Do We Need to Model the Ray Profile in Iterative Clinical CT Image Reconstruction?

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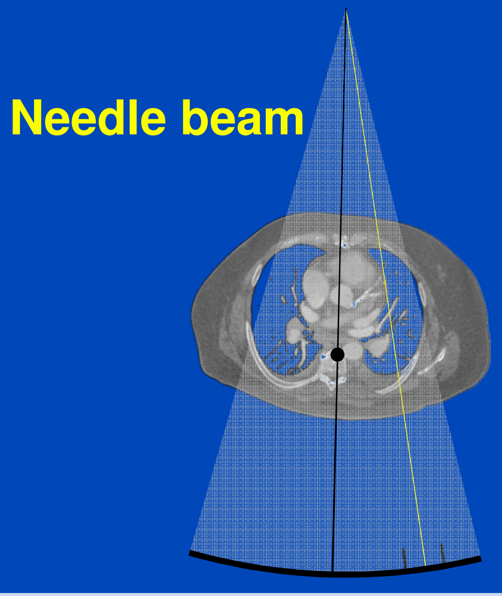
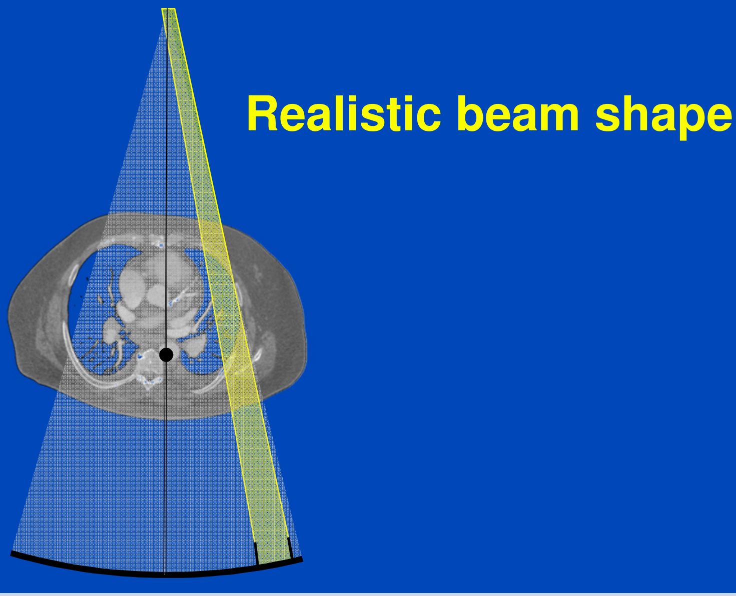
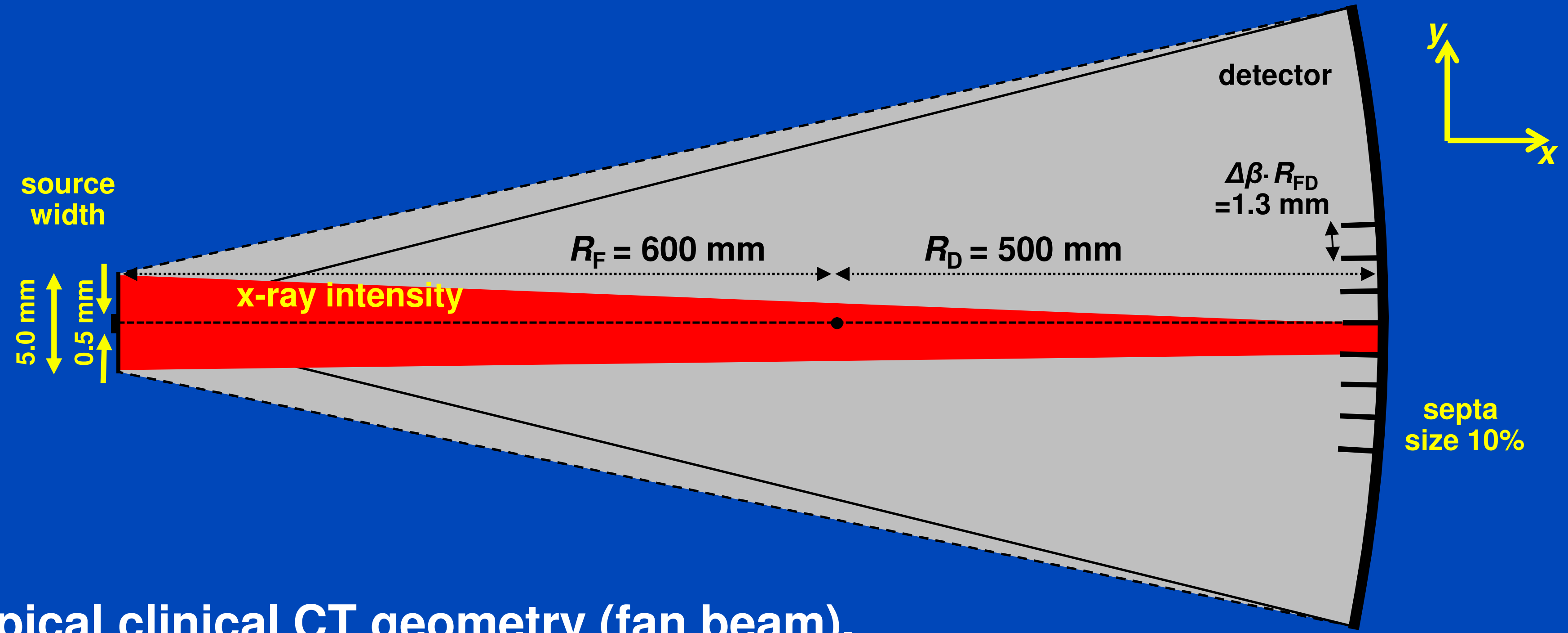
Motivation

