Optical Tomography: The Survey on Optical Tomographic Techniques

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Abstract - Tomography technique is displaying a representation of a cross section through a human body or other solid object using x-rays or ultrasound. Whereas, it is a form of computed tomography that creates a digital volumetric model of an object by reconstructing images made from light transmitted and scattered through an object. It is widely used in medical imaging research. In our paper, we present the different kinds of optical tomography techniques to show their procedure for reconstruction, advantages with applications and followed by their limitations, to make a comparison between each of them. The techniques which we have considered here are Atom Probe tomography, confocal microscopy tomography (Laser scanning confocal microscopy), magnetic resonance imaging or nuclear magnetic resonance tomography, Single photon emission computed tomography, Seismic tomography and X-ray computed tomography.

Keywords- Atom Probe Tomography, Confocal Laser Scanning Microscopy, Magnetic Resonance Imaging or Nuclear Magnetic Resonance Tomography, Diffusion MRI, Real-Time MRI, Single-Photon Emission Computed Tomography.

I. INTRODUCTION

Tomography is taking the images by sections or we sectioning, through the use of any kind of penetrating wave. The method is used in radiology, archaeology, biology, atmospheric science, geophysics, oceanography, plasma physics, materials science, astrophysics, quantum information, and other areas of science. The word tomography is derived from Ancient Greek, "slice, section" and “to write”. A device used in tomography is called a tomogram, while the image produced is a tomogram. In many cases, the production of these images is based on the mathematical procedure tomographic reconstruction, such as X-ray computed tomography technically being produced from multiple projectional radiographs. Many different reconstruction algorithms exist. Most algorithms fall into one of two categories: filtered back projection (FBP) and iterative reconstruction (IR). These procedures give inexact results: they represent a compromise between accuracy and computation time required. FBP demands fewer computational resources, while IR generally produces fewer artifacts (errors in the reconstruction) at a higher computing cost. Although MRI and ultrasound are transmission methods, they typically do not require movement of the transmitter to acquire data from different directions. In MRI, both projections and higher spatial harmonics are sampled by applying spatially-varying magnetic fields; no moving parts are necessary to generate an image. On the other hand, since ultrasound uses time-of-flight to spatially encode the received signal, it is not strictly a tomographic method and does not require multiple acquisitions at all.

II. LITERATURE REVIEW

A. Atom probe tomography (APT)

Field evaporation is an effect that can occur when an atom bonded at the surface of a material is in the presence of a sufficiently high and appropriately directed electric field, where the electric field is the differential of electric potential (voltage) with respect to distance. Once this condition is met, it is sufficient that local bonding at the specimen surface is capable of being overcome by the field, allowing for evaporation of an atom from the surface to which it is otherwise bonded. Ion flight, whether evaporated from the material itself, or ionised from the gas, the ions that are evaporated are accelerated by electrostatic force, acquiring most of their energy within a few tip–radii of the sample. Subsequently, the accelerative force on any given ion is controlled by the electrostatic equation, where \( n \) is the ionisation state of the ion, and \( e \) is the fundamental electric charge.

\[
F = ne\nabla \phi
\]

This can be equated with the mass of the ion, \( m \), via Newton’s law (\( F = ma \)):

\[
ma = q\nabla \phi
\]

\[
a = (q/m)\nabla \phi
\]

Relativistic effects in the ion flight are usually ignored, as realisable ion speeds are only a very small fraction of the speed of light. Assuming that the ion is accelerated during a very short interval, the ion can be assumed to be travelling at constant velocity. As the ion will travel from the tip at voltage \( V_i \) to some nominal ground potential, the speed at which the ion is travelling can be estimated by the energy transferred into the ion during (or near) ionisation. Therefore, the ion speed can be computed with the following equation, which relates kinetic energy to energy gain due to the electric field, the negative arising from the loss of electrons forming a net positive charge.

\[
E = (1/2)mU_{ion}^2 = -neV_i
\]
Where \( U \) is the ion velocity. Solving for \( U \), the following relation is found:

\[
U = \sqrt{\frac{2eV_f}{m}}
\]

