and Kachelrieß, “Cardio-respiratory motion-compensated micro-CT image reconstruction using an artifact model-based motion estimation” Med. Phys. 42(4):1948-1958, 2015.

Illustration of Workflow



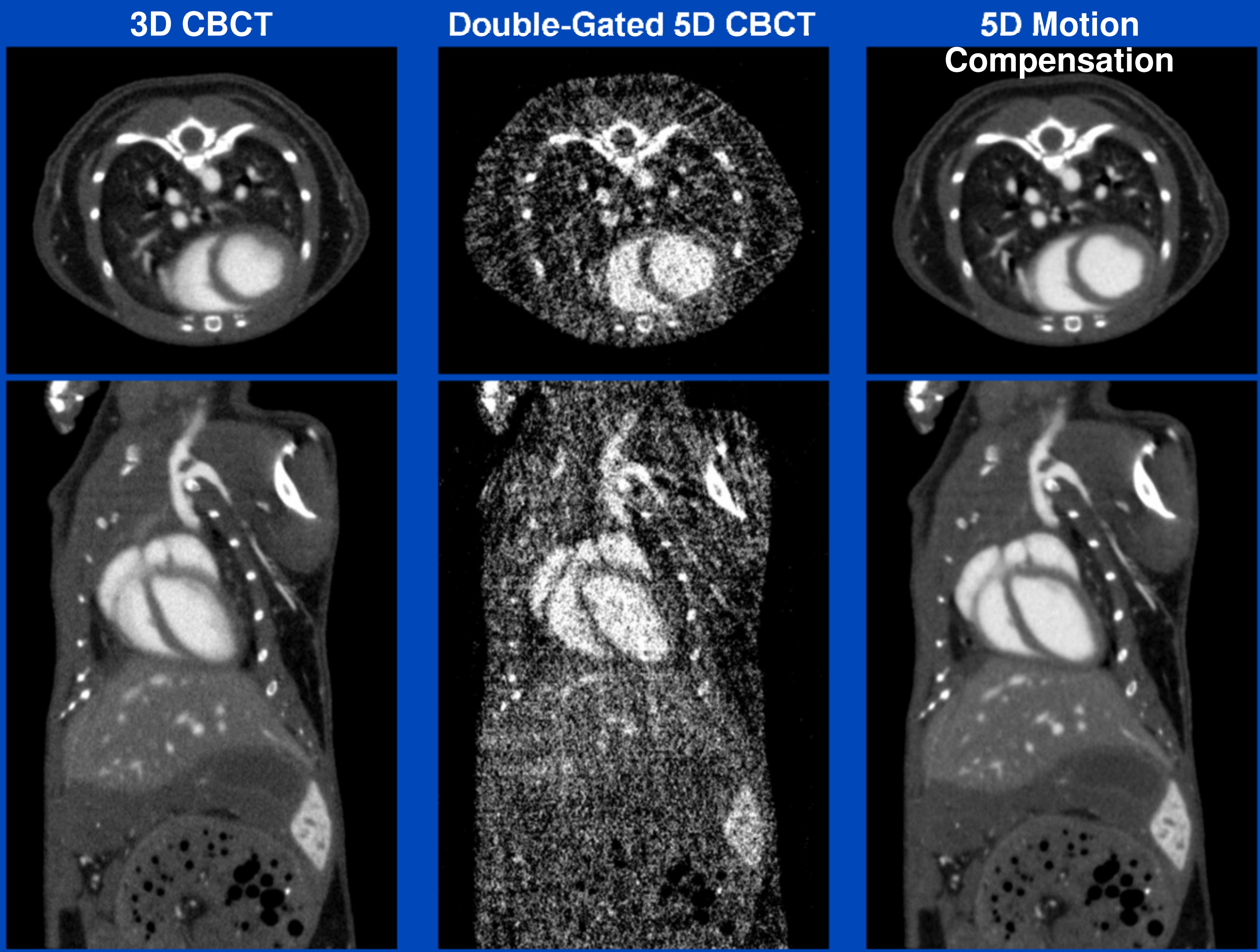
7200 Projections



The images show a fixed respiratory and cardiac phase.

