SPECT / CT

Basics, Technology Updates, Quality Assurance, and Applications

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Educational Objectives

1. Understand the underlying principles of SPECT/CT image acquisition, processing and reconstruction
2. Understand current and future clinical applications of SPECT/CT imaging
3. Familiarization with commercially-available SPECT/CT systems

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Outline

- Review of SPECT principles
- Iterative SPECT reconstruction
- Hybrid SPECT/CT imaging
- SPECT/CT quality assurance
- Commercial SPECT/CT systems
- SPECT/CT clinical applications

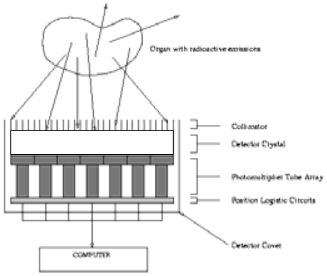
SPECT Basics

- Single Photon Emission Computed Tomography
  - Radio-pharmaceutical administration – injected, ingested, or inhaled
  - Bio-distribution of pharmaceutical – uptake time
  - Decay of radionuclide from within the patient – the source of information
  - Gamma camera detects gamma rays and images (tomography) the radio-pharmaceutical distribution within the patient – SPECT
  - Used for visualization of functional information based on the specific radio-pharmaceutical uptake mechanism

SPECT Hardware

Anatomy of a Gamma camera

1. Collimator
2. Scintillation Detector
3. Photomultiplier Tubes
4. Position Circuitry
5. Data Analysis Computer



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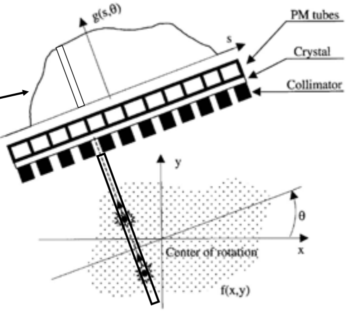
SPECT Back-Projection Model



g(s,θ) = Σf(x,y) along an in-plane line integral

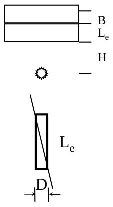
© Bruyant, P. P., J Nucl Med 2002; 43:1343-1358

Crystal Thickness

- Thinner crystals → ↑ spatial resolution
  - interactions occur at a better defined depth
  - multiple interactions less likely
  - less light spread
  - ↓ interaction likelihood for higher energy γ's
- Thicker crystals → ↑ sensitivity
  - ↑ interaction likelihood (esp. for higher E γ's)
  - ↑ likelihood of multiple interactions
  - greater light spread → ↓ spatial resolution

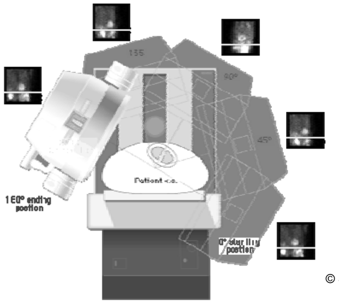
Spatial Resolution

- Intrinsic Spatial and Energy Resolution
  - # of scintillation photons, N ∝ Gamma-ray energy, E
  - Spatial Resolution = 100 × σ/N ∝ 1/√N ∝ 1/√E
  - Energy Resolution = 100 × FWHM/E ∝ 1/√E
- Collimator Resolution $R_g = \frac{D(L_e + H + B)}{L_e}$
- System Resolution $R_s^2 = R_i^2 + R_g^2$