- The finite focal spot size, finite detector element width and detector crosstalk lead to a finite beam width. This may impair spatial resolution.
- Iterative reconstruction may be designed to correctly account for the realistic beam shape and to, potentially, achieve resolution recovery.
- Is resolution recovery in clinical CT possible or not?





Ray-Modeling Approach



- **Typical clinical CT geometry (fan beam).**
- **Effects of ray-modeling (RM) on resolution recovery for source widths ranging from 0.5 mm (realistic) to 5.0 mm (unrealistic) are investigated.**
- **The realistic beam shape is realized by simulating many needle beams (0.05 mm spacing at the source and at the detector side).**
- **Detector crosstalk is not simulated (its effect is small in clinical CT due to reflective coatings).**

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Reconstruction Algorithms

- **Ground truth: noise-free ten-fold spatial resolution analytical reconstruction of our analytical phantoms**
- **Reconstructions[1]:**
  - **FBP: Filtered backprojection as a reference**
  - **No RM: Ordered subsets expectation maximization (OSEM) without ray modeling**
  - **With RM: OSEM + ray modeling**
- **Postprocessing:**
  - **Match iterative results to the noise of the FBP reconstruction (by applying a post reconstruction Gaussian filter)**
  - **No postprocessing (images compared at convergence as they are)**
  - **Match the noise of the "No RM" images to the "With RM" images (by unsharp masking)**

[1]Reconstructions use a 0.2 mm pixel grid. Iterations were carried out until near convergence (about 500 iterations).

Phantom and Analysis

- Analytical phantom simulated with 40 HU Poisson noise in the FBP reconstruction.
- Image quality was quantified by computing the normalized cross correlation with ground truth:

