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Advances in X-ray detectors for clinical and preclinical Computed Tomography

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A R T I C L E I N F O

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A B S T R A C T

Computed tomography (CT) is a non-invasive X-ray diagnostic technique that allows reconstructing cross sections of a patient's body, providing detailed information about structure and anatomy of organs and, in some extent, also about their functionality. Since the development of the first CT scanner for clinical use in the '70s, several improvements especially in solid-state X-ray detector technology with growing detection efficiency and fast response have led to the current configuration of modern ultra-fast, low dose whole body CT scanners. Such developments brought great advantages in the clinical settings in terms of image quality, dose effectiveness, imaging throughput, but also extending considerably the field of clinical application that were initially foreseen. Parallel to the roadmap of clinical CT technology, dedicated systems for high-resolution preclinical CT (or micro-CT) have seen a considerable growth in the last two decades, taking advantage of the modern technology of high granularity flat-panel X-ray detectors (FPD). This article aims at reviewing the milestones of the evolution of X-ray detector technology that have traced the roadmap of development of CT and micro-CT. An outlook of the current and future trends on energy resolved clinical and preclinical CT with photon counting detectors will be also given.

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1. Introduction

Since its invention in the '70s by Sir Godfrey N. Hounsfield and Allan M. Cormack (Nobel laureates in Physiology or Medicine in 1979), Computed Tomography (CT) has had almost immediately an enormous impact in the medical field [1], although it has also found applications in many other fields, from non-destructive testing (NDT) in industrial processes [2,3], cultural heritage [4], to geophysics [5] and homeland security [6,7]. The mathematical foundations of image reconstruction from projections came after the work of the Austrian mathematician J. Radon [8], even though many years had to pass by his seminal work before the advent of modern computers to see a practical application of his studies. Since the construction of the first commercial head-CT scanner EMI Mark I by Hounsfield at EMI [9], a constant race for technological improvement from manufactures and researchers worldwide have allowed the development of increasingly faster scanners, providing images with finer details and wider fields of view. The first whole-body CT scanner (ACTA – Automatic Computerized Tomography Axial) was realized soon after by Ledley et al. [10] at Georgetown University, USA. In the late '80s, the introduction of slip-rings and continuously rotating scanners added a major step forward in CT technology. Soon after, Kalender et al. have introduced the mathematical basis of helical (or "spiral") CT reconstruction [11], which combined to the development of multi-row detectors progressively led CT to the today's era of ultrafast, low-dose whole-body imaging.

Parallel to the development in the clinical arena, a consistent effort in CT development was devoted to high-resolution imaging of small samples. Elliot and Dover firstly described in 1983 a microscopy system based on the concept of X-ray CT (now commonly known as micro-CT, or X-ray microtomography), reaching a resolution of 15 μm with a field of view limited to 0.5 mm [12]. Due to the pivotal role of animal models of human diseases in understanding the underlying mechanism of pathologies, as well as assessing the effectiveness of new drugs and therapeutic approaches [13], a wide range of high-resolution CT systems dedicated for the study of small rodents, mainly mice and rats, have been built and validated in the last two decades [14–16]. Just like "macro" CT, also micro-CT scanners have found a plenty of applications outside the *in vivo* imaging, such as fossil analysis [17], NDT [18], material analysis and tissue engineering [19]. These scanners feature spatial resolutions of the order of 1–10 μm [20] or even sub-micron [21].

In this paper, the main milestones in the advances of CT technology will be reviewed, with particular emphasis on the evolution of X-ray detection that has led to the current configuration and performance of clinical and preclinical CT scanners. After discussing the pros and cons of detecting materials (mainly high-pressure Xenon and solid-state scintillators or semiconductors), considerations of geometric arrangements of the detector arrays

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will be done, also in relationship with their impact on performance (i.e., image quality vs. speed and dose). The current status of photon counting CT and energy-resolved imaging (now sometimes called "color CT") and the factors that are slowing down its definitive introduction in the clinical market will be also discussed.

2. Evolution of CT detectors

2.1. Basic principles of CT

The basic principles of CT can be found in an extensive body of review articles and textbooks [22–25]. In this context, we just summarize the main factors involved in the CT image formation in order to better understand the key role of X-ray detectors. The goal in CT acquisition is to collect the line integrals, p, of the three-dimensional distribution of the object's attenuation coefficient, $\mu(x,y,z,E,t)$, through all the possible lines l crossing the object

$$p_l(E,t) = \int_l \mu(x,y,z,E,t)dl \quad (1)$$

The above notation takes into account that the attenuation coefficient μ depends on the photon energy, E, and on the time t (this must be considered especially in contrast enhanced CT protocols or when the object motion during the scan time cannot be neglected). Assuming an ideal monoenergetic distribution of the photons in the incident beam, following the Beer's law of exponential attenuation we get

$$p_l(E,t) = -\ln\frac{I(t)}{I_0} \quad (2)$$

where I(t) and I_0 are the measured X-ray intensities of the attenuated and unattenuated beams, respectively, recorded along the line l at time t. After collecting a complete set of line integrals, a reconstructed distribution of the attenuation coefficients can be obtained via several approaches (see, for instance, Ref. [26]) All reconstruction methods are based on complex computer calculations, and hence the physical measurements required to gather the line integrals must be digitized in order to be stored in a computer and then processed. Apart from some pioneering work in classical tomography such as those of the Italian radiologist Vallebona [27] and few other documented experiments, CT as we know it today relied on analog-to-digital conversion systems since its early days. Natterer and Ritman authored a good review article covering also CT historical developments [28].

2.2. From scintillators to gas detectors, and back again

The EMI Mark I head scanner consisted in a rotating-translating system in which a pair of NaI detectors were employed, where each element of the pair was devoted to the acquisition of a different slice (i.e., the EMI scanner was a two-slice tomograph) [9]. A total of 160×180 (radial $\times$ angular) readings were taken for each slice, and then processed by a minicomputer and reconstructed via Algebraic Reconstruction Technique (ART) methods. See, for instance, the classic textbook of Kak and Slaney for a detailed overview of ART [29]. A big limitation of this first generation CT was in the dynamic range of the detection system, which required a water bath surrounding the patient's head (physically separated from the head by mean of a rubber cap) to avoid saturation of the detector when the pencil beam is shifted outside the skull's margins (see Fig. 1). As reported in the original Hounsfield's paper, it took about 5 min with this system to get a single slice, which was barely sufficient for the computer "to keep the pace