SPECT Acquisitions

- SPECT acquires 2-D projections of a 3-D volume



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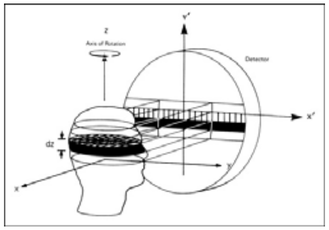
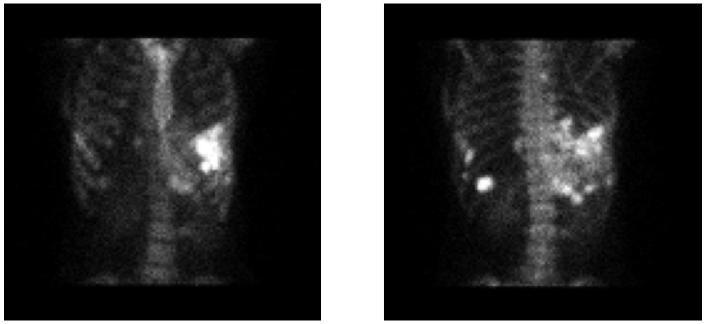


© SPECT in the year 2000: Basic principles, JNMT 24:233, 2000

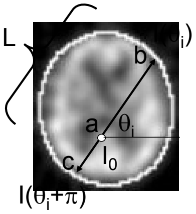
Radon transform angular symmetry violated in SPECT

$P(\theta) \neq P(\theta+\pi)$



Radon transform angular symmetry violated in SPECT

- Why ?
- Due to Differential Attenuation



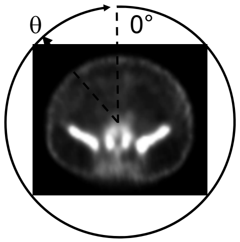
$$I(\theta_i) = I_0\, e^{-\int_a^b \mu(L)dL}$$

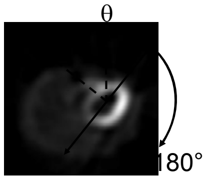
$$I(\theta_i+\pi) = I_0\, e^{-\int_a^c \mu(L)dL}$$

- Other mediating factors:
  - distance-dependent resolution
  - depth-dependent scatter

SPECT Acquisitions

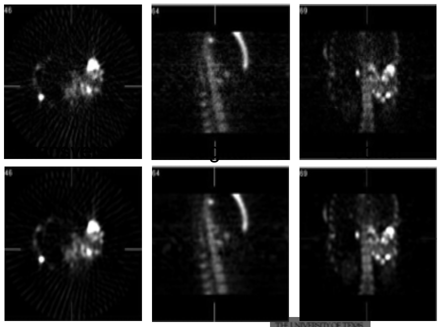
- SPECT projections acquired over 360°
- Exception: Cardiac SPECT acquired over 180°





SPECT images have isotropic voxel size

2-D filter of projections ≡ 3-D post-reconstruction filter



No volume smoothing

Butterworth: 0.6 Nyquist, 10th order

SPECT Filtered Back-Projection

- FBP based on ideal Radon inversion formula
  - assumes a linear, shift-invariant system and angular symmetry of projections
- SPECT imaging systems are neither angularly symmetric nor shift-invariant
  - SPECT projection data affected by attenuation, scatter, and spatial resolution that are all depth-or distance-dependent
- Thus, FBP reconstruction cannot adequately model the physics of SPECT

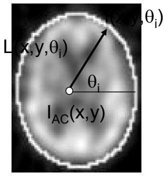
Conventional SPECT Corrections

Attenuation: Chang post-processing algorithm

$I(x,y)$ = SPECT image w/o AC

$I(x,y,\theta_i) = I_{AC}(x,y).e^{-\mu L(x,y,\theta_i)}$

$I_{AC}(x,y) = I(x,y) / \{(1/M).\Sigma_i e^{-\mu L(x,y,\theta_i)}\}$; $i = 1, M$



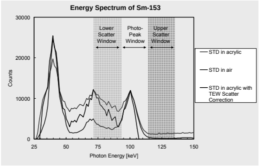
Scatter: Energy window subtraction

$P(x,y)$ = projections w/ scatter

$P_{LE}(x,y)$ = projection at lower energy

$P_{HE}(x,y)$ = projection at higher energy

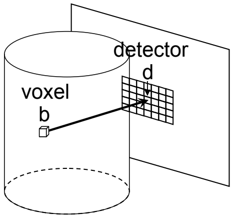
$P_{SC}(x,y) = P(x,y) - k_L.P_{LE}(x,y) - k_H.P_{HE}(x,y)$



