Positron Emission Tomography
Principals and History

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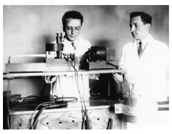
Overview

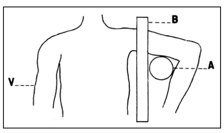
The Tracer Principle

- Use of a radioactive compound to measure in-vivo processes
- 1913 First use of radioactive isotopes by George de Hevesy – works on Pb-salt soluability with a Pb-isotope.
- 1923 (De Hevesy) Pb-salt uptake in Plants
- 1924 Blumgart and Yens / Weiss – Bi 212 into arm of patient – measured arrival in other


www.nobelprize.org


Patton, JNM 2003

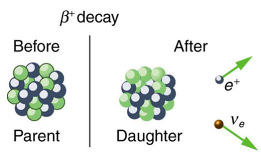
STUDIES ON THE VELOCITY OF BLOOD FLOW
I. THE METHOD UTILIZED[1]
BY HERRMANN L. BLUMGART AND OTTO C. YENS
(From the Thorndike Memorial Laboratory, Boston City Hospital and the Department of Medicine, Harvard Medical School)
(Received for publication October 4, 1926)

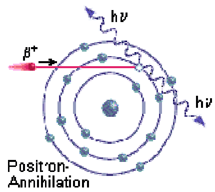


A radioactive tracer can be tracked to study dynamic processes

The Beta^{+} decay

For nuclei with a fixed mass number a proton can transform to a neutron = β^{+} (and conversely = β^{-})





- e^{+} antiparticle to e^{-} -> annihilation
  - E = mc^2 = 2 x 511keV
  - Momentum conservation -> ~ 180°

Beta+ decay: Changing a proton into a neutron

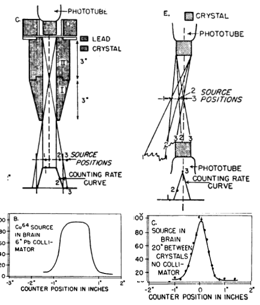
Coincidence Counting

- First described by Wrenn et al. (Science 1951) for localization of Brain tumors
- e^{+} decay leads to 2 perpendicular γ-rays
- If detected, decay took place on a line between detectors
- Equivalent to a projection

The Use of Positron-emitting Radioisotopes for the Localization of Brain Tumors[1]

Frank R. Wrenn, Jr.,[2] Myron L. Good, and Philip Handler

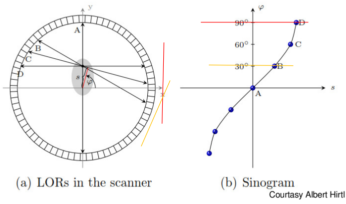
Division of Neurosurgery, Departments of Physics and Biochemistry and Nutrition, Duke University, Durham, North Carolina

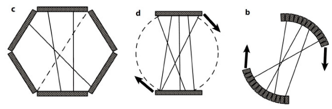


Coincidence counting requires no collimation for directional information

PET Principle

- PET relies on detection of two photons from positron annihilations
- Most common: ring geometry
- Annihilation on a line (LOR) between the detectors
- Parallel lines belong to same projection
- Data most common represented as "Sinograms"





No need for a collimator to gain projections in coincidence counting

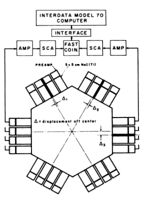
The first PE(T)T

- First described by Phelps et al. JNM 1975
- Prototype of a positron emission transaxial tomograph (PETT)
- Tested on phantoms and dogs
- Reconstructed using Fourier transformation and attenuation corrected using ^{64}Cu source

APPLICATION OF ANNIHILATION COINCIDENCE DETECTION TO TRANSAXIAL RECONSTRUCTION TOMOGRAPHY

Michael E. Phelps, Edward J. Hoffman, Nizar A. Mullani, and Michel M. Ter-Pogossian

Washington University School of Medicine, St. Louis, Missouri



The first positron emission tomograph was described in 1975

Reconstruction

General Problem:

Get a 2D image out of 1D projections

- ➢ In principle solved mathematically by Radon 1917
- ➢ Radon transformation

SITZUNG VOM 30. APRIL 1917.

Über die Bestimmung von Funktionen durch ihre Integralwerte längs gewisser Mannigfaltigkeiten.

Von

Johann Radon.