Brehm, Sawall, Maier, and Kachelrieß, "Cardio-respiratory motion-compensated micro-CT image reconstruction using an artifact model-based motion estimation" Med. Phys. 42(4):1948-1958, 2015.

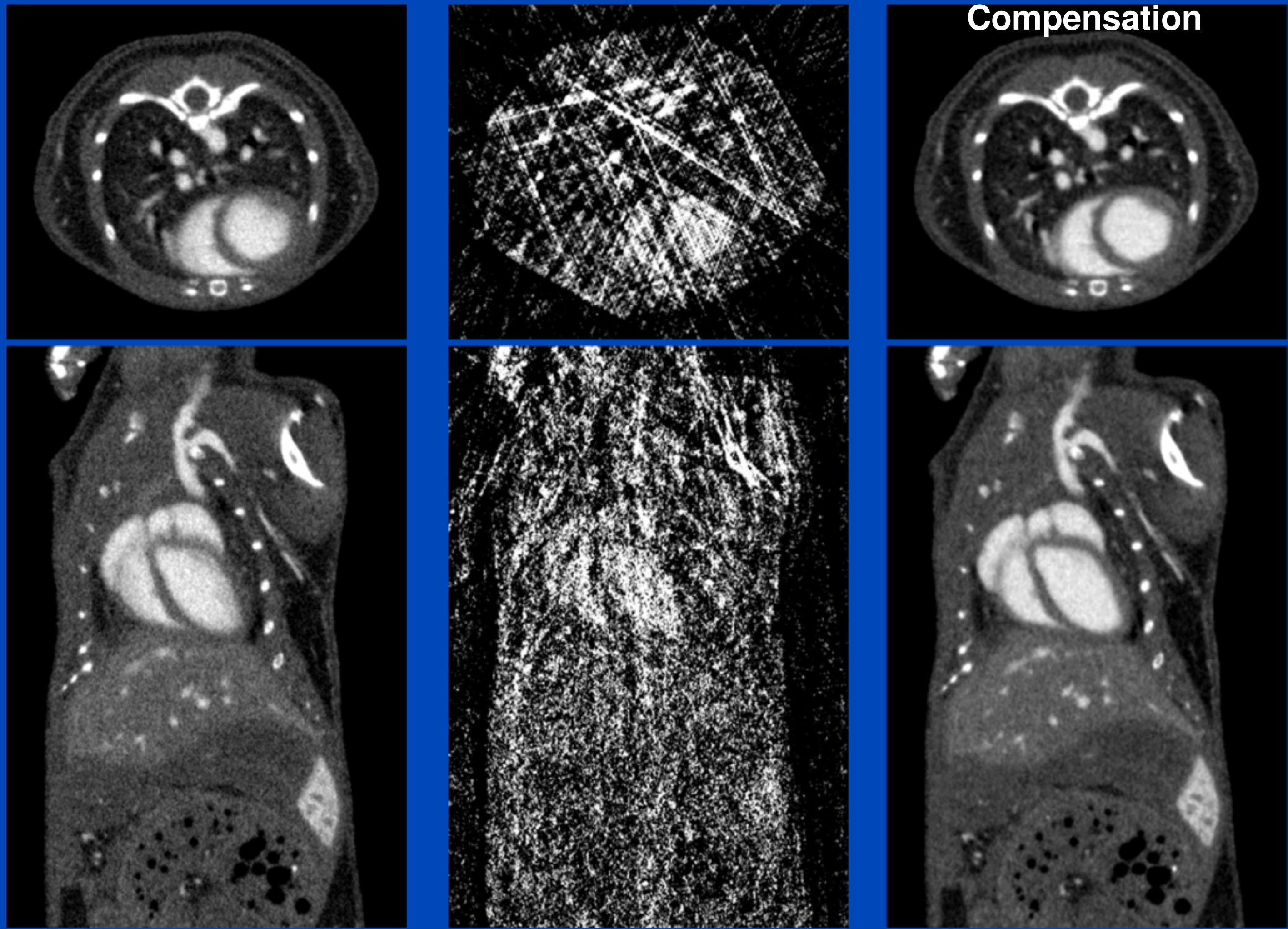
3600 Projections



The images show a fixed respiratory and cardiac phase.