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

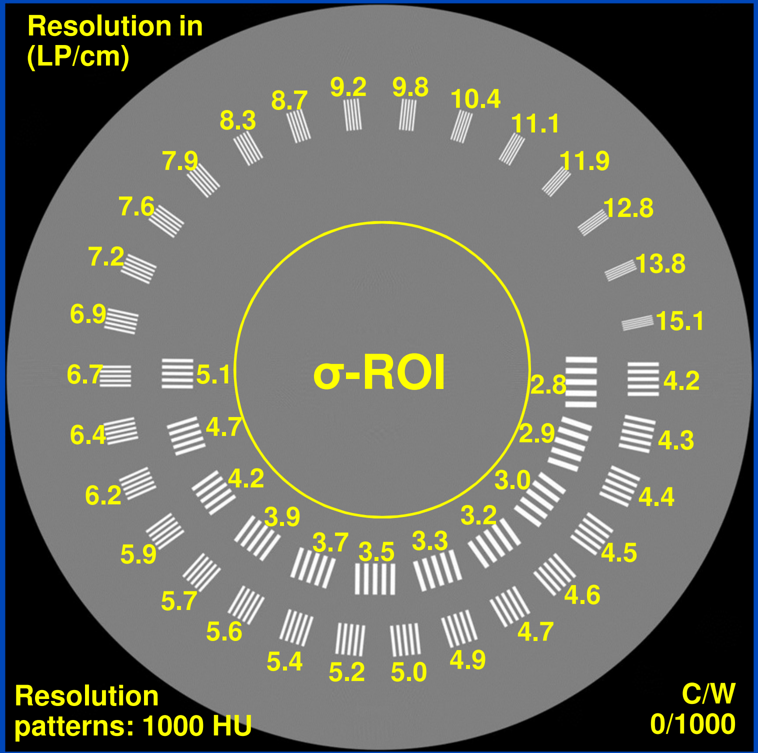
  - f = reconstructed image, g = ground truth
  - σ_f, σ_g = corresponding standard deviations
  - Ω region for NCC analysis

- Resolution recovery potential assessed by using the contrast factor:

$$\mathrm{CF} = \frac{\mathrm{MeanMax}(i) - \mathrm{MeanMin}(i)}{B - A},$$

  - MeanMax(i) = mean of maxima of one resolution pattern
  - MeanMin(i) = mean of minima of one resolution pattern
  - B = 1000 HU, A = 0 HU

Ground truth

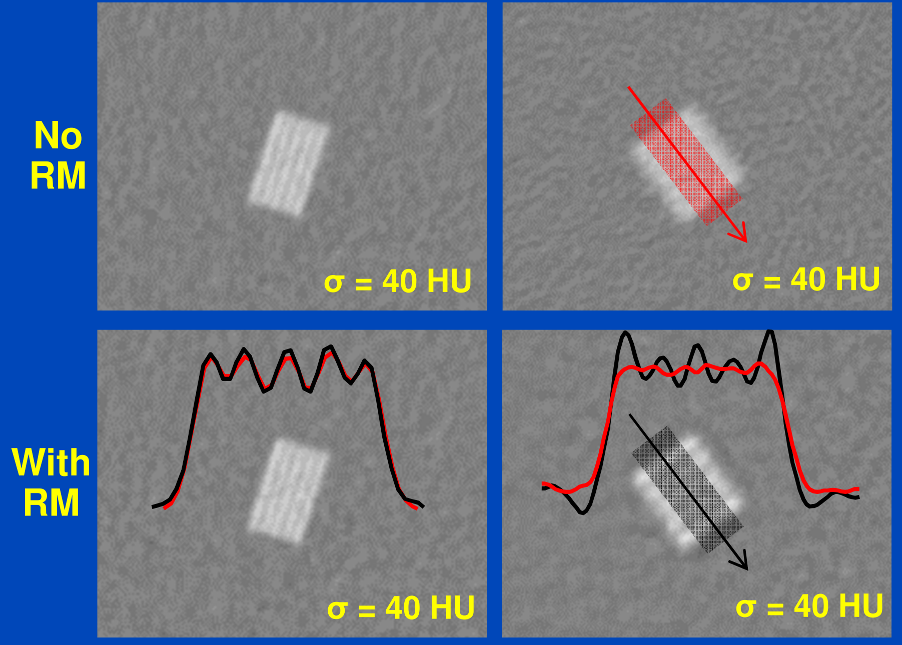


Noise Matched to FBP (1)