SPECT Iterative Reconstruction

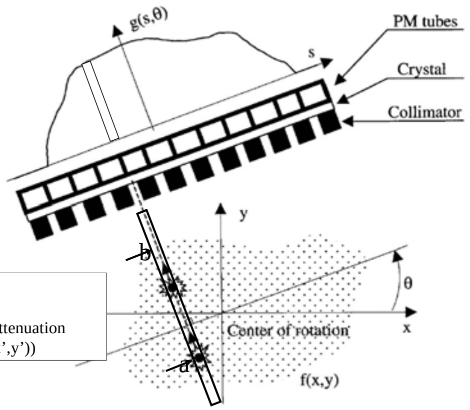
Maximum Likelihood-Expectation Maximization (ML-EM)

- Accounts for the statistical nature of SPECT imaging
- Incorporates the system response p(b,d) – the probability that a photon emitted from an object voxel *b* is detected by projection pixel *d*
- p(b,d) captures…
  1. Depth-dependent resolution
  2. Position-dependent scatter
  3. Depth-dependent attenuation



- Use a measured attenuation map along with models of scatter and camera resolution to perform a far more accurate reconstruction

SPECT Iterative Recon: Attenuation Modeling

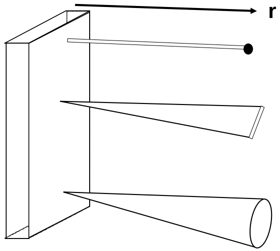


along a line integral …
$g(s,\theta) = \Sigma f(x,y) * p_{attn}(x,y,s,\theta)$
$p_{attn}(x,y,s,\theta)$ = probability due to attenuation
$p_{attn}(x,y,s,\theta) = \exp(-\Sigma_{ab}\mu(x',y')\Delta(x',y'))$

SPECT Iterative Recon: System Resolution Modeling

Distance-dependent collimator beam

$$R_s = \sqrt{R_i^2 + R_c^2}$$



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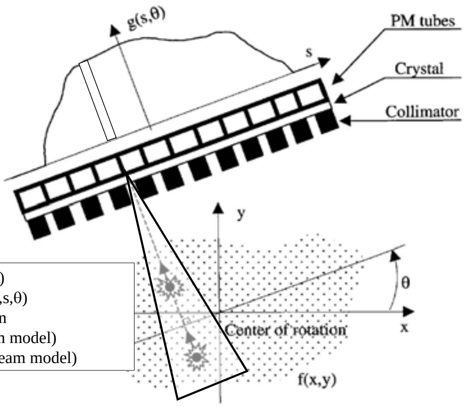
Intrinsic Detector Resolution R_i

Pencil Beam (FBP)

Fan Beam (2D iterative)

Cone Beam (3D iterative)

SPECT Iterative Recon: Resolution Modeling



2D: $g(s,\theta) = \Sigma f(x,y) * p_{res}(x,y,s,\theta)$
3D: $g(s,\theta) = \Sigma f(x,y,z) * p_{res}(x,y,z,s,\theta)$
p_{res} = probability due to resolution
"fan of acceptance" (2D fan beam model)
"cone of acceptance" (3D cone beam model)

SPECT Imaging: Scatter

- Scatter compensation occurs before attenuation
  - the photopeak window contains scatter
  - attenuation accounts for the removal of photopeak photons
- Scatter contribution estimated as a weighted sum of one or more adjacent energy window images, $C_i(x,y,\theta)$
  $S(x,y,\theta) = \Sigma_i\, k_i \times C_i(x,y,\theta)$
- Subtract scatter prior to reconstruction
  $P_{corr}(x,y,\theta) \rightarrow P(x,y,\theta) - S(x,y,\theta)$
- Incorporate scatter into forward projection
  $P(x,y,\theta) \rightarrow P_{corr}(x,y,\theta) + S(x,y,\theta)$