Brehm, Sawall, Maier, and Kachelrieß, "Cardio-respiratory motion-compensated micro-CT image reconstruction using an artifact model-based motion estimation" Med. Phys. 42(4):1948-1958, 2015.

720 Projections



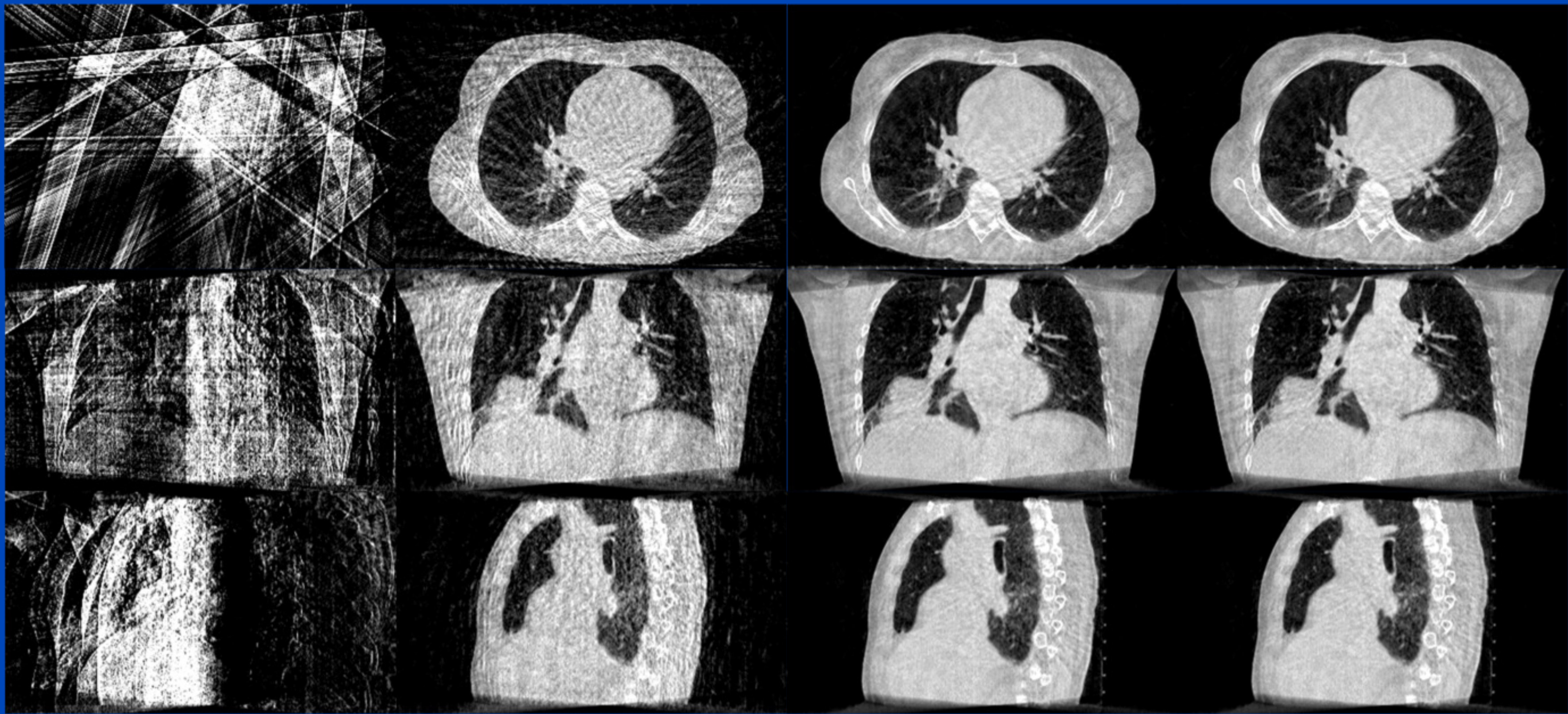
The images show a fixed respiratory and cardiac phase.

