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

• Standard FDK-based cardiac reconstruction might have an insufficient temporal resolution introducing strong motion artifacts.

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 \( \cong 180° + \text{fan angle} \)).
- Especially beneficial in cases of patients with high or irregular heart rates or non-optimally chosen gating positions.

\[ C = 300 \text{ HU}; \quad W = 1500 \text{ HU} \]
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 reconstruction of a single cardiac phase
    - Iterative reconstruction algorithms dealing with limited angle artifacts:
      1. Temporal Resolution Improvement Method (TRIM[1])
- 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])
    3. Motion Compensation based on Partial Angle Reconstructions (PAMoCo[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
- **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 ($p' \approx 120^\circ$).
- **Workflow:**

```
Initial short-scan reconstruction

SART-based iterative reconstruction with regularization

Final reconstruction
```
• Minimize the cost function

$$C(f) = C_1(f) + \beta C_2(f).$$

raw data fidelity  regulation term

• Extension of an SART, which optimizes the raw data fidelity

$$C_1(f) = |Xf - p'|^2.$$  Image to be reconstructed

by a regularization term $C_2(f)$.

• Optimizing only $C_1(f)$ would introduce limited angle artifacts, since $\rho' \approx 120^\circ$ covers only a subset of the short-scan range.
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
• **Aim:** Improve the image quality of best phase images.
• **Idea:** Estimate motion vectors by measuring the amount of motion artifacts in the reconstruction (adaptation of auto focus concept from photography).

• **Workflow:**

```
Initialization

Motion model
Motion estimation
Motion compensation

Final reconstruction
```
• 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) $\Omega_s$ incorporating all motion artifacts associated to the chosen coronary artery.
• Motion inside $\Omega_s$ 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 Shift at time $t$ and voxel $x$

• Between the control points, the MVF is approximated by spline interpolation and a dense MVF is generated.
The MVFs are subject to the cost function optimization:

\[
\hat{s} = \arg\min_{s \in \mathbb{R}^{N_p}} \mathcal{L}
\]

Entropy and Positivity are used as image artifact measuring cost functions and are optimized in an alternating manner using a gradient descent method.
Motion-compensated reconstruction is obtained by taking the motion during the backprojection step into account (Schäfer’s method\cite{1}).

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

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

Data courtesy of Dr. Stephan Achenbach
PAMoCo
Generation of 2K+1 PARs

- **Initial segmented stack volume**
  - $t_{\text{res}} \approx \frac{t_{\text{rot}}}{2} \approx 150$ ms

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

- **Partial angle reconstructions** $f_k(r)$
  - $f_{-K}(r)$
  - $f_0(r)$
  - $f_K(r)$

- **Equations**
  - $p_k(\vartheta, \xi) = w_k(\vartheta)p'(\vartheta, \xi)$
  - $w_k(\vartheta) = \Lambda((\vartheta - \vartheta_k)/2\Delta\vartheta)$

- **FWHM** = $\Delta\vartheta$
- **K = 15**
PAMoCo
Motion Model

• Motion inside $\Omega_s$ 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 ($P, L \leq 3$)

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

• A dense MVF is generated by assigning a center line parameter $\lambda$ to each voxel.
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

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

$$f_{\text{MoCo}}(\mathbf{x}, s) = \sum_{k=-K}^{K} f_k(M(s, \mathbf{x}, t_k)).$$

and adding the warped images.
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

Siemens CT system

Motion robot

Vessel phantom

Water tank

Body phantom

Stents

Calcified plaques

50 HU @ 120 kV

d = 1.5 mm  2.5 mm  3 mm
Phantom Measurement

Applied motion
70 bpm

Data acquisition

- Rotation Time $t_{\text{rot}} = 250$ ms $\Rightarrow t_{\text{res}}(\text{FBP}) \approx 125$ ms
- Low pitch spiral scanning: $p \approx 0.2$
- $\Rightarrow$ Reconstruction of multiple cardiac phases possible.
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.
• We characterize the best phase, by the simulated phase featuring least absolute motion:

<table>
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<tr>
<th>70 bpm</th>
<th>Phase featuring strong motion</th>
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• MAM and TRIM aim at increasing the image quality close to the best phase.

• → Perform reconstructions at slightly shifted cardiac phases.
Results
Best Phase

Vessel phantom

\( d = 2.5 \text{ mm} \)

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

HR = 70 bpm, c = 67%, C = 400 HU, W = 1500 HU

Vessel phantom

Stent

d = 2.5 mm
Results
10% off Best Phase

FBP
TRIM
MAM
PAMoCo

axial
sagittal
coronal

Vessel phantom

HR = 70 bpm, c = 62%,
C = 400 HU, W = 1500 HU

d = 2.5 mm
Clinical Case

HR = 74 bpm, c = 74%, C = 400 HU, W = 1500 HU
**Clinical Case**

HR = 74 bpm, c = 74%,
C = 400 HU, W = 1500 HU
Clinical Case

Slice 46

sagittal view

HR = 74 bpm, c = 74%,
C = 400 HU, W = 1500 HU
Clinical Case

Slice 31
sagittal view

FBP

PAMoCo

HR = 74 bpm, c = 30%, C = 400 HU, W = 1500 HU
Clinical Case

Slice 44 - sagittal view

FBP

PAMoCo

HR = 74 bpm, c = 30%, C = 400 HU, W = 1500 HU
Clinical Case

HR = 74 bpm, c = 30%,
C = 400 HU, W = 1500 HU
Clinical Case

HR = 74 bpm, c = 30%,
C = 400 HU, W = 1500 HU
Clinical Case

**FBP**

**PAMoCo**

curved MPRs of the RCA

HR = 74 bpm, c = 30%, C = 400 HU, W = 1500 HU
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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