SC techniques:
DEW
TEW
ESSE

SPECT Iterative Reconstruction

- True projection intensity = sum of true voxel intensities weighted by detection probabilities

Forward Projection

$$y(d) = \sum_{b=1}^{B} \lambda(b)\, p(b,d)$$

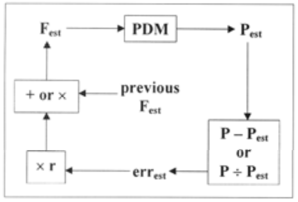
- True voxel intensity = sum of true detector intensities weighted by detection probabilities

Back Projection

$$\lambda(b) = \sum_{d=1}^{D} y(d)\, p(b,d)$$

Iterative Reconstruction Flow Diagram

$$\lambda^{[k+1]}(b) = \frac{\lambda^{[k]}(b) \sum_{d=1}^{D} \frac{y(d)\, p(b,d)}{\boxed{\sum_{b'=1}^{B} \lambda^{[k]}(b')\, p(b',d)}}}{\sum_{d=1}^{D} p(b,d)}$$



In clinical practice, the stopping criteria is number of iterations (a time constraint) instead of a convergence criteria.

Ordered Subset EM (OSEM)

- Each OSEM iteration is a ML-EM iteration using an ordered subset of n (out of N) projections (eg: 4/36 views - 9 subsets, start with 0°,90°,180°,270° views)
- The next OSEM iteration starts with the result of the previous OSEM iterations but uses a different ordered subset of n projections (next set uses 10°,100°,190°,280° views)
- ↑ rate of convergence by using an ordered subset of all N projections for each iteration
- m OSEM iterations with n subsets each ≅ m×n ML-EM iterations using all N each time

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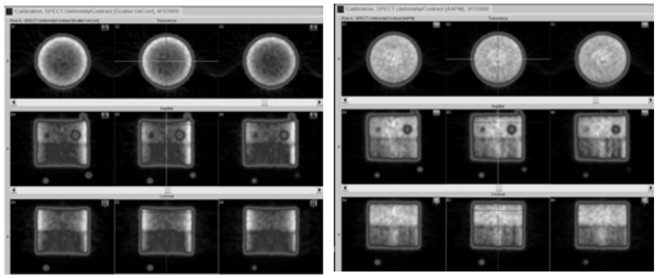
OSEM Iterative SPECT Reconstruction: Attenuation and Scatter Correction

