Dose Reduction Strategies for SPECT/CT and PET/CT

Adam Alessio, PhD, DABSNM

aalessio@uw.edu

Department of Radiology
University of Washington

http://faculty.washington.edu/aalessio/

DISCLOSURE:
Dr. Alessio has received grant/research support from GE Healthcare

© Adam Alessio 2015, aalessio@uw.edu
Image Quality Tradeoffs in NM

Image Quality
Diagnostic Utility
Information Density

Radiation Dose (a.u.)
Scan Duration (minutes)

Technology
Dose Savings

Technique/Scanner 1
Technique/Scanner 2
New Technology?
Image Quality Tradeoffs in NM

- Image Quality
- Diagnostic Utility
- Information Density

Operational Dose Savings

Radiation Dose (a.u.)

Scan Duration (minutes)
Goal of Dose Optimization?

A. Make prettiest image possible
B. Minimize radiation dose
C. Maximize physician’s happiness
D. Maximize technologist’s happiness (i.e., shortest acquisition time)
E. Acquire with maximum image quality at minimum of dose
F. Define a task and sufficient image quality to achieve task
Dose Optimization in Nuclear Medicine

• Is all about
  – Injected Activity?
  OR
  – Defining the desired task and the necessary image quality to achieve that task
  – Dose Optimization = Rational Protocol Selection
    ▪ !! More than just a question of injected activity!!
    ▪ Appropriate protocol for the appropriate scanner, clinical resources, study, and patient
    ▪ We need better approaches for rational protocol selection…
Diagnostic Reference Levels

“Diagnostic reference levels (DRLs), which are a form of investigation levels, represent an important tool to optimize image quality and the radiation dose delivered to patients.”

DRL’s help promote (not dictate) good practice for a more specific medical imaging task; and

- Proposed 20 years ago. Used extensively in Europe for Quality Assurance

- ICRP 73 (1996)
Diagnostic Reference Levels

- DRLs are set at approximately the 75th percentile of similar studies for similar patients
- Achievable doses, AD, represent the median (50$^{th}$ percentile) of doses

- ICRP 73 (1996)
• 75% of doses below Diagnostic Reference Level
• 50% of doses below Achievable Dose (encourage dose optimization for sites below 75% level)
For nuclear medicine, the 75th percentile maximum RLs should be used as guidelines to limit unnecessary radiation dose as long as diagnostic-quality nuclear medicine studies are obtained, but not as absolute limits.

<table>
<thead>
<tr>
<th>Examination</th>
<th>DRL (mCi)</th>
<th>AD (mCi)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tc99m-Tetrofosmin (Stress)</td>
<td>39.0</td>
<td>25.0</td>
</tr>
<tr>
<td>Tc99m-Tetrofosmin (Rest)</td>
<td>29.0</td>
<td>18.0</td>
</tr>
<tr>
<td>99mTc-MAG3</td>
<td>10.0</td>
<td>7.5</td>
</tr>
<tr>
<td>Tc99m-MDP</td>
<td>32.0</td>
<td>23.0</td>
</tr>
<tr>
<td>F18-FDG</td>
<td>19.0</td>
<td>15.0</td>
</tr>
</tbody>
</table>

Many pages of DRL’s for CT based primarily on ACR CT Accreditation Materials

| TABLE 6.5—Adult body CTDI$_{w}$ (milligray) from the ACR CT Accreditation Program. |
|---------------------------------|-------|-------|-------|-------|
| Mean                            | 18.7  | 19.2  | 17.0  | 18.4  |
| Standard deviation              | 8.0   | 8.7   | 7.6   | 8.3   |
| 75th percentile                 | 22.6  | 23.4  | 21.1  | **22.2** |
| 90th percentile                 | 29.4  | 30.6  | 25.8  | 29.5   |

Potential Diagnostic Ref Level
SAM Question:
Diagnostic Reference Levels (DRLs) can be used in clinical practice to:

A. Provide legal justification in event of malpractice law suit
B. Set standards to identify normal, average doses
C. Set standards to identify unusually low doses
D. Compare local practice with peer institutions and national levels
E. Provide required protocol settings for local practice
Technology Dose Savings

Current technologies providing genuine improvements:

• Improved collimators (SPECT)
• Improved solid-angle coverage (SPECT, PET)
• Improved detectors and electronics (SPECT, PET, and CT)
• Improved data processing (SPECT, PET, and CT)
  – Iterative image reconstruction
• Improved review software/workstations
Collimator Efficiency

• Collimators typically absorb well over 99.95% of all incident photons.

• Trade-off between spatial resolution and detection efficiency (sensitivity).