Let’s say that for at a certain ionization voltage, a singly charged hydrogen ion acquires a resulting velocity of \( X \) \( \text{ms}^{-1} \). A singly charged deuterium ion under the sample conditions would have acquired roughly \( X/1.41 \) \( \text{ms}^{-1} \). If a detector was placed at a distance of 1 m, the ion flight times would be \( 1/X \) and \( 1.41/X \) s. Thus, the time of the ion arrival can be used to infer the ion type itself, if the evaporation time is known. From the above equation, it can be re-arranged to show that,

\[
\frac{m}{n} = \frac{-2eV_f}{U^2}
\]
given a known flight distance, \( F \), for the ion, and a known flight time, \( t \),

\[
\frac{m}{n} = -2eV_f \left( \frac{1}{t} \right)^2
\]

and thus one can substitute these values to obtain the mass-to-charge for the ion.

Thus for an ion which traverses a 1 m flight path, across a time of 2000 ns, given an initial accelerating voltage of 5000 V (\( V \) in Si units is \( \text{kg.m}^2\text{s}^{-2}\text{A}^{-1} \)) and noting that one amu is \( 1 \times 10^{-27} \) kg, the mass-to-charge ratio (more correctly the mass-to-ionisation value ratio) becomes \(~3.86 \text{amu/charge}\). The number of electrons removed, and thus net positive charge on the ion is not known directly, but can be inferred from the histogram (spectrum) of observed ions.

**B. Confocal Laser Scanning Microscopy (CLSM)**

Confocal microscopy, most frequently confocal laser scanning microscopy (CLSM), is an optical imaging technique for increasing optical resolution and contrast of a micrograph by means of adding a spatial pinhole placed at the confocal plane of the lens to eliminate out-of-focus light. Resolution enhancement can be done as follows, CLSM is a scanning imaging technique in which the resolution obtained is best explained by comparing it with another scanning technique like that of the scanning electron microscope (SEM). CLSM has the advantage of not requiring a probe to be suspended nanometers from the surface, as in an AFM or STM, for example, where the image is obtained by scanning with a fine tip over a surface. The Uses are CLSM is widely used in numerous biological science disciplines, from cell biology and genetics to microbiology and developmental biology. It is also used in quantum optics and nano-crystal imaging and spectroscopy. The Variants and enhancements can be done by improving axial resolution. The point spread function of the pinhole is an ellipsoid, several times as long as it is wide. This limits the axial resolution of the microscope.

**C. Magnetic Resonance Imaging or Nuclear Magnetic Resonance Tomography (MRI)**

The Magnetic resonance imaging is a medical imaging technique used in radiology to form pictures of the anatomy and the physiological processes of the body in both health and disease. MRI scanners use strong magnetic fields, radio waves, and field gradients to generate images of the organs in the body. MRI has a sensitivity of around 10–3 mol/L to 10−5 mol/L, which, compared to other types of imaging, can be very limiting. This problem stems from the fact that the population difference between the nuclear spin states is very small at room temperature. For example, at 1.5 teslas, a typical field strength for clinical MRI, the difference between high and low energy states is approximately 9 molecules per 2 million.

The reconstruction methods are Fundamental electromagnetism equation, In Consideration of Measurable RF Magnetic Fields and Local SAR estimation. The core equation of Electronic Properties Tomography (EPT), which directly links (Electronic properties) and Radio frequency (RF)-coil-induced magnetic fields, can be obtained by combining the Ampere’s law and Faraday’s law as,

\[
-\nabla \times H = \omega \varepsilon_0 \mu_0 \frac{\partial E}{\partial t} \times (\nabla \times H) \quad \cdots \quad (1)
\]

Where His the RF-coil-induced magnetic field strength vector in the Cartesian coordinate, \( \varepsilon_0 \) and \( \mu_0 \) the complex permittivity as a function of the electrical conductivity \( \sigma \) and the relative permittivity \( \varepsilon_r \) and \( \varepsilon_a \), and \( \mu_0 \) are the free space permittivity and permeability, respectively. Through which EPs can be computed from measurable information of \( H+1 \) and/or \( H\)−1 distributions, as discussed later.