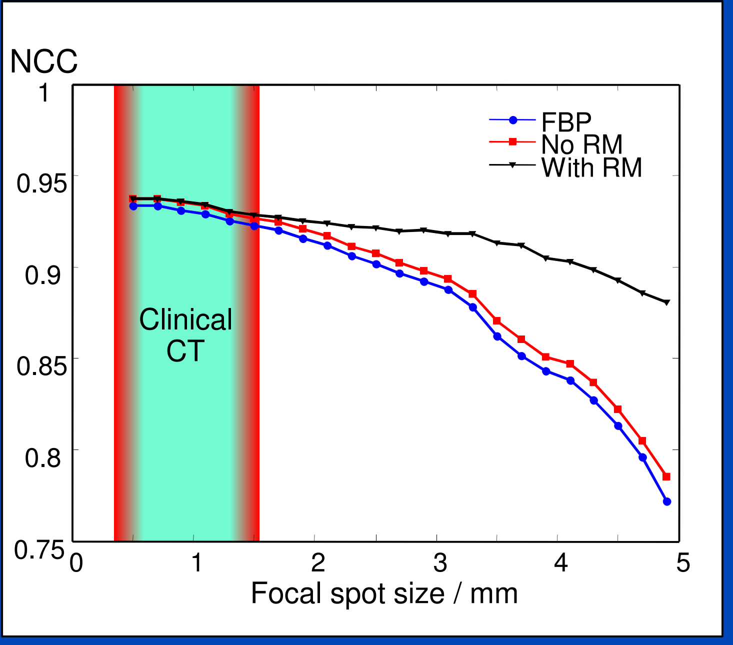
- 0.5 mm focal spot: RM does not improve image quality.
- 5.0 mm focal spot: RM recovers higher frequencies.

0.5 mm focal spot (8.3 LP/cm)

5.0 mm focal spot (4.2 LP/cm)



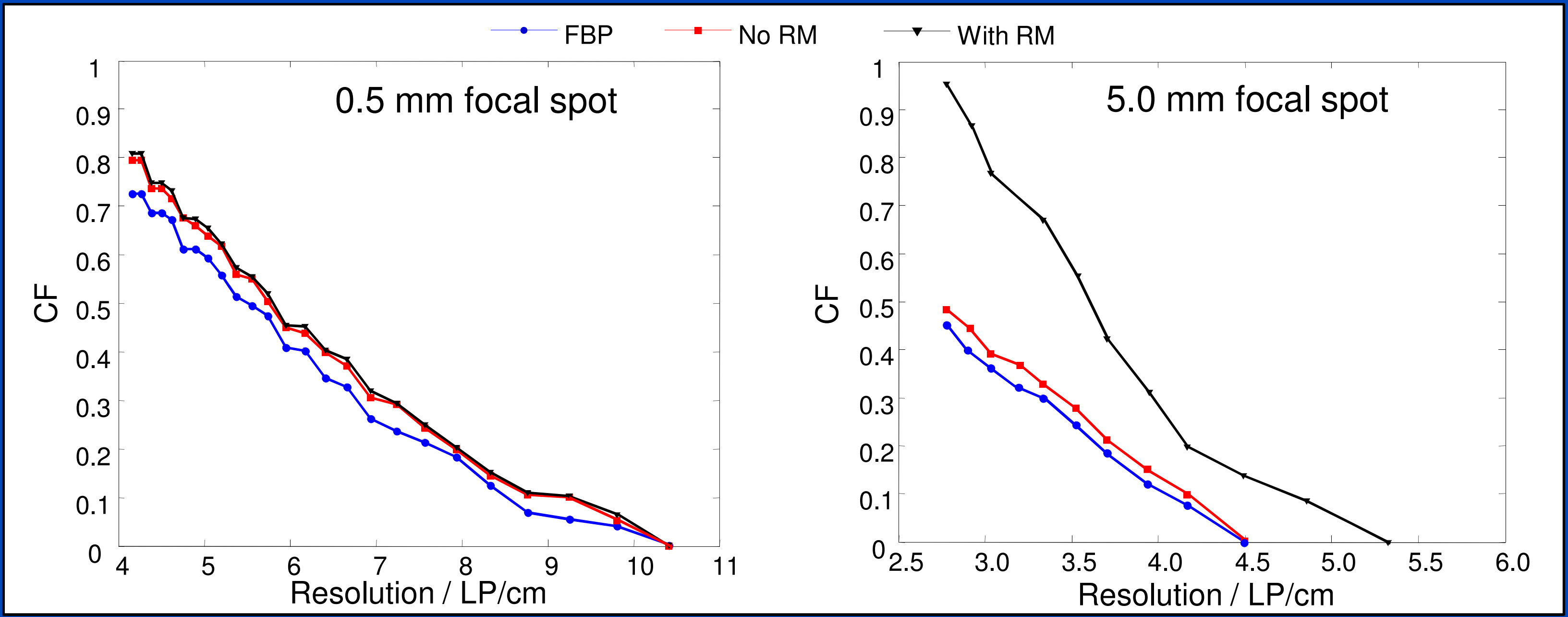
Agreement of resolution patterns with ground truth



Noise Matched to FBP (2)

- 0.5 mm focal spot: RM does not improve image quality.
- 5.0 mm focal spot: RM recovers higher frequencies.

