CT Scanner and Applications in Soil Science: Fundamentals, X-ray Sensors and Detectors, Architectures and Algorithms for Image Reconstruction and Scientific Visualization

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What is X-Ray Computed Tomography (CT)?

• Considering a short definition CT, which is also referred to as CAT, for Computed Axial Tomography, utilizes X-ray technology and sophisticated instrumentation, sensors & computers to create images of cross-sectional “slices” through a body under analysis.

• CT exams and CAT scanning provide a quick overview of morphologies and its quantification (also those related to pathologies) and enable rapid analysis and plans for prognostics and support for decision maker.

• Tomography is a term that refers to the ability to view an anatomic section or slice through the body.

• Anatomic cross sections are most commonly refers to transverse axial tomography
What is displayed in CT images?

CT# = \( \frac{\mu_T - \mu_{\text{water}}}{\mu_{\text{water}}} \times 1000 \text{HU} \)

- **Bone window**: C/W 1000, 2500
  - +2250 HU
  - -250 HU

- **Mediastinal window**: C/W -50, 400
  - +150 HU
  - -250 HU

- **Lung window**: C/W -600, 1700
  - +250 HU
  - -1450 HU

**Water**: 0 HU

**Air**: -1000 HU
Some Words About the History of X-Ray Computed Tomography (CT)

"We could limit the story of the beginnings of computed tomography to mentioning Cormack and Hounsfield, the authors of this groundbreaking invention, and to placing their achievements on a timeline, from Cormack’s theoretical idea in the late 1950s to Hounsfield’s development of a practical device in the late 1970s."

- Cormack developed the mathematical technique to reconstruct images using the backprojection method based on a finite number of projections [Cormack, 1963];

- Hounsfield developed the first commercial tomograph [Housfield, 1973].

Radon (1887-1956) presents the mathematical principles of a body reconstruction from their projections considering a space with order equal to $n$ [Radon, 1917];

In mathematics, the **Radon transform** is the integral transform which takes a function $f$ defined on the plane to a function $Rf$ defined on the space of lines in the plane, whose value at a particular line is equal to the line integral of the function over that line (introduced in 1917 and also provided a formula for the inverse transform).

Johann Karl August Radon

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CT uses electromagnetic wave, i.e., X-radiation.

Michael Faraday (1791–1867) observed the phenomenon of electromagnetism and in 1831 formulated the laws of electromagnetic induction.

Twenty-nine years later, in 1860, James Clark Maxwell (1831-1879), formulates the Maxwell’s equations, comprehensively expressed the ideas of electricity and magnetism, which led to the development of the later technologies of radio and television and of course, radiology.
Wilhelm Röntgen (1845-1923) was the first to systematically study the X-rays in 1895.

Production of X-rays and Bremsstrahlung (stopping radiation) – thermal electron emission in vacuum ($10^{-6}$ mbar) and target bombardment

*White X-ray spectrum (gamma quanta with all energies)* and its final view (after tube filtration)

Wilhelm C. Röntgen
The Nobel Prize in Physiology or Medicine 1979
Region of Interest (electromagnetic spectrum) for CT
Introduction to the physical concepts of the tomographic process

- The bases of the X-ray transmission tomography are related to a narrow beam of monoenergetic photons with energy $E$ and a flux of photons $N_0$ passing through a homogeneous body of thickness $x$ (in cm):

\[ N = N_0 \exp(-\mu(\rho,Z_N,E)x) \]
Introduction to the physical concepts of the tomographic process

\[ N = N_0 e^{\int_{\rho} (-\mu(\rho,Z_N,E)x)dp} \]

- If the study body is a chemical component or a mixture, its mass attenuation coefficient can be roughly evaluated from the coefficients of the elements.

| Attenuated photons intensity | Initial photons intensity | Linear attenuation coefficient (in cm\(^{-1}\)) | Material density (g/cm\(^3\)) | Atomic number |