1) By assuming the EPs distribution is locally homogeneous, (1) can be simplified and reorganized into the Helmholtz Equation as,

\[
-\Delta H^1 = \omega^2 \varepsilon_0 \mu_0 \frac{E}{\varepsilon_c} \quad \cdots \quad (2)
\]

in which absolute EP values can be directly computed via the fraction derived from measured \( B1 \) fields.

2) The spatial variation of EPs distribution is considered. Other than EP values themselves, which are taken into account only as in (2), (1) is further expanded with terms of EPs’ gradients as follows:

\[
-\Delta H^1 = \omega^2 \varepsilon_0 \mu_0 \frac{E}{\varepsilon_c} - \left( \frac{\partial H^1}{\partial x} \right) \left( \frac{\partial E}{\partial x} / \varepsilon_c \right) - \left( \frac{\partial H^1}{\partial y} \right) \left( \frac{\partial E}{\partial y} / \varepsilon_c \right) - i \left( \frac{\partial E}{\partial x} / \varepsilon_c \right) \quad \cdots \quad (3)
\]

In which the curvature of RF-coil-induced \( H_z \) is assumed negligible within coils, e.g., birdcage, transverse electromagnetic (TEM), and microstrip coils, and only the transverse components of RF magnetic fields are taken into account as in [15] and [16]. The unknown EPs information, shown as terms associated with \( \varepsilon_c \) in (3), can be calculated through solving the linear equations formed by multiple \( H_z \) measurements. In Consideration of Measurable RF Magnetic Fields, comprises of Quadrature Coil Application with Transceiver Phase Assumption and Multichannel Transmit Coil Application WithB1 Phase Deduction.
\[ \nabla^2 H_i^+ = \omega^2 \mu_0 \varepsilon_c H_i^+ + (\nabla H_i^+ )^2 \left[ i \begin{array}{ccc} 1 & 0 & 0 \\ 0 & 0 & 1 \\ 0 & 1 & 0 \end{array} \right] \left( \nabla \ln \varepsilon_c \right) \]

\[ \nabla^2 H_i^- = \omega^2 \mu_0 \varepsilon_c H_i^- + (\nabla H_i^- )^2 \left[ i \begin{array}{ccc} 1 & 0 & 0 \\ 0 & 0 & 1 \\ 0 & 1 & 0 \end{array} \right] \left( \nabla \ln \varepsilon_c \right) \]

Local SAR Estimation: \( \text{SAR} = \sigma \left( |E_x|^2 + |E_y|^2 + |E_z|^2 \right) / 2 \rho \)

Where \( E_x, E_y, \) and \( E_z \) are the Cartesian components of the RF-coil-induced electric fields, and \( \rho \) is material mass density.

**D. Single Photon Emission Computed Tomography (SPECT)**

Single-photon emission computed tomography (SPECT, or less commonly, SPET) is a nuclear medicine tomographic imaging technique using gamma rays. It is very similar to conventional nuclear medicine planar imaging using a gamma camera (that is, scintigraphy). However, it is able to provide true 3D information. The Quality control, here the overall performance of SPECT systems can be performed by quality control tools such as the Jaszczuk phantom. SPECT image reconstruction, the SPECT reconstruction problem, In SPECT, the goal of image reconstruction is to determine accurately the three-dimensional distribution of administered radiopharmaceutical in the patient. Assume an ideal situation in which the emission photons do not experience attenuation and scatter in the patient and the collimator-detector has perfect spatial resolution without blurring effects in the measured data. A naive approach to the SPECT reconstruction is to consider the projections \( p(t, \theta) \) of radiopharmaceutical activity as the simple Radon transform, which for a two-dimensional distribution \( f(x, y) \) is,

