COMPUTERIZED TOMOGRAPHY SCANNERS

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I - INTRODUCTION -

Among the medical imaging techniques, these last years saw an important newcomer with the advent of Computerised Tomography.

The imaging process used to obtain classical radiograms had not changed in its rough lines since X-rays discovery in 1896 : a point source irradiates an object, which casts a shadow on a planar detector.

All the modern radiological tools keep these fundamentals, with improvements on the source (X-ray tubes), on the detectors (film changers, image intensifiers), or on the way to superpose such different two-dimensions projections by mechanical devices (tomography).

The CT Scanners also tend to provide radiological images ; but the major difference brought by CT is that it is an indirect and quantitative method : a large set of individual measures is obtained, then processed to finally provide a digital representation of the object on which the measures had been made.

All that concerns a "slice" of the object, as in classical tomography ; but it allows to obtain true measures of X-ray attenuation coefficients (densitometry). That's why it has also been proposed to name this technique TomoDensitoMetry (TDM).

II - THEORY -

The principle under which the CT machines work is the reconstruction of a 2-dimensionnal function from all its linear projections inside its definition plane.

A - PROJECTIONS-

The absorption of monochromatic X-ray photons by objects obey the Beer-Lambert law (fig. 1).

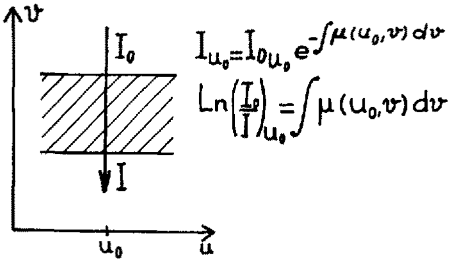

Fig. 1
X-Rays Photons Attenuation

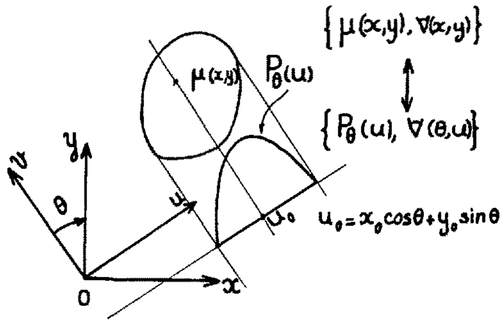

Fig. 2
Projections

μ is the linear attenuation coefficient.

If we consider a section of a finite object, it is caracterised by $\mu(x,y)$ in any point. The set of integrals of μ along $u = u_0$ for all values of u_0 constitutes the projection $P_\theta(u)$ of the object along the direction Ov defined by $(Ov, Oy) = \theta$ (see fig.2).

The relation giving all $P_\theta(u)$ from all $\mu(x,y)$ is linear ; it admits an inverse which gives $\mu(x,y)$ in any point from all projections.

B - ALGORITHMS

After some trys on iterative methods, which had proven to be valuable in similar problems with high symetries (Electron microscopy of viruses), the generally used algorithm is the filtered back-projection, deriving from the works of Radon (1), Ramachandran (2), and Shepp and Logan (3).(Application of "Central Slice" theorem).

If $\mathcal{P}_\theta(\omega)$ is the Fourier Transform of $P_\theta(u)$,

$$\mu(x,y)=\frac{1}{4\pi^2}\int_0^\pi d\theta\int_{-\infty}^{+\infty}\mathcal{P}_\theta(\omega)e^{i\omega u}|\omega|\,d\omega \qquad \text{with } u=x\cos\theta+y\sin\theta$$

Obviously, it is possible to replace the ω integral (inverse Fourier Transform) by a direct convolution product of $P_\theta(u)$ by $\varphi(u)$, where $\varphi(u)$ is the inverse Fourier Transform of $\Phi(\omega)=|\omega|$.

From over, it is possible to decompose the algorithm in two phases :

- Computing $(P_\theta * \varphi)(u)$, directly or by Fourier Transform, $|\omega|$ multiplication, then inverse F.T.

- For a given θ (that is for one projection) adding the contribution of $(P_\theta * \varphi)$ (u) in any point (x,y) so that $x\cos\theta + y\sin\theta = u$, i.e along a line identical to the projection line.

The first operation is generally called convolution (or deconvolution or filtering), and the second : back-projection.

In real procedures, the needed precision is for now uncompatible with analog methods. It is, so, necessary to turn to digital, and, for that, to sample along projection angles (θ), along each projection (u), and inside the finite reconstruction domain.

This asks for a limitation (Shannon frequency) on the maximum ω in the physical measuring process, and drives to replace the $|\omega|$ value of Φ by a new Φ function, product of $|\omega|$ by a low pass filter (and similarly for $\varphi(u)$). Special care must be given to this filter, because it will determine the bandwidth of the system, and the "damping" of its transfer function. It will play a role similar to the apodizing functions in optics.