• Collimator choices: LEGP, LEHR, MEGP, High Energy
  - balance the trade-off
  - used for different isotopes
Collimator Sensitivity
Point Source Geometric Efficiency in Air
Collimator Design

- Thicker septa lead to more attenuation
- Low sensitivity

- Unique septa design enables industry-leading NEMA sensitivity* (up to 26% higher)

*Vendor Statement: Slide provided from Siemens Healthcare
A Benefit of Application Specific Geometries: Solid-Angle Coverage

Parallel Collimator:
Same detection efficiency, Different resolution

Focused geometries can provide significantly better solid angle coverage
→ Many more counts detected at a time
Detectors: NaI vs Cadmium Zinc Tellurium (CZT)

- Inexpensive
- Energy Resolution ~9%
- Spatial Resolution ~4mm

- Relatively expensive
- Energy Resolution ~5%
- Spatial Resolution ~2mm
- Compact

Figure from GE Healthcare, Alcyone Technology White Paper
The Reconstruction Problem: An Inverse Problem

\[ y = Px + n \]

\[ x = P^{-1}(y - n) \]

DIFFICULT: Requires Iterative Solution

1. Each Vendor can have unique representation for \( y, P, x, n \)
2. And, how they solve \( P^{-1} \)

Main Point: Not all “OSEM” algorithms the same

Not all Vendors Recon algorithms are the same…
Iterative Image Reconstruction in SPECT

Faster, Better Images:


Increasing Applications For Quantitative SPECT:

2. Beauregard et al, Quantitative 177Lu SPECT (QSPECT) imaging using a commercially available SPECT/CT system, Cancer Imaging 2011.
Quantitative SPECT Reconstruction

Key Components:
1. Attenuation correction
2. Intra-Patient Scatter correction
3. Accurate System Model (includes collimator-resolution modeling)
4. Intra-Collimator Scatter correction
5. View-dependent decay correction

- ALL incorporated into reconstruction algorithm
Collimator Resolution
Dependent on source-collimator distance

**FIGURE 21-12.** Line spread function (LSF) of a parallel-hole collimator as a function of source-to-collimator distance. The full-width-at-half-maximum (FWHM) of the LSF increases linearly with distance from the source to the collimator; however, the total area under the LSF (photon fluence through the collimator) decreases very little with source to collimator distance. (In both figures, the line source is seen “end-on.”)
Collimator Resolution

Dependent on source-collimator distance

- All collimators suffer from depth dependent resolution response
- Iterative reconstruction methods can model, and therefore somewhat compensate for, the resolution response of the collimator

From: The Essential Physics of Medical Imaging (Bushberg, et al)
Example of Iterative Recon Trial
Stress/rest Tc-99m tetrofosmin single-isotope study

Full Duration, Filtered Backprojection

Half Duration, Wide Beam Reconstruction from UltraSPECT

Conclusion from this study: cardiac SPECT perfusion studies may be performed with the WBR algorithm using half of the scan time without compromising qualitative or quantitative imaging results.

SAM Question

Technology dose savings can be achieved in SPECT imaging through all the following except:

A. Improved collimator designs
B. Higher resolution detectors made from materials such as CzT
C. Improved data processing and reconstruction algorithms
D. Faster rotation of detector heads
E. Greater solid angle coverage
Trends in PET Technology

• Larger Bore Sizes (70cm towards 78cm and more…)

• More Reproducible Quantitation
  – Better Calibration (ex: Siemen’s Quanti-QC)
  – Respiratory Compensation
  – Better image reconstruction (ex: GE’s Q.CLEAR)

• Better Signal to Noise through:
  – Better Time-of-Flight (ex. Philip’s Digital PET)
  – Larger axial sampling
  – Better image reconstruction
Digital Photon Counting PET

Converts scintillation light directly to a digital signal, with zero analog noise.

Allows for Faster Timing Resolution

- ≈ 2x volumetric resolution
- ≈ 2x sensitivity gain
- ≈ 2x quantitative accuracy

Vendor Statements from Philips Healthcare
Variations in resolution loss vs. size and smoothing

**FBP**

**OSEM**

Increasing smoothing
How to reduce partial volume effect in PET?

PSF-Based Iterative Reconstruction

Each radial location has blur in radial direction

$P_{PSF}(s_v, s)$

$S_v = 2\text{mm}$

$S_v = 348\text{mm}$

Measured profile in black, parameterized profile in red
Measured Spatially Variant System Modeling (PSF) in Iterative Reconstruction

Images with “clinical” reconstruction parameters: 2.7mm/pixel, 7mm post-filter, 28 subsets

Contrast Recovery vs. Size

Contrast Recovery vs. “True” Noise across 50 scans

Proposed Method

Prior “best” method

Observations: Addition of PSF...

• Leads to roughly 7% bias improvement at matched true noise levels across all sphere sizes

FDG PET Exam, 109kg patient

Time-Of-Flight PET

- Measures time difference of detection of photons
  - If time difference = 0, annihilation at center of field of view

- Timing resolution 500 ps = 7.5 cm

Conventional backprojection

TOF backprojection
Time-of-Flight PET

Contrast recovery coefficient versus noise for 27 cm diameter cylinder

Contrast recovery coefficient versus noise for 35 cm diameter cylinder

TOF gain for matched noise levels, averaged over 6–9 lesions (1- to 2-cm diameter) for each patient, is plotted as function of patient mass. Error bars reflect the range of TOF gains seen for this patient.