Brehm, Sawall, Maier, and Kachelrieß, “Cardio-respiratory motion-compensated micro-CT image reconstruction using an artifact model-based motion estimation” Med. Phys. 42(4):1948-1958, 2015.

MoCo 5D Results

20 respiratory phases of 10% width, 10 cardiac phases of 20% width

PCF 5D	PCF 5D	acMoCo 5D	acMoCo 5D
Respiratory & Cardiac Gated	Respiratory Compensated & Cardiac Gated	Respiratory & Cardiac Compensated r-loop, $c = 0\%$	Respiratory & Cardiac Compensated $r = 0\%$, c-loop



C=-250 HU, W=1400 HU

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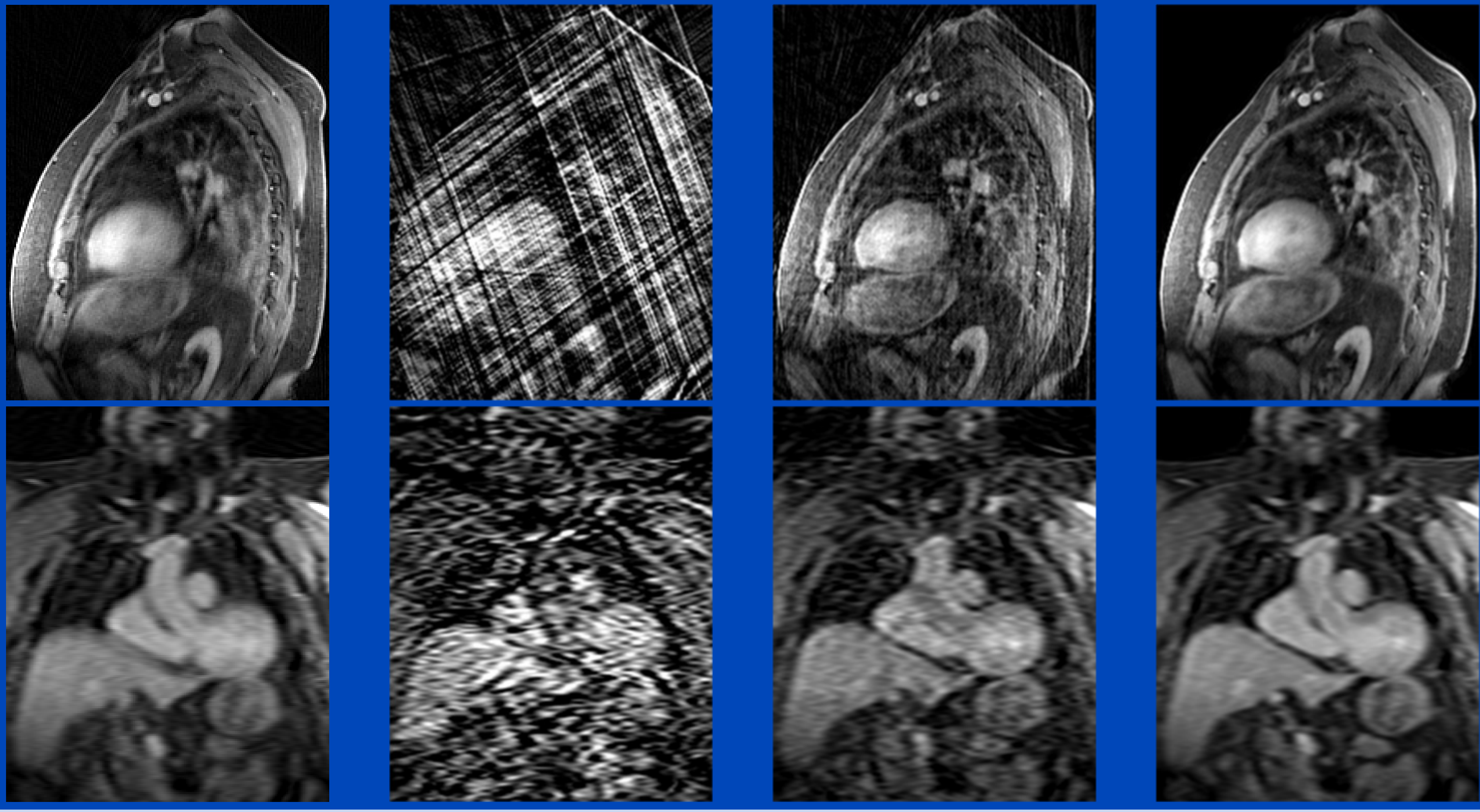
Spin-Off Effects?