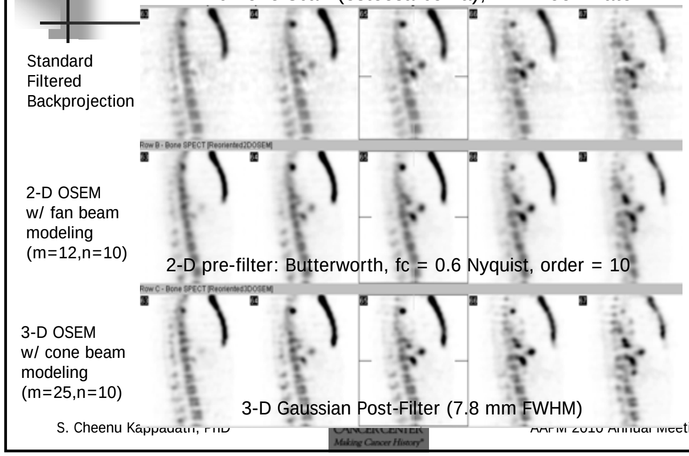


Un-Corrected Corrected

Note the “hot-rim” artifact

OSEM Iterative SPECT Reconstruction: Collimator Resolution Modeling

^{99m}Tc Bone Scan (osteosarcoma), LEHR Collimator



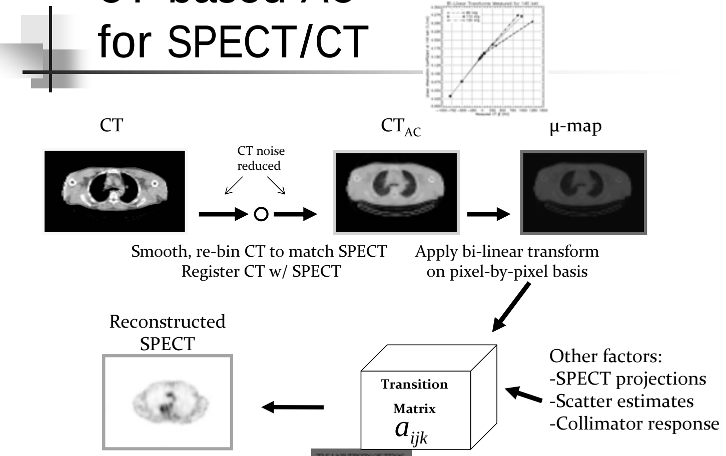
2-D pre-filter: Butterworth, fc = 0.6 Nyquist, order = 10

3-D Gaussian Post-Filter (7.8 mm FWHM)

SPECT/CT Hybrid Imaging: Why?

- Non-uniform attenuation maps required
  - Previous methods used constant μ maps that work for brain but are problematic for thorax and pelvis
  - radioactive source-based transmission CT – time penalty
- Functional-anatomical overlay (image fusion)
  - Improve localization of uptake regions
  - Increase confidence in interpretation

CT-based AC for SPECT/CT



CT

CT noise reduced

CT_{AC}

μ-map

Smooth, re-bin CT to match SPECT
Register CT w/ SPECT

Apply bi-linear transform
on pixel-by-pixel basis

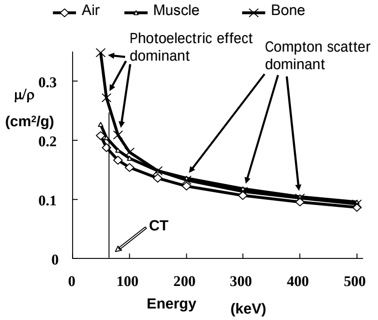
Reconstructed SPECT

Transition Matrix a_{ijk}

Other factors:
-SPECT projections
-Scatter estimates
-Collimator response

CT-based μ values

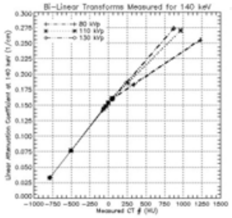
Material attenuation versus Energy

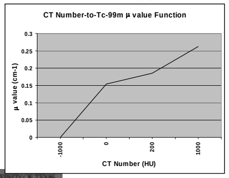


- m = k × CT-HU
(simple but not accurate)
- Compton Scatter probability proportional to e$^-$ density
- Photoelectric effect probability proportional to $(Z/E)^3$
- Attenuation mismatch between PE and CS with energy for high Z

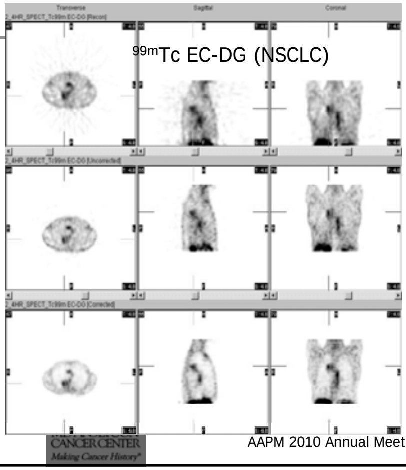
CT-based μ values

- HU-to-cm^{-1} conversion
  - not linearly related
  - piece-wise linear
  - bi- or tri-modal
- Effective energy differences
  - CT (~ 70 – 80 keV)
  - SPECT (nuclide dependent)
    eg: 140 keV for Tc-99m