Another important problem comes from the determination of the value to be back projected in the (x_i, y_j) position, as in general $U_{ijm} = x_i \cos\theta_m + y_j \sin\theta_m$ is not equal to a u_n position along a projection. A compromise must be found between the precision and the computing time to interpolate $P_{\theta_m} * \varphi$ so as to know its value in U_{ijm} from the nearest positions u_n and $u_n + 1$, where $u_n \leqslant U_{ijm} \leqslant U_n + 1$.

III - BASIC STRUCTURE AND DESIGN PRINCIPLES OF A CT MACHINE

A CT-Scanner must :

- 1 - make densitometric measurements useable as projections, with :
  - X-ray producing equipments, beam conditionning devices, and X-ray detector(s), measuring the photons after object traverse.
  - A mechanical gantry, supporting the preceidings, and moving it along to allow the measure of all points of all projections.
  - An object support (the patient couch).
- 2 - ensure data management and algorithm computation, with :
  - Measuring electronics connected to the detector(s),
  - General electronics to ensure various supplies, synchronisations, interfaces, transfers, and receive orders from the operator.
  - Computing systems (including fast processors if needed).
- 3 - provide the results as interpretable pictures :
  - The reconstructed values are sent to a storage peripheral and a visualisation console (see later).

The quality of the results will be the consequence of all characteristics and performances.

Some of those limitations are pure technology, but the main ones will result from the interaction between physical limits and some severe restrictions given by the nature of the object, generally a living patient.

The theorical precision would be infinite, less the counting statistics of X-ray photons in measures. But if their number goes too high, patient exposures will also be too high.

A compromise must be found between acceptable doses and precision, for both low contrast and high frequencies resolution.

As the quantum noise on counting N events can be described by a Poisson law with a variance N, it is possible to evaluate the noise on the resulting image, for a given bandwidth. It can be established that the standard deviation is $\sigma \propto \sqrt{\omega_s^3/D}$ where D is the patient dose. This shows that for a given dose, it is important to choose the noise/bandwidth trade off.

Another important problem is the choice of photons energy. All the theory is based upon monochromatic photon sources. The optimum energy is a compromise between transmission and contrast, to achieve the best signal to noise ratio for a given dose. The nature of the detected contrasts will vary with the energy because of the different attenuation mechanisms (photoelectric absorption-elastic or inelastic scatter). The values of these contrasts ask for very high amounts of photons (typically $1.5\ 10^{10}$ to 10^{12} photons per slice without object), with energies in the 60-90 keV range. The last point to determine the X-Ray source is the measuring time, which must be short for economic reasons (patient throughput) and to avoid patient moving. Typical times go from 5 minutes to 20 Seconds for head examinations, and must be less than 5 seconds (down to about 50 ms if possible for body examinations.

The only practically useable X-ray source is the vacuum X-ray tube. Its large emission spectrum is adapted by using filters, but with low selectivity to keep high efficiency. It is then necessary to correct for the polychromaticity errors by digital ways.

The detectors must too have high efficiency ; the photon rates (5.10^7 to 10^{12}per second) are too high for a counting use ; so, it is necessary to convert their analog signal to digital before use.

<u>The detectors</u> in use can be - photodectors associated with scintillators,
- gas ionisation chambers.

In the first category, the scintillators had been NaI(Tl), then plastic scintillators or fluorite, and now ICs or, mainly, Bismuth Germanate. The points to be careful with are : high efficiency, low light absorption, and low decay time. The associated photodetectors are generally Photo Multipliers, and sometimes now photodiodes.

The second type,used only for multicells detectors are Xenon ionisation chambers used in the true ionisation mode. The high efficiency is obtained by high pressure (10 to 30 atmospheres) and thickness (10 to 100 mm). The main advantage is the simple structure giving directly an electric current. The collection time must remain low, but can be corrected (pure delay).

The scanning movements, combined with the detectors's structure, must allow to obtain the set of projections under a total half turn (minimum).

The number of projections and the number of samples per projection must be determined in relations with the desired precision on the reconstructed image ; current values are 128 to 512 projections by half-turn (sometimes completed on a whole turn or more), with 128 to 1024 samples per projection.

The simplest (and oldest) way to obtain a projection is to dispose one detector facing the tube, with the beam shaped by collimations. They are fixed on the same mechanical parts, and translate linearly, the beam going throuhg the field ; after each translation, the gantry rotates of an elementary angle, then a new translation is performed, and so on. The sequence of signals during the translation constitutes one projection. This way is cheap and simple, but slow :about 1 second per traverse will give a many minutes exploration time (fig. 3).