5D MR Motion Compensation

Results Patient c12

3D reconstruction	5D reconstruction	5D reconstruction	5D MoCo
motion average	resp & card gated	resp MoCo & card gated	resp & card MoCo
	$r = 1$, c-loop	$r = 1$, c-loop	$r = 1$, c-loop



total acquisition time: 1 min 55 s, radial undersampling = 36

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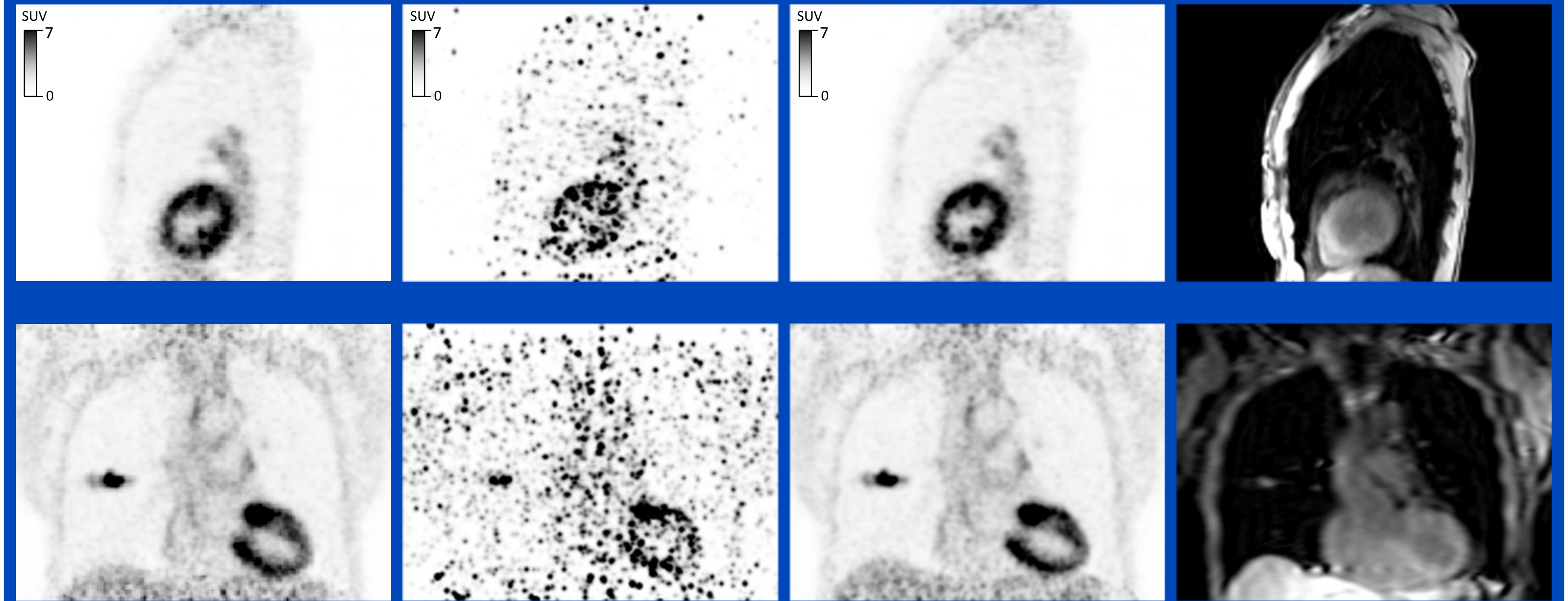
5D PET/MR Motion Compensation