Contrast plots of line resolution patterns



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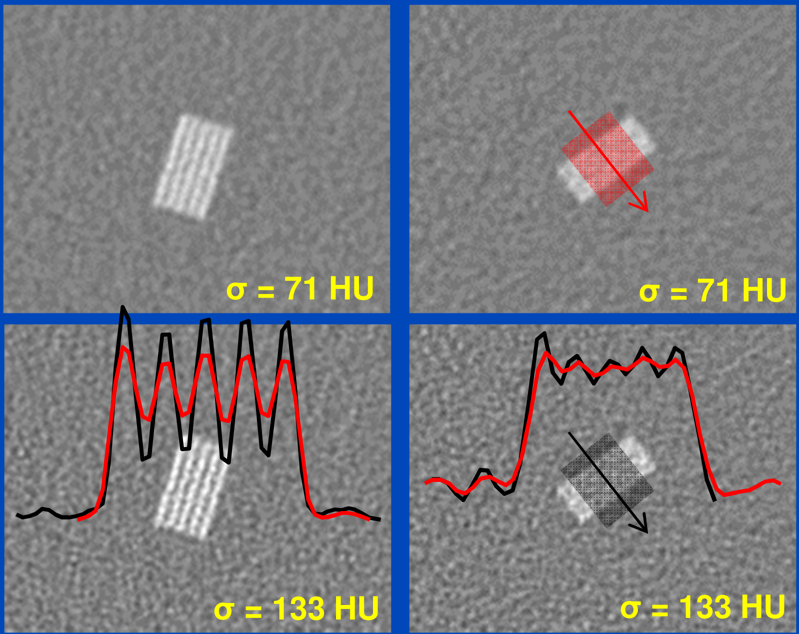
Images as Converged (0.5 mm Source)

- RM results in a higher contrast factor (CF) at the price of higher noise.
- Resolution limit is the same with and without RM.