- Initial photons intensity
- Linear attenuation coefficient (in cm\(^{-1}\))
- Material density (g/cm\(^3\))
- Atomic number
Introduction to the physical concepts of the tomographic process

\[
\frac{\mu}{\rho} = \sum_i w_i \left( \frac{\mu_i}{\rho_i} \right)
\]

where \( w_i \) is proportional to the weight of the \( i_{th} \) constituent of the material. The mass attenuation coefficient of a component or mixture can be calculated from the mass attenuation coefficient of the components.
Introduction to the physical concepts of the tomographic process

The differences in linear attenuation coefficients for different materials are energy dependent.

This fact leads to a definition of the contrast in X-ray computed tomography, which is a function of the linear attenuation coefficient values and the mapping process.
Introduction to the physical concepts of the tomographic process

The quality of a tomographic image is correlated to the work energy, as well as the physical characteristics of the samples or bodies under analysis (function of the chemical constituents).
Noise (most significant) of the tomographic process - Poisson noise

• The probability of detecting $N$ photons in an exposure time interval $t$ can be estimated by the Poisson probability distribution function, given by:

$$
\bar{N}_0 = \xi R t
$$

The uncertainty or noise is given by the standard deviation, given by:

$$
\sigma = \sqrt{N_0}
$$
An event can occur 0, 1, 2, … times in an interval. The average number of events in an interval is designated $\lambda$. It is the event rate, also called the rate parameter. The probability of observing $k$ events in an interval is given by the equation:

$$P(k, t) = \frac{\lambda^k e^{-\lambda}}{k!}$$

where:

- $\lambda$ is the average number of events per interval
- $e$ is the number 2.71828... (Euler's number) the base of the natural logarithms
- $k$ takes values 0, 1, 2, …
- $k! = k \times (k - 1) \times (k - 2) \times \ldots \times 2 \times 1$ is the factorial of $k$. 

Probability mass function (PMF) for a Poisson distribution
Sample scanning process

- Two-dimensional reconstruction algorithms use projections to reconstruct the tomographic sections;

- Projections are collected in the interval
  \[ 0^\circ \leq \theta < 180^\circ \]
i.e., getting the Radon transform of the object;

- Through the inverse transform of Radon one can obtain the reconstructed image of the object, based on the attenuation coefficients
Image Reconstruction from Projections

Projection

Detector

X-Ray Source

Inverse path

Projections

$P_0$

$P_m$

$P_n$
Parallel projection of $f(x,y)$ for Radon Transform.
Considering the time domain and take into account the Inverse Fourier Transformation

\[ f(x, y) = \int \int F(u, v)e^{j2\pi(ux+uy)}\,dudv \]

Changing from rectangular coordinate system \((u, v)\) to polar coordinate system \((\omega, \theta)\) and changing the differential variables

\[ dudv = \omega d\omega d\theta \]

\[ f(x, y) = \int \int F(\omega, \theta)e^{j2\pi\omega(x\cos\theta+y\sin\theta)}\omega d\omega d\theta \]
Using discrete forms to represent de projections and rewritten the equations in the frequency domain, is possible to find:

\[ Q_\theta(n\tau) = \tau \times \text{IFFT} \{ \text{FFT}[P_\theta(n\tau)] \times \text{FFT}[h(n\tau)] \} \]

where \( \tau \) is the sampling interval

\[ Q_\theta(n\tau) = \tau \times \text{IFFT} \{ \text{FFT}[\hat{P}_\theta(n\tau)] \times \text{FFT}[h(n\tau)] \} \]

\[ \hat{f}(x, y) = \frac{\pi}{K} \sum_{i=1}^{K} Q_\theta(x\cos\theta_i + y\, \text{sen}\theta_i) \]
Reconstruction from projections

backprojection without convolution  backprojection with convolution
CT without covers
Typical CT architectures

Typical Schematic Representation for Scanning Geometry of a CT System

What are inside the gantry?