Results Patient s04

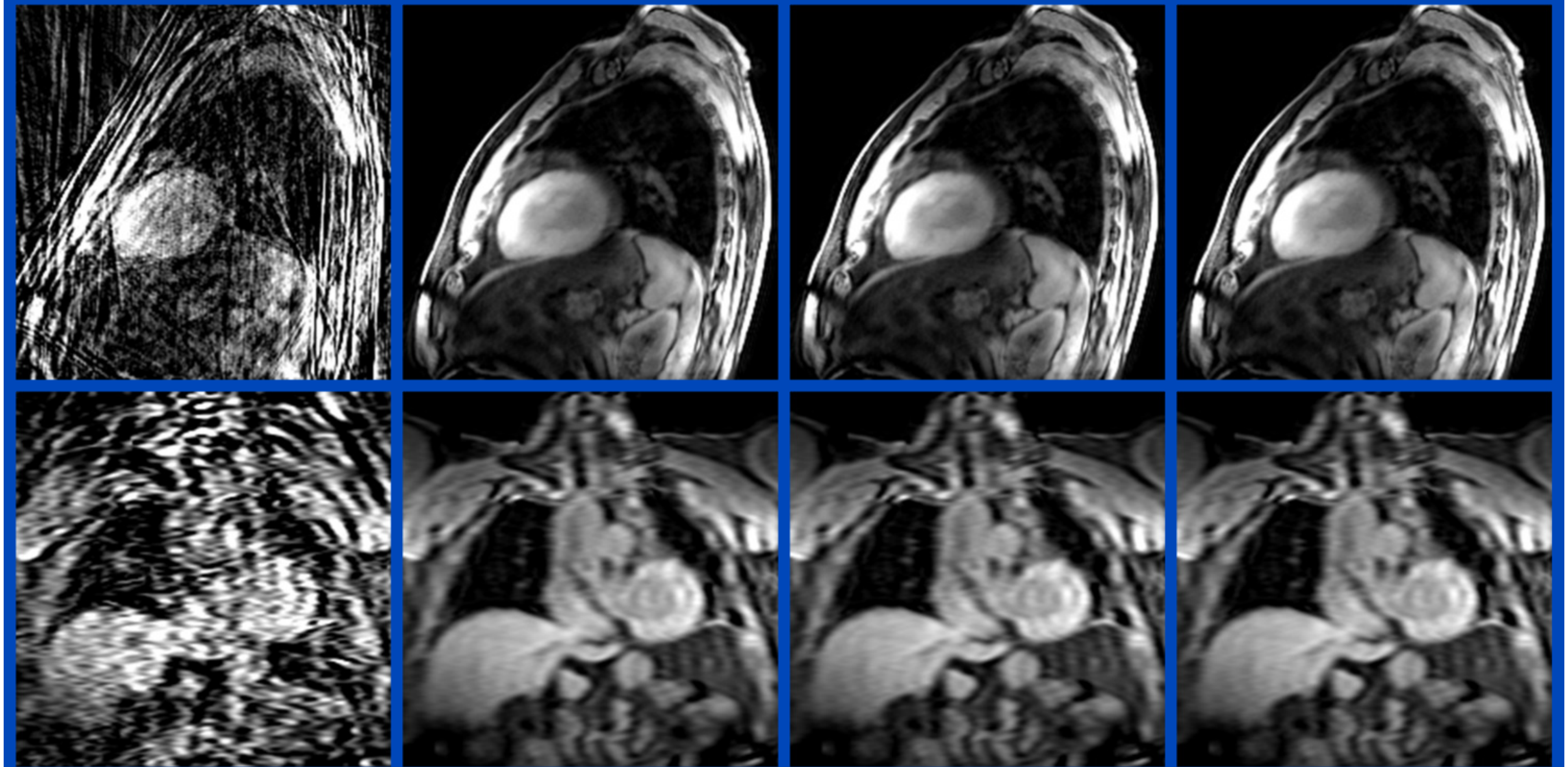
3D PET	**5D double-gated PET**	**5D MoCo PET**	**5D MoCo MR**
motion average	$r = 1$, c-loop	$r = 1$, c-loop	$r = 1$, c-loop