Next method on the same principle uses a number of detectors, to make the same number of projections at each translation, the elementary rotation being also multiplied by the same factor. Generally, this number is from 3 to 60, and it takes from 1 minute to 10 seconds per exploration.

Another principle is to suppress the translation, with as many detectors as samples per projection. This can be interpreted as either a conical projection in place of the parallel one, or by rearranging the samples to give back parallels projections.

These machines use 256 to 1024 detector cells, but suppress one mechanical movement. Acquisition time is down to 1 to 5 seconds (fig.4).

The last used principle is to dispose the detectors continuously on a stationnary circle ; only the X-ray tube rotates to give the different projection angles (Fig. 5). It asks for 700 to 2 000 detectors, with the same times as the preceiding.

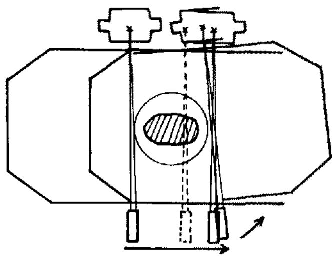
Fig. 3

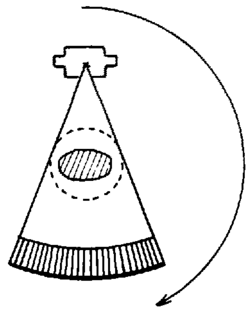
Fig. 4

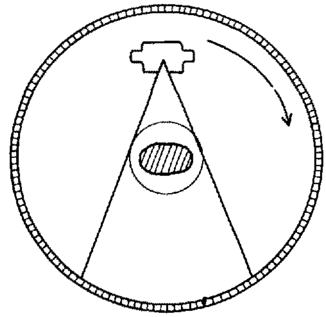
Fig. 5

IV - VISUALISATION

From the set of projections, the algorithm procedure allows to reconstruct the value of μ (x,y) on a sampling raster ; the number of reconstructed points is now generally between 256 × 256 and 512 × 512 (actually this number multiplied by $\pi/4$ because it has sense only in a circle).

The number of bits per point must be consistent with the precision available, which asks for 10-12 bits. The values of μ are generally expressed in a conventionnal scale called Hounsfield scale, which derives linearly from the linear attenuation values. In this scale, air is -1000, and water 0.

These values are systematically represented as continuous pictures by a zero-order interpolation (square picture elements-pixels-), analog conversion from a refresh memory and CRT viewing.

To avoid pseudo contouring effects, it is preferable to convert a minimum of 6 bits (or 8 for safety). As the values interesting the user are generally concentrated on a small interval, the digital values are treated between refresh memory and conversion by a multiplication plus offset and truncation. This is called contrast (or density) window ; the window width is the difference between the first non black value and the first white value ; the window mean is the middle of this interval. It is completely real-time adjustable to exploit the information.

Some other classical digital image treatments are also applied : smoothing, edge enhancement, enlargements ; some digital measuring is provided too : distances, angles, densities on Regions of Interest, density profiles, density histograms etc... All theses functions are called by the user on real time image consoles.

V - RESULTS - DISCUSSION

Results quality is evaluated in terms of imaging process : geometric resolution, noise, and contrast transmission ; but any performance figure has meaning only if accompanied by equivalent energy (to know the theorical values for a given object) and patient dose. The noise is described by the standard deviation on a uniform zone ; it is given in percentage of water to air difference. Standard values are between 0.1 and 1 %, with repartitions close to Gaussian.

Geometric resolution and contrast transmission are given by the Modulation Transfer function (or the Line Spread function). The cut off frequency can be found between 3 and 10 line pairs per cm, with LSF Full Width at Half Maximum between 1 and 5 mm.

These features are obtained with 60-75 keV equivalent energy and 0,1 to 20 rads maximum skin dose.

At this day, the only use of CT scanners has been for medical imagery. The first available machine (EMI 1971) was dedicated to head examination. This field has remained the best adapted, both for technical reasons (limited dynamic, fixed organs) and medical reasons (lack of other non invasive examinations). Body examinations begin to be interesting too, but the asymptotic quality in this field has not yet been reached, and other techniques are available (for example Ultra-Sound echography). The problems in this application are : large field (about 500 mm) and heavy attenuation (10^{-3} transmission), compared to 250 mm and 10^{-2} for head, and, mainly, the need for the shortest possible examination time ; the fastest human organs movements would ask for 10 to 50 ms examination time to avoid any cinetic blurring. Modern machines with 1 second time however provide a large amount of information, and prove to be quite valuable ; some improvements now on lab study will probably put them soon to a stable phase, with further developements mainly tied with new progresses in general technology.

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