Frontal view:
- X-ray tube
- Gantry opening
- Field of measurement
- Anti scatter collimator
- Detector array

Lateral view:
- Shaped filter
- Fixed collimator
- Adjustable collimator
- Center of rotation
- Adjustable collimator
- Fixed collimator
<table>
<thead>
<tr>
<th>Generation</th>
<th>Source</th>
<th>Source Collimation</th>
<th>Detector</th>
<th>Detector Collimation</th>
<th>Source-Detector movement</th>
<th>Advantages</th>
</tr>
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<tbody>
<tr>
<td>1&lt;sup&gt;st&lt;/sup&gt;</td>
<td>single</td>
<td>Pencil beam</td>
<td>single</td>
<td>no</td>
<td>Translation + Rotation</td>
<td>No scatter</td>
</tr>
<tr>
<td>2&lt;sup&gt;nd&lt;/sup&gt;</td>
<td>single</td>
<td>Fan-beamlet</td>
<td>multiple</td>
<td>yes</td>
<td>Translation + Rotation</td>
<td>Faster than 1&lt;sup&gt;st&lt;/sup&gt; Generation</td>
</tr>
<tr>
<td>3&lt;sup&gt;rd&lt;/sup&gt;</td>
<td>Single</td>
<td>Fan-beam</td>
<td>many</td>
<td>no</td>
<td>Rotates together</td>
<td>Faster than 2&lt;sup&gt;nd&lt;/sup&gt; Generation</td>
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<tr>
<td>4&lt;sup&gt;th&lt;/sup&gt;</td>
<td>Single</td>
<td>Fan-beam</td>
<td>Stationary ring</td>
<td>no</td>
<td>X-Ray Source rotates only</td>
<td>Higher efficiency than 3&lt;sup&gt;rd&lt;/sup&gt; Generation</td>
</tr>
<tr>
<td>5&lt;sup&gt;th&lt;/sup&gt;</td>
<td>multiple</td>
<td>Fan-beam</td>
<td>Stationary ring</td>
<td>no</td>
<td>No movement</td>
<td>Ultrafast for dynamic analysis (for instance in cardiac movements)</td>
</tr>
<tr>
<td>6&lt;sup&gt;th&lt;/sup&gt;</td>
<td>single</td>
<td>Fan-beam</td>
<td>many</td>
<td>yes</td>
<td>3&lt;sup&gt;rd&lt;/sup&gt; Generation + bed translation</td>
<td>Faster 3D imaging</td>
</tr>
<tr>
<td>7&lt;sup&gt;th&lt;/sup&gt;</td>
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<td>Narrow cone-beam</td>
<td>Multiple arrays</td>
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<td>3&lt;sup&gt;rd&lt;/sup&gt; Generation + bed translation</td>
<td>Faster 3D imaging</td>
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<tr>
<td>8&lt;sup&gt;th&lt;/sup&gt;</td>
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<td>Wide cone-beam</td>
<td>Flat Panel detector (FDP)</td>
<td>no</td>
<td>3&lt;sup&gt;rd&lt;/sup&gt; Generation</td>
<td>Large 3D</td>
</tr>
</tbody>
</table>
CT Detector module based on integrated CMOS photodiodes

Matrix Array Detector
"Lightspeed" ceramic scintillator

Adaptive Array Detector
"Ultrafast" ceramic scintillator (UFC)

Slice definition by near-focus collimators and electronic switching

Coincidence Processing Unit
Sinogram/Listmode Data

Annihilation
Image Reconstruction

(Helical CT)
SNR is dependent on dose, as in X-ray. Notice how images become grainier and our ability to see small objects decreases as dose decreases.
CT in Agriculture

• In 1982 and 1983 studies were conducted respectively by Petrovic & Hainsworth, and Aylmore. They demonstrated the possibility of using a computerized X-ray tomograph to measure the bulk density of soils;

• In 1986, Crestana and collaborators presented results related to the applications of CT in agriculture for:

1. Detection of soil heterogeneities;
2. Compaction of soils resulting from the use of agricultural machinery;
3. Dynamic three-dimensional simulation of drip irrigation in a soil column;
4. Studies on seed germination in situ.