5D MR Motion Compensation

Results Patient s10

5D double-gated MR	5D MoCo MR	5D MoCo MR	5D MoCo MR
r = 1, *c*-loop	*r* = 1, *c*-loop	*r*-loop, *c* = 1	*r*-loop, *c*-loop

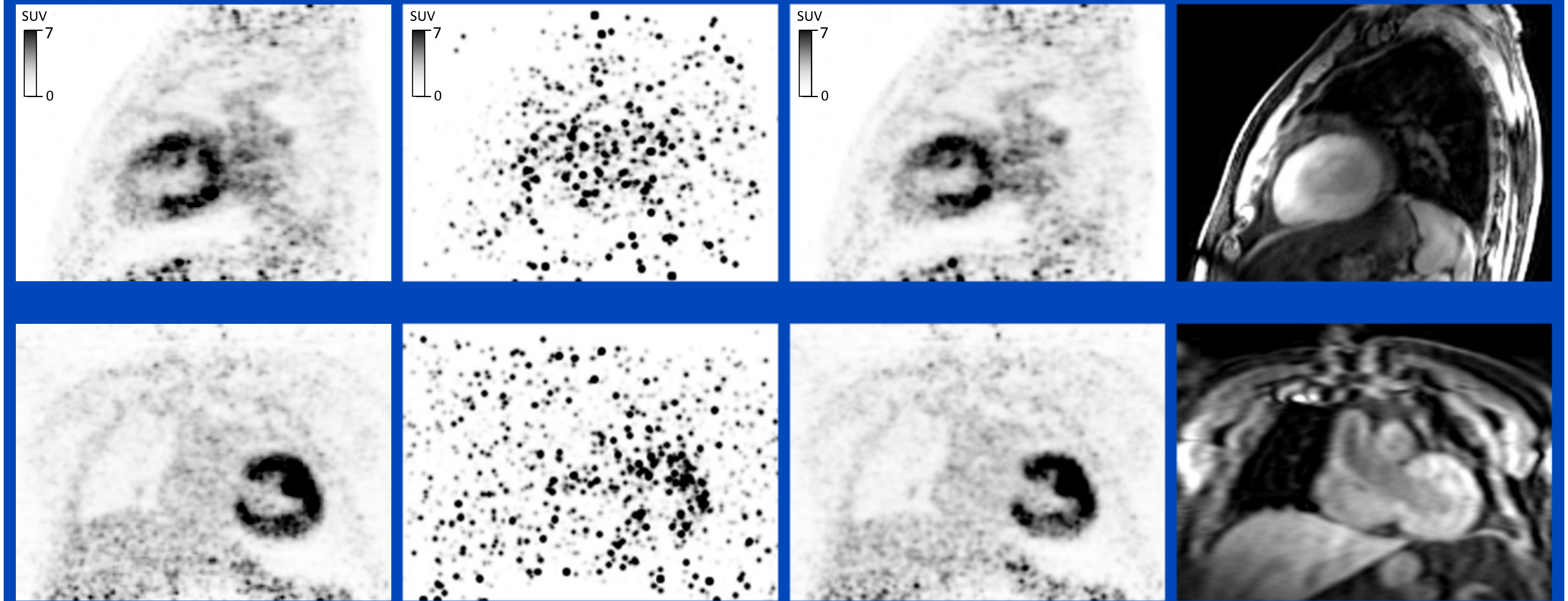


total acquisition time: 5 min

5D PET/MR Motion Compensation

Results Patient s10

3D PET	**5D double-gated PET**	**5D MoCo PET**	**5D MoCo MR**
motion average	*r* = 1, *c*-loop	*r* = 1, *c*-loop	*r* = 1, *c*-loop



total acquisition time: 5 min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Parts of the reconstruction software were provided by RayConStruct® GmbH, Nürnberg, Germany.