SPECT/CT Hybrid Imaging:
Iterative Reconstruction



^{99m}Tc EC-DG (NSCLC)

FBP w/
Butterworth 0.4/5

3-D OSEM w/
resolution modeling

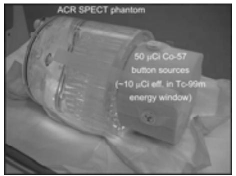
3-D OSEM w/
resolution and attenuation
modeling

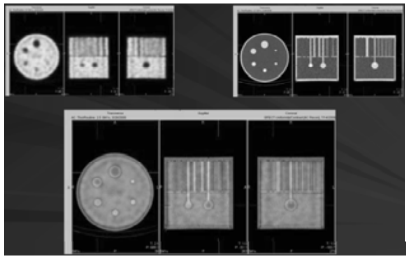
SPECT/CT QA/QC

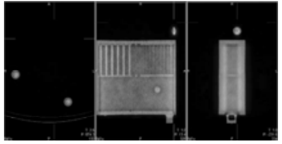
- Planar (AAPM Reports 6 and 9; NEMA NU 1-1994)
  - Inherently includes all planar gamma camera QA
  - Energy/Spatial resolution, uniformity, deadtime, sensitivity, rotational uniformity, opposed-head registration, etc.
- SPECT (AAPM Report 22 and 52)
  - Uniformity and Contrast
  - Resolution
- SPECT/CT (AAPM TG 177: Jim Halama)
  - NM-CT registration
  - CT-HU to linear attenuation (μ) transformation

NM-CT Registration

- Use Co-57 button sources w/ SPECT phantom



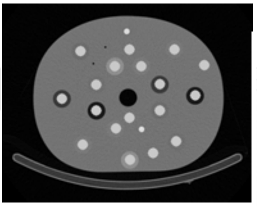




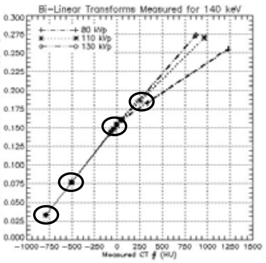
CT-HU to μ-map transformation

- Use an electron density phantom

CIRS Inc.



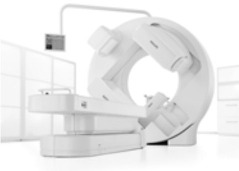
CT image: -790 to 235 HU



Commercial SPECT/CT systems

GE Hawkeye
(1- or 4-slice CT)

Philips BrightView
(Flat-panel CT)



Siemens SymbiaT
(1-, 2-, 6, 16-slice CT)

GE – Millennium VG Hawkeye

- NM
  - 3/8” and 1” NaI(Tl) crystals
  - 16 simultaneous energy windows
  - Slip-ring gantry
  - Body-contouring based on infrared-based transmitters
- CT
  - Co-planar, dental tube, 4-slice 20 mm beam
  - no additional real estate needed
  - Resolution: 3.5 or 1.75 mm (transaxial); 5 or 10 mm (axial)
  - Time-averaged: 23 s per rotation (slow-scan)
  - kVp: 120 – 140; mA: 1 – 2.5

Phillips – BrightView XCT

- NM
  - 3/8” and ¾” NaI(Tl) crystals
  - Energy-independent flood calibration (up to 300 keV)
  - 15 simultaneous energy windows
  - Body-contouring based on tissue impedance
- CT
  - Co-planar, flat-panel detector, 14 cm axial FOV
  - no additional real estate needed
  - High-resolution: 0.33 mm isotropic voxels
  - Time-averaged: 12 s or 24 s per rotation (slow-scan)
  - kVp: 120; mA: 5 – 80