Their results were compared with usual techniques in the agricultural area the use of CT showed greater precision.

CT in Agriculture
CT in Agriculture
CT in Agriculture

Model and Algorithm Conception

Diagram with the high level model of work processes.
Filtering Prior to Image Reconstruction

One of the main problems in CT measurement is the improvement of the Signal/Noise ratio of the collected projections and the reconstructed image. This involves:

1. Poisson noise;
2. Noise from electronics;
3. Table vibrations;
4. Noise of reconstruction and visualization algorithms.
Regarding Poisson noise, the possible solutions are:

1. Increase the exposure time to radiation, to improve the signal-to-noise ratio;
2. Applying filtering to reduce Poisson noise, working on the projections, or a posteriori in the reconstructed image.
Why to use Anscombe Transform?

• Poisson noise is characterized by being signal dependent;

• By means of the Anscombe (AT) Transform the Poisson noise is transformed into one that is approximately Gaussian, additive, with zero mean and unit variance [Anscombe, 1948] [Mascarenhas et al, 1999];

• Enables the use of noise reduction methods with stationary Gaussian distribution.
For the random variable $x$ (Poisson distribution) its AT will be defined as:

$$y_i = 2\sqrt{x_i + \frac{3}{8}} \iff y_i = 2\sqrt{x_i + \frac{1}{8}} + v_i$$

Approximately independent
In the 1940s, Norbert Wiener pioneered a filter that would produce an optimal estimate of a noisy signal [Wiener, 1949]:

\[ x(n) = d(n) + v(n) \]

Minimize the mean square error of the estimation of \( d(n) \)
Filtering Prior to Image Reconstruction (using Wiener-FIR)

• For a FIR filtering, one may have the systems of Wiener-Hopf equations given by:

\[
\begin{bmatrix}
    r_x(0) & r_x(1) & \cdots & r_x(p-1) \\
    r_x(1) & r_x(0) & \cdots & r_x(p-2) \\
    \vdots & \vdots & \ddots & \vdots \\
    r_x(p-1) & r_x(p-2) & \cdots & r_x(0)
\end{bmatrix}
\begin{bmatrix}
    w(0) \\
    w(1) \\
    \vdots \\
    w(p-1)
\end{bmatrix}
= 
\begin{bmatrix}
    r_{dx}(0) \\
    r_{dx}(1) \\
    \vdots \\
    r_{dx}(p-1)
\end{bmatrix}
\]

Autocorrelation of the signal
Weights for the FIR filter
Cross-correlation between the desired signal \( d(n) \) and the input \( x(n) \)
Wiener Filtering by Linear Prediction

• From observations without noise it is sought to estimate the value of $\hat{x}(n+1)$ in terms of a linear combination of $p$ values prior to $x(n+1)$
Re-evaluating the cross-correlation between \(d(n)\) and \(x(n)\), is possible to obtain:

\[ r_{dx}(k) = r_x(k + 1) \]

Thus, the Wiener-Hoeflt equations for the linear predictor are defined as:

\[
\begin{bmatrix}
    r_x(0) & r_x(1) & \ldots & r_x(p-1) \\
    r_x(1) & r_x(0) & \ldots & r_x(p-2) \\
    \vdots & \vdots & \ddots & \vdots \\
    r_x(p-1) & r_x(p-2) & \ldots & r_x(0)
\end{bmatrix}
\begin{bmatrix}
    w(0) \\
    w(1) \\
    \vdots \\
    w(p-1)
\end{bmatrix}
= 
\begin{bmatrix}
    r_x(1) \\
    r_x(2) \\
    \vdots \\
    r_x(p)
\end{bmatrix}
\]
3D Reconstruction for Agricultural Soil Samples