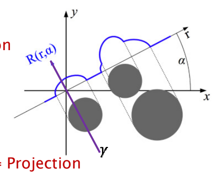
Integriert man eine geeigneten Regularitätsbedingungen unterworfene Funktion zweier Veränderlichen x, y — eine Punktfunktion

Activity distribution

$$R(\alpha, r) = \int_{\gamma} f(x,y)d\gamma$$

Projections

Line integral through activity concentration = Projection



Invers Radon transformation = FBP

Reconstruction: FBP

Back projection -> Blurred images

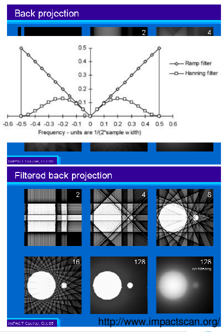
Solution: FBP

Inverted radon transformation

$$f(x,y) = \frac{1}{4\pi^2}\iint G(\alpha,\omega)\, e^{i\omega(x\sin\alpha - y\cos\alpha)}|\omega|\, d\omega\, d\alpha$$

Filter!!!

Projections in Frequency space



Filter:

➢reduces low frequencies = blurring

➢enhances high frequencies = sharp edges

!Enhances noise

!Limited # projections -> Cut of frequency

FBP is fast and easy but sensitive to noisy data

Iterative Reconstrucion

Getting a projection from an image can be written as a matrix multiplication

System-matrix: representing the PET

$$g_i = \sum_{j=1}^{K} c_{ij}\lambda_j$$

Projection

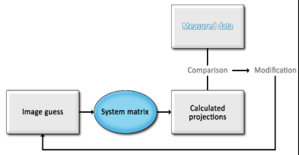
Image (activity distribution)

Problem:
inversion of matrix not possible!!!

Solution:
Iteratively finding the image which fits the measured projections

Advantages:

- less affected by noice
- physical properties (scattering, absorbtion, resolution...) can be directly implemented into sytem-matrix



Changing an image until the resulting projections fit the measured data

The OS-EM algorithm

Maximum likelihood expectation maximisation

Algorithm for iterative reconstruction described by Sepp and Vardi 1982

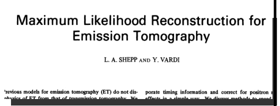
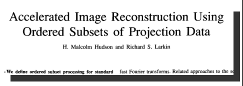


Image update:

comparing and correcting with all projections

➢slow convergence



Ordered Subset:

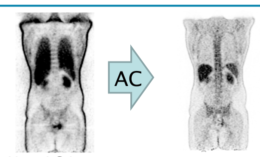
A compromise described by Hudson and Larkin 1994

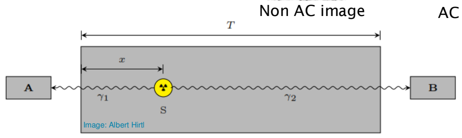
- a subset of projections is taken for correction
- ➢ Good noise properties + fast convergence

EM is the how to update the image, OS is for acceleration

Attenuation Correction

- Photons get attenuated within the body
- Attenuation in coincidence counting is only depending on the total attenuation along a LOR

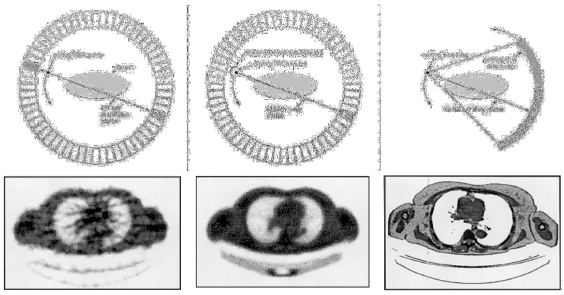




$$P_{detect} = P_{detect\gamma_1} \cdot P_{detect\gamma_2} = e^{-\mu x} \times e^{-\mu(T-x)} = e^{-\mu T}$$

Attenuation correction is essential for quantitative images

Attenuation correction



(a) ^{68}Ge/^{68}Ga positron source (b) ^{137}Cs gamma-ray source (c) 120 kVp x-ray source

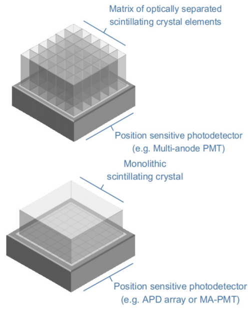
AC is done using a transmission scan

PET Detector

Transfer 511keV photons into electric signal

Most common devices consist of:

- Scintillator:
  converts high energy photons into visible light
- Photo detector:
  converts visible light into electric signal



A detector converts a 511 keV photon into an electrical signal

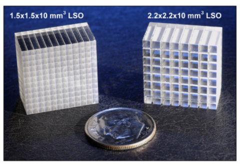
Scintillator material

Scintillator material needs certain properties

•High interaction property with 511keV
- ➢ Density -> stopping power

•High light output
- ➢ Energy resolution
- ➢ Spatial resolution

•Fast scintillation
- ➢ Dead time
- ➢ Timing resolution



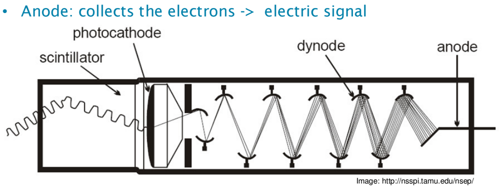
Scintillator	Decay time (ns)	Attenuation coefficient (cm^{-1})	Light output (photons/MeV)
NaI(Tl)	230	0.35	41000
BGO	300	0.95	7000
GSO	60	0.70	10000
BaF_2	2	0.45	2000
LYSO, LSO	40	0.86	26000
$LaBr_3$	20	0.47	60000

Fast detectors with high light output

Photo multiplier tubes

Most commonly used

- Photocathode: converts visible light quants into free electrons
- Electric field: accelerates electrons towards the dynode (an anode)
- Dynode: electron impact causes secondary electron production (amplification)
- Anode: collects the electrons -> electric signal

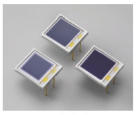


Electron acceleration and amplification

Some PMT alternatives

APD: Averlange photo detectors

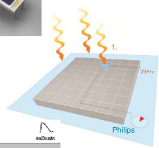
- Based on semiconducturs
- \+ better spatial resolution
- \- low gain
- \- slow compared to PMTs



APD

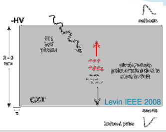
SiPM: Silicon Photomultiplier

- Grid of APDs operated in “Geiger-mode”
- \+ analog or digital readout
- \+ fast



Direct semiconductor detectors

- \+ high spatial resolution
- \- low efficiency



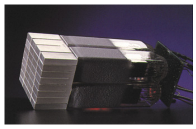
There are alternatives to PMTs based on semiconductors

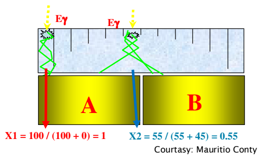
PET system design

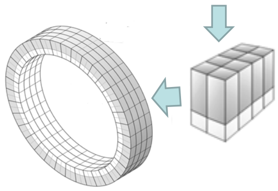
One scintillator cristall mounted on four PMT

➢Detector block

Position of photon interaction – Anger logic







All clinical PET/CT today are whole-body ring systems

Type of Events

True coincidences are usable

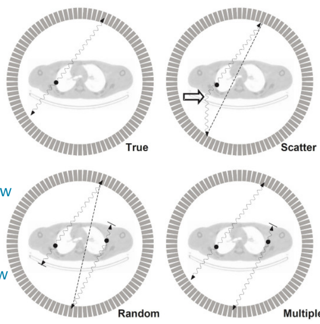
- Time window ~4ns
- Energy window ~450keV-630keV

Scatter

- minimized by energy window

Random

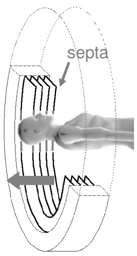
- Minimized by timing window



PET/MR quantification is not straight forward

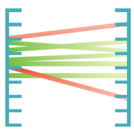
2D- versus 3D-PET

Courtesy T.Beyer (Vienna)



Multi-ring tomograph
\+ retractable septa

2D (septa extended)



3D (septa retracted)

- increased sensitivity
- increased background
- increased deadtime
- out-of-field activity
- 3D reconstruction
- non-uniform response
- BGO works for brain
- LSO/GSO for whole-body

State-of-the-art PET: 3D (no septa) for increased sensitivity

Limiting Factors

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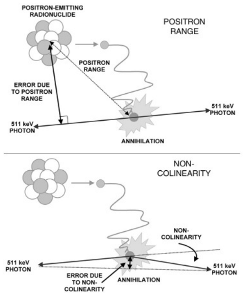
Positron Range:

Depending on E_{kin} of positron

Nuclide	Mean range
18-F	0.6 mm
68-Ga	2.9 mm
11-C	1.1 mm

Non collinearity:

- •Residual momentum of positron causes deviation from 180° - preservation of momentum
- •Nearly Gaussian (FWHM~0.5°)
- •Depending on ring diameter



Positron emission ≠ positron annihilation

Limiting Factors

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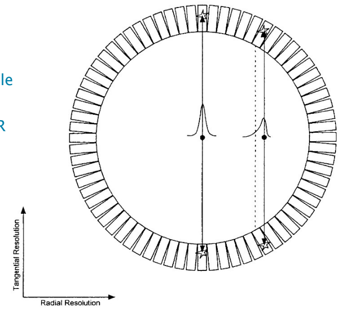
Detector parallax:

- Detectors shifted in angle
- Photon passing first scintillator -> wrong LOR
- asymmetric response function



Resolution depends on system design and geometry

Point spread funciton: PSF

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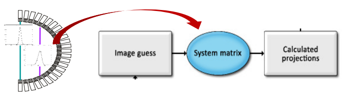
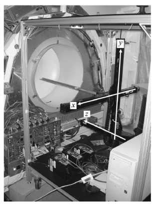
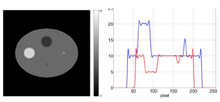
Incorporation of the PSF into system matrix





Panin et al. IEEE 2006

- •PSF:measured, MC simulations, modeled
- •accounting for varying spatial resolution
- •E.g. Siemens TrueX Algorithm (now "HD PET")
  - \+ better resolution -> detectability of small volumes
  - \- Gibbs artifacts at sharp edges



Tong et al. IEEE 2011

PET/MR quantification is not straight forward

Time-of-flight (TOF)

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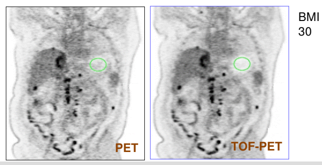
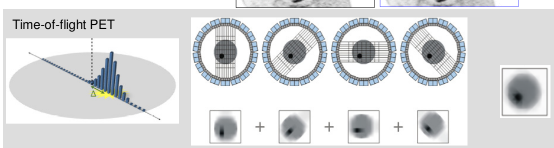
$$SNR_{TOF} \cong \sqrt{\frac{D}{\Delta x}} \cdot SNR_{conv}$$

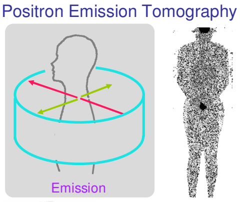




TOF-PET improves SNR (not spatial resolution)

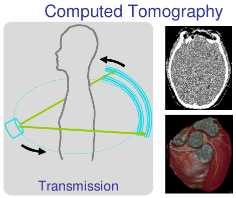
Why PET/CT ?

Positron Emission Tomography



Functional anatomy
Changes in metabolism
High functional resolution
Early detection is possible

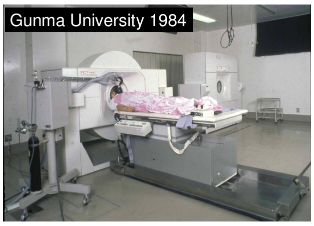
Computed Tomography



Spatial anatomy
Changes in anatomy
High spatial resolution
Late(r) anatomical changes

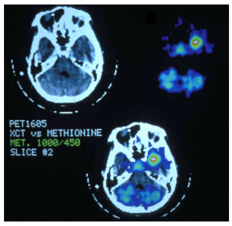
PET/CT is *function* plus *anatomy*

Early PET/CT development



A combined CT and PET system developed by Prof. Teruo Negai (Dept of Radiology, Gunma University, Japan) in 1984.

The device incorporated CT and PET scanners from Hitachi Inc and the patient bed moved on floor-mounted rails between the PET and CT.

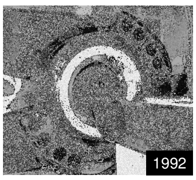


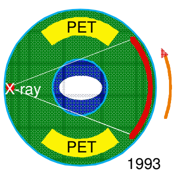
Fusion images of a patient with a brain tumor.

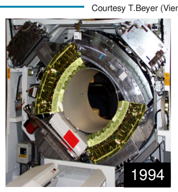
^{11}C-methionine PET and CT acquired in the same examination.