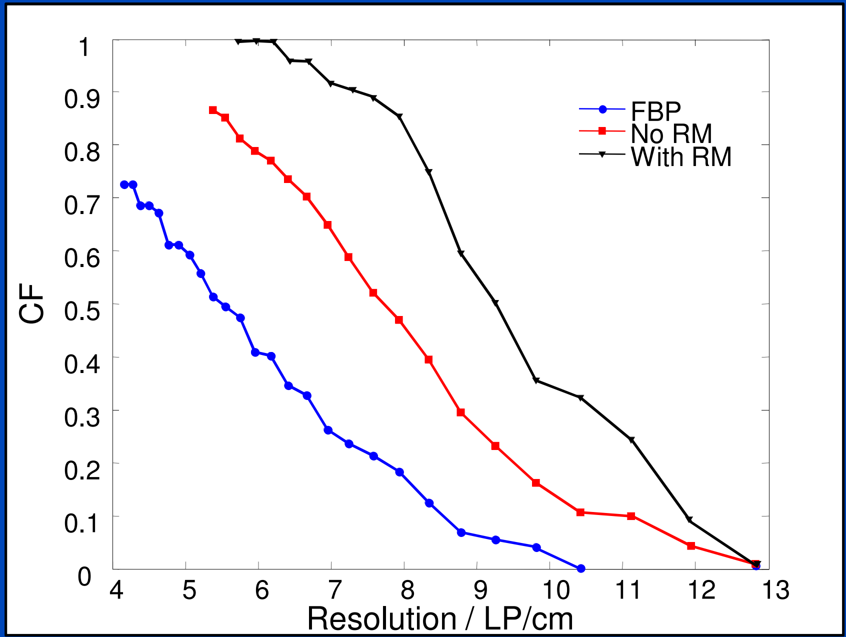
8.3 LP/cm　　11.9 LP/cm

No RM

With RM



Contrast plots of line resolution patterns



C = 0 HU / *W* = 2000 HU

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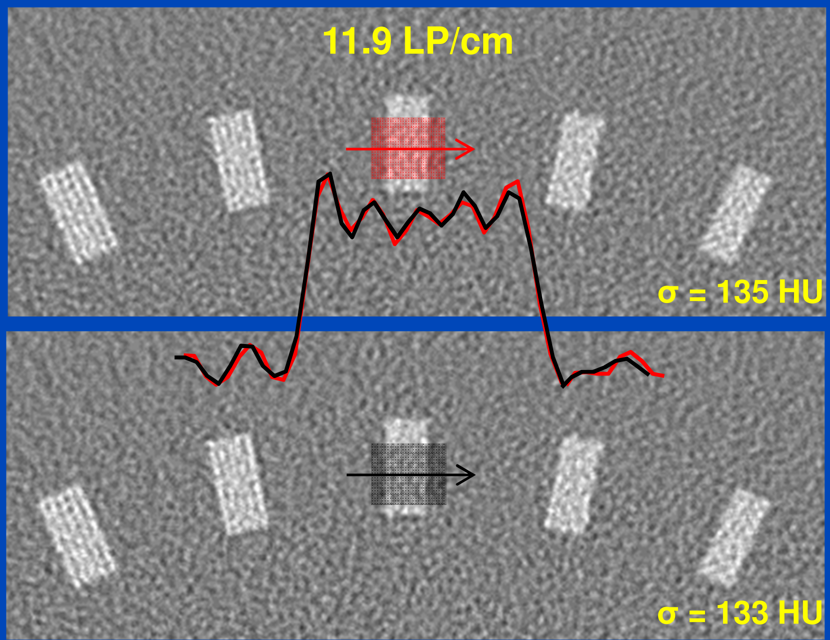
Matching the "No RM" Noise (0.5 mm Source)

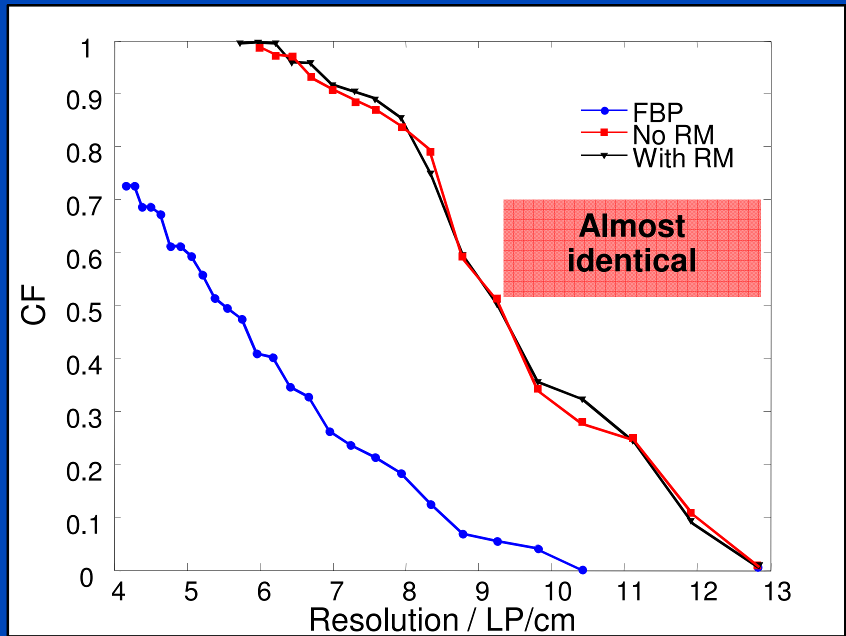
- "No RM" is being sharpened post reconstruction until the noise is matched to noise in the reconstruction with ray-modeling.
- The contrast of the resolution patterns is now almost the same over the whole resolution range.
- The resolution limit is the same.

No RM + Sharpen

With RM



Contrast plots of line resolution patterns



C = 0 HU / W = 2000 HU

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Summary & Conclusion

- When the ray cross-section is much larger than the sampling distance higher frequencies can be recovered with ray-modeling than without ray-modeling.
- In clinical CT with effective beam cross-sections similar to the sampling distance the effects of ray-modeling are negligible.


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