- Tomographic data are acquired without displacement of the sample under analysis. The position during the scanning process remains the same;

- Completion of the intervals between acquisition plans (A) and virtual plans (V);

- This feature allows to use interpolation and to increase resolution of 3D objects, i.e., called as Volumetric Reconstruction.
3D Reconstruction of Agricultural Samples - B-Spline-Wavelet Interpolation

- Function for interpolation

\[ f(u) = \sum_{i=0}^{N} a_i B(Nu - i) \]

Where \( u \) represents the step in interpolation and \( N \) is the number of known points.

Blending Function

\[ B(x) = \begin{cases} 
\frac{1}{6} (2 + x)^3 & -2 < x \leq -1 \\
\frac{1}{6} (4 - 6x^2 - 2x^3) & -1 < x \leq 0 \\
\frac{1}{6} (4 - 6x^2 + 2x^3) & 0 < x \leq 1 \\
\frac{1}{6} (2 - x)^3 & 1 < x < 2 \\
0 & 2 \leq |x|
\end{cases} \]
3D CT Image Reconstruction (interpolation procedures)
Parallel 2D reconstruction algorithm used in the proposed model.

Agglomeration of the parallel three-dimensional reconstruction tasks.
Modeling the parallel algorithm for 2D and 3D reconstruction - Mapping in DSP platform

– Interface between PC and DSP modules
– Interconnection of a network of modules
Filtering with Wiener

- Assays performed with additive Gaussian noise and evaluation based on the resulting variance before and after the filtration;
- In the study, a homogeneous phantom and a heterogeneous phantom were used;
Fig. 11. Typical reconstructed CT images for the analyzed soil samples, which are the dystrophic clay Dark-Red Latosol. (a) Noisy image; (b) Filtered image with the Wiener filter based on the linear prediction with 6 weights.
User Interface Image Reconstruction

Scanning Parameters

Data about the Linear Attenuation Coefficient and Color Threshold selection
User Interface for 3D Image Visualization

Navigate between the Folders to select the slices

Choice of the slices in the range of interesting

Opening the slices for visualization
Agricultural Soil Sample Analysis

- To evaluate the potential of the 3D visualization model a study of a soil sample was carried out. These results demonstrate the potentials of the model as an analysis tool for application in research related to soil science;

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Total translation</td>
<td>15.000 mm</td>
</tr>
<tr>
<td>Linear step</td>
<td>0.083 mm</td>
</tr>
<tr>
<td>Angular step</td>
<td>1.000°</td>
</tr>
<tr>
<td>Window in time</td>
<td>4 seconds</td>
</tr>
<tr>
<td>Energy</td>
<td>58.5 keV</td>
</tr>
</tbody>
</table>

Depth of 88 mm
Depth of 158 mm
Agricultural Soil Sample Analysis

\[ \mu(44,119,53) = 0.261 \text{ cm}^{-1} \]

\[ \mu(88,119,27) = 1.044 \text{ cm}^{-1} \]

\[ \mu(87,119,55) = 0.938 \text{ cm}^{-1} \]
Soil Porosity Analysis
CT in Agriculture

SKYSCAN 1172: HIGH RESOLUTION DESK-TOP MICRO-CT

- DESCRIPTION

  - Fully distortion corrected 11Mp X-ray camera.
  - Up to 8000x8000 pixels in every slice.
  - Down to 0.5μm isotropic detail detectability.
  - Dynamically variable acquisition geometry for shortest scan at any magnification.
  - World's fastest hierarchical reconstruction (InstaRecon®) and GPU-accelerated FDK reconstruction.
  - Software for 2D/ 3D image analysis and realistic visualization by surface and volume rendering.
Acknowledgements
Thank You!