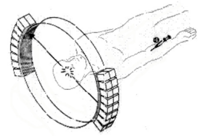
Courtesy Prof. Y Sasaki

Prototyping PET/CT







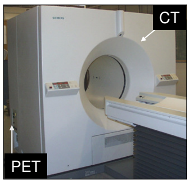


A rotating PET scanner using BGO detectors, Townsend DW et al, JNM 30, 1993

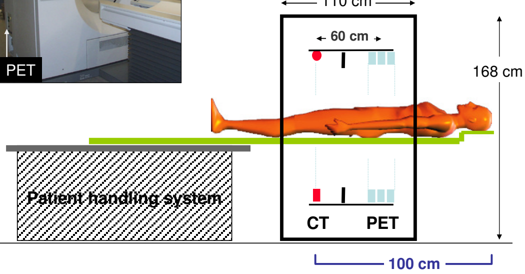
The evolution of a PET/CT concept:1992-4

- Routine oncology imaging (whole-body)
- Complementary PET/CT
- Accurate, fast CT-based quantitation

The SMART PET/CT

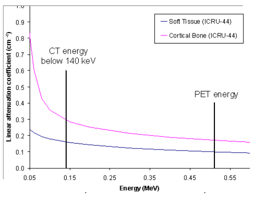


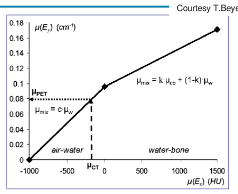
SMART
→ Somatom CT
→ Advanced Rotating Tomograph (PET)



CT-based attenuation correction

Courtesy T.Beyer (Vienna)





Townsend DW, PMB 53: R1-39, 2007

bi-linear segmentation-scaling approach to CT-AC

Early PET/CT concepts

Courtesy T.Beyer (Vienna)

1998-2000

300 oncology patients

~50% with CT contrast

Torso exam in 60 min

CT: 80 cm = 4 min,

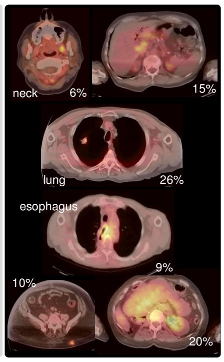
PET: 7 beds = 50 min

BGO: 7x7x20 mm³

1-slice CT, 15 kW

Power of fused images

CT-based artifacts



PET/CT systems

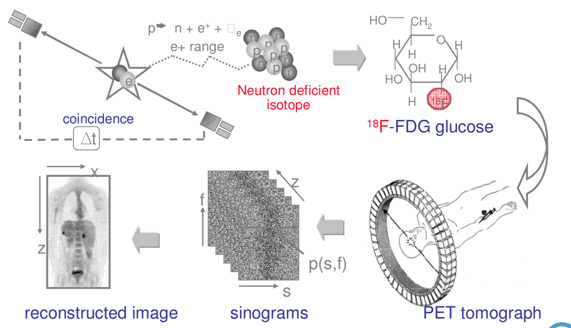
Courtesy T.Beyer (Vienna)

Discovery VCT	Ingenuity TF	AnyScan	Biograph mCT
CT: 16-128 slices	CT: 16-128 slices	16-slice CT	CT: 20-128
70 cm patient port	70 cm (85 cm) patient port	70 cm diameter patient port	78 cm patient port
250 kg table weight limit	215 kg table weight limit	250 kg table weight limit	250 kg table weight limit
170 cm co-scan range	190 cm co-scan range	360 cm co-scan range	170 cm co-scan range
24 rings of LYSO(Ce)	44 rings of LYSO(Ce)	24 rings of LYSO(Ce)	52 rings of LSO (Ce) crystals
4.2 x 6.3 x 25 mm³	4.0 x 4.0 x 22 mm³	3.9 x 3.9 x 20 mm³	4.0 x 4.0 x 20 mm³
Time-of-flight	Time-of-flight		Time-of-flight
15.1 cm axial FOV	18 cm axial coverage	23 cm axial coverage	21.6 cm axial coverage
70 cm transaxial FOV	67 cm transaxial FOV	55 cm transaxial FOV	70 cm transaxial FOV
PET resolution model	PET resolution model		PET resolution model

High-end PET combined with high-end, multi-slice CT

Summary

Courtesy D Townsend (UT Knoxville, USA)



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