Question:

Time-of-flight PET is especially beneficial for:

A. High resolution brain imaging
B. Smaller pediatric patients
C. Depth of interaction detectors
D. Obese patients

Answer: D. Larger patients will have more signal to noise gains than smaller objects.
CT: Instrumentation/Processing

• CT Detectors
• Improved data/image processing
  – Discussion of CT image enhancement
  – Iterative image reconstruction
Modern Systems use Solid State Scintillation Detectors

- Scintillation Materials: CdW0₄, Gd₂O₂S, HiLight™, GEMSStone™, CsI
- Coupled to photodiodes

- Flat-panel detectors usually use Cesium-Iodide (CsI) coupled to amorphous silicon photodiodes
  - Originally developed for angiography
  - Used in
    - C-Arm Conebeam CT Systems
    - Philips BrightviewXCT SPECT/CT
  - Cons: Low contrast resolution and Slow acquisition
  - Pros: High spatial resolution and Large area
Each vendor is offering “iterative” methods to reduce image noise, effectively allowing for reduced dose acquisitions at matched image quality.

“Iterative” data enhancement methods can be applied at any step in imaging chain.

Image-Based Iterative Methods:
- Philips: iDose
- Siemens: IRIS (iterative reconstruction in image space)
- Toshiba: AIDR (adaptive iterative dose reduction)
- Third Party Solutions: Clarity™ CT from Sapheneia (Sweden); ContextVision Inc. (Sweden)

“More Fully” iterative reconstruction methods:
- Toshiba: AIDR^3D (adaptive iterative dose reduction)
- GE Healthcare: ASIR (Adaptive Statistical Iterative Recon), ASIR-V, Veo
- Philips: iDose^4
- Siemens: SAFIRE (Sinogram Affirmed Iterative Recon)
Image-Based Enhancement

Marketing Brochure for Clarity CT Solutions,
www.claritysolutions.org
Soft Tissue Conspicuity

Increased noise → FBP → Increased conspicuity

FBP

GE:ASIR

GE:VEO

120 kVp, variable mAs (NI=36), 1.375 pitch. 0.625/0.8 mm slice: Width = 400, Level = 40 HU
65 YO female, 83.7 kg, 160 cm, BMI = 32.7
Coronary CT angiography

Dose Reduction Techniques

CT: Operational

• Factors that affect radiation dose with CT
• Appropriate protocols
  – Diagnostic CT vs Localization CT vs Attenuation Correction CT
Factors Affecting CT Dose

*Direct Influence on Dose*
- X-ray beam energy (kVp)
- X-ray tube current (mA)
- Rotation or exposure time
- Slice thickness
- Object thickness
- Pitch or spacing
- Dose-reduction techniques
- X-ray source to isocenter distance

*Indirect Influence on Dose*
- Reconstructed slice thickness
  image statistics require higher kVp and/or mAs in thinner slices to achieve equivalent level of noise as in thicker slices.
- Reconstructed image resolution
  algorithms enhancing spatial resolution also increase image noise- higher kVp and/or mAs may be used to compensate.
Factors Affecting CT Dose

CTD1_w measured in 16cm head & 32 cm body phantoms

As you decrease dose, you increase noise (usually decrease image quality) – No Free Lunch

Automatic Exposure Control

• Modulate Tube Current based on patient specific information in the
  – Longitudinal (z-axis) for example: AutoMA (GE), Z-DOM (Philips), CareDose (Siemens)
  – Angular Direction

• On average, can achieve ~20% (3-45%) dose reductions at matched quality*

## Comparison of Typical Doses Hybrid CT Acquisitions

<table>
<thead>
<tr>
<th>Study</th>
<th>Technique</th>
<th>Effective Dose (mSv)</th>
</tr>
</thead>
<tbody>
<tr>
<td>NM Bone Scan</td>
<td>20 mCi Tc99m MDP (740 MBq)</td>
<td>4.2</td>
</tr>
<tr>
<td>High-low Myocardial Perfusion</td>
<td>Stress:Rest Tc99m trofosmin (40:10 mCi)</td>
<td>13</td>
</tr>
<tr>
<td>CT for diagnostic purposes</td>
<td>[110-200] mAs(^1) CTDI(_{vol}) = [8-14] mGy</td>
<td>11-20</td>
</tr>
<tr>
<td>CT for anatomic localization</td>
<td>[30-60] mAs(^3) CTDI(_{vol}) = [2-4] mGy</td>
<td>3-6</td>
</tr>
<tr>
<td>CT for attenuation correction only</td>
<td>[5-10] mAs(^4) CTDI(_{vol}) = [0.3-1.0] mGy</td>
<td>0.5-1.0</td>
</tr>
</tbody>
</table>

For ease of comparison, all CT studies performed with 120kVp, pitch 1.375, 40mm collimation, 900 mm scan range, average tube current-time product is presented.


Summary

Dose Reduction is possible through

• Operational Dose Savings:
  – Rational Protocol Selection
  – Potentially use reference levels to align with peer institutions

• Technology Dose Savings:
  – Application-specific geometries (SPECT, PET)
  – Improved collimators (SPECT)
  – Improved detectors (SPECT, PET, and CT)
  – Improved image reconstruction (SPECT, PET, and CT)

• Review software is critical part of realizing potential of hybrid devices

Thank You
Adam Alessio
aalessio@uw.edu