\[ p(t, \theta) = c \int_{-\infty}^{+\infty} f(x, y) dx \]

where \( t \) is the position on the projection array, \( \theta \) is a particular projection angle, and \( c \) is the gain factor that transforms radioactivity concentration to detected signals. The SPECT reconstruction problem is different from the classical problem of image reconstruction from projections because each photon is attenuated and scattered by the material between its source and the detector. When attenuation is taken into consideration, the two-dimensional attenuated Radon transform can be written as,

\[ p(t, \theta) = c \int_{-\infty}^{+\infty} f(x, y) \exp \left( [- \int_{\theta}^{+\theta} a(u, v) \, ds] \right) ds \]

where \( a(u, v) \) is the two-dimensional attenuation coefficient distribution and \( f_{(x,y)} \) is the attenuation factor for photons that originate from \( (x,y) \), travel along the direction perpendicular to the detector array, and are detected by the collimator-detector.

However, in realistic situations, the projection data in SPECT are often measured in two dimensions and are affected by the three-dimensional effects of attenuation and scatter of photons in the patient's body and the three-dimensional spatial resolution response of the collimator-detector. When these three-dimensional effects are taken into consideration, the measured projection data are given by a more complicated attenuated Radon transform,

\[ p(t, \theta) = c \int_{t}^{+t} h(s, \omega, r) \left( \int_{\theta}^{+\theta} a(u, v) \, ds \right) ds \, da \]

where \( t = (x,y) \) is a point on the two-dimensional projection image \( p(t, \theta) \) at viewing angle \( \theta \), and \( f(r) \) and \( a(u) \) are the three-dimensional radioactivity and attenuation distributions in the patient, respectively. At each point on the projection image, unattenuated and scattered photons that fall within the field of view of the collimator-detector are detected. The inner integral of equation 5.3 considers attenuation of these photons through the patient with a three-dimensional attenuation distribution \( a(u) \). The combined geometric collimator-detector and scatter response are represented by a three-dimensional response function, \( h(s, \omega, r) \), which is a function of the position \( r \) at which the photon originates in the patient and its distance from the collimator-detector. The function is non-zero within a solid angle defined by \( \Omega \).

The SPECT Image reconstruction methods uses iterative reconstruction algorithm. The compensation methods and three dimensional reconstruction methods for special collimator designs is used.

**E. Seismic Tomography (ST)**

For solving the forward problem, in travel time tomography, solution of a forward problem is required in order to determine the ray path and the travel time residual which is to be minimized by inversion. Given a velocity structure and source and receiver locations, the problem usually is to determine the minimum travel time between the two endpoints. This is because first-arrival times are most reliably picked from seismic records since they are not contaminated by signal-generated noise that can complicate the identification and picking of later-arriving phases. The shooting and bending methods of ray tracing have been popular in the past, but the more recently introduced finite difference methods and network theory techniques are beginning to become popular. In Shooting Method, Ray tracing is a boundary value problem because the two endpoints (source and receiver) are known. The shooting method involves keeping one endpoint fixed and "shooting" the ray out in a specified direction. This initial propagation direction is then modified iteratively until the ray emerges at the target (figure 1). Julian and Gubbins (1977) define a system of six first-order differential equations:

\[ r' = v' \sigma \]

\[ \sigma = - \frac{\nabla v}{v} \]
Which can be integrated numerically to find the ray path. \( r \) is the position vector and \( o \) is the slowness where time \( t \) is the only independent variable, \( v \) is the wave speed. The starting direction which results in the ray passing through the desired endpoint may be determined by solving two nonlinear simultaneous equations:

\[
\begin{align*}
h(i_0,j_0) &= H \\
g(i_0,j_0) &= G
\end{align*}
\]

\( h \) and \( g \) are the calculated coordinates of the end of the ray (example: latitude and longitude) with starting incidence angle \( i_0 \) and starting azimuth \( j_0 \). \( H \) and \( G \) are the coordinates of the end of the correct ray.

**Figure 1: Principle of shooting method**

**F. X-Ray Computed Tomography (CT)**

Fundamental Principles of X-ray Computed Tomography (CT): Tomographic imaging consists of directing X-rays at an object from multiple orientations and measuring the decrease in intensity along a series of linear paths. This decrease is characterized by Beer's Law, which describes intensity reduction as a function of X-ray energy, path length, and material linear attenuation coefficient. A specialized algorithm is then used to reconstruct the distribution of X-ray attenuation in the volume being imaged.

The simplest form of Beer's Law for a monochromatic X-ray beam through a homogeneous material is:

\[
I = I_0 \exp(-\mu x)
\]

Where \( I_0 \) and \( I \) are the initial and final X-ray intensity, \( \mu \) is the material's linear attenuation coefficient (unit 1/length) and \( x \) is the length of the X-ray path. If there are multiple materials, the equation becomes:

\[
I = I_0 \exp\left(\sum_i (-\mu_i x_i)\right)
\]

Where each increment \( i \) reflects a single material with attenuation coefficient \( \mu_i \) with linear extent \( x_i \). In a well-calibrated system using a monochromatic X-ray source (i.e. synchrotron or gamma-ray emitter) this equation can be solved directly.

If a polychromatic X-ray source is used, to take into account the fact that the attenuation coefficient is a strong function of X-ray energy, the complete solution would require solving the equation over the range of the X-ray energy \( E \) spectrum utilized:

\[
I = \int I_0(E) \exp\left(\sum_i (-\mu_i(E) x_i)\right) dE
\]

However, such a calculation is usually problematic, as most reconstruction strategies solve for a single \( \mu \) value at each spatial position. In such cases, \( \mu \) is taken as an effective linear attenuation coefficient, rather than an absolute. This complicates absolute calibration, as effective attenuation is a function of both the X-ray spectrum and the properties of the scan object. It also leads to beam-hardening artifacts: changes in image gray levels caused by preferential attenuation of low-energy X-rays.

Hierarchy of the different types of optical tomography is shown in figure 1:

<table>
<thead>
<tr>
<th>Tomographic Techniques</th>
<th>Reconstruction Methods</th>
<th>Applications</th>
<th>Limitations</th>
</tr>
</thead>
<tbody>
<tr>
<td>Atom probe tomography</td>
<td>Ionization process</td>
<td>Metallurgy, Semiconductors.</td>
<td>Specimen geometry uncontrolled, Ion overlap, Contingent Results.</td>
</tr>
<tr>
<td>Laser scanning confocal microscopy</td>
<td>optical sectioning</td>
<td>Biology and medicine, Optics and crystallography.</td>
<td>Limits the axial resolution, Achieve resolution below the diffraction limit.</td>
</tr>
</tbody>
</table>

The critical review of the discussed tomographic techniques are shown in table 1:
III. ANALYSIS OF OPTICAL TOMOGRAPHIC TECHNIQUES

The analysis states that, the above reconstruction techniques which are forwarded are having their respective reconstruction methods which are unique to their data, which are to be computed, processed and reconstructed to give the results as the outputs. The processing capability differ from one technique to the other. Also there are numerous advantages along with the different applications and limitations. Each and every technique is best to their own approach towards reconstructing the images, which is depending on the data to be taken as an input. When keenly observed to the techniques based on the minor to major points, we can find in the analysis that they vary in minor from each other. Keeping this as the major point in our paper, we compared these reconstruction techniques and came to a conclusion with a graph. Which shows not only the performance but also the advantages as well as the limitations with respect to all these reconstruction techniques.

IV. CONCLUSION

Basics of these tomography techniques have been briefly reviewed. It is a multi-disciplinary area involving physics, mathematics, computational sciences etc. All these reconstructions come under the computational sciences. The input raw data for reconstruction depends on that particular technique used for reconstructing that particular image. Various modern reconstruction imaging techniques are discussed along with the procedure, advantages, performance, applications and limitations. The avenue of work would be to investigate the applications of the proposed methods under different and less standard types of illumination.
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