Siemens - SymbiaT

- NM
  - 3/8” and 5/8” NaI(Tl) crystals
  - Energy-independent flood calibration (up to 300 keV)
  - 6 simultaneous energy windows
  - Body-contouring based on infrared-based transmitters

- CT
  - Diagnostic CT scanner
  - kVp: 80/110/130; mA: 20 – 345 (T16) & 30 – 240 (T6)
  - Scan time: 0.5, 0.6, 1, 1,5 s per rotation
  - 1-, 2-, 6-, and 16-slice CT scanners

Clinical SPECT/CT Imaging

- Stress/Rest Myocardial Perfusion Imaging
  - Stress: ^{99m}Tc-sestaMIBI or ^{99m}Tc-Tetrafosmin
  - Rest: ^{99m}Tc-labeled agents or ^{201}Tl-chloride
- ^{99m}Tc-MDP: bone diseases, bone metasteses
- ^{99m}Tc-sestaMIBI: parathyroid adenomas
- ^{99m}Tc-sulphur colloid: liver/spleen, lymphoscintigraphy
- ^{111}In-Pentetreotide: neuroendocrine cancers
- ^{111}In-ProstaScint: prostate cancer
- ^{123}I/^{131}I-MIBG: pheochromocytoma, neuroblastoma
- ^{131}I-NaI: thyroid cancer

Clinical SPECT/CT Imaging

- ^{99m}Tc-CEA: colorectal cancer
- ^{99m}Tc-RBCs: hemangioma
- ^{99m}Tc-HMPAO, -ECD: brain perfusion
- ^{111}In-WBC: infection
- ^{67}Ga-citrate: inflammation, lymphoma
- ^{201}Tl-chloride: tumor perfusion

Clinical Benefits of SPECT/CT

- Visualization, diagnosis and interpretation of primary and metastatic diseases
  - higher sensitivity and contrast than Planar imaging
  - CT scan increases confidence in interpretation of SPECT examination
- Surgical planning and IMRT treatment planning
- ^{90}Y-microspheres radio-embolotherapy (selective internal RT or micro-brachytherapy)
- Internal radio-pharmaceutical therapy planning

SPECT/CT: Limitations

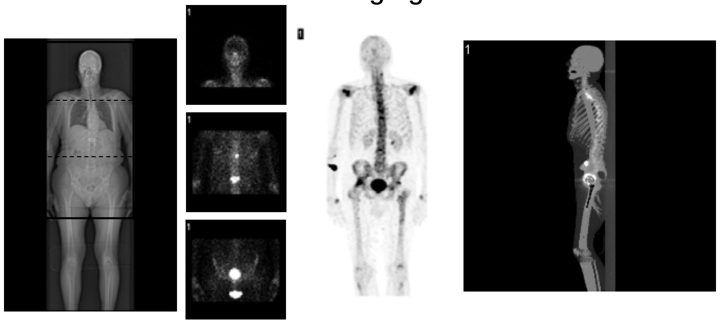
- Patient motion
  - between SPECT and CT scans
  - respiratory and cardiac motion during SPECT acquisitions
- Contrast CT
  - contrast introduces electron density-material mismatch
  - μ map algorithms do not yet account for contrast CT
- Absolute quantification (Bq/ml) not yet fully developed
  - radionuclide-dependent
  - acquisition/reconstruction technique-dependent
  - calibration techniques not yet standardized

SPECT/CT: Future Applications

- Whole body SPECT/CT (analogous to PET/CT)
- Quantification of absolute activity (like PET)
- Compensation for CT contrast in μ map
- Compensation for respiratory, cardiac motion
- SPECT/CT-based 3-D dosimetry/treatment planning

Future: Whole-body Bone SPECT/CT

- Tc-99m MDP Bone Imaging



Future: Multi-nuclide SPECT/CT

- Maximum Intensity Projection (MIP) of a dual-isotope (Tc-99m and I-123) SPECT/CT mouse study.
- Published by the Molecular Imaging Center for Excellence newsletter, SNM publication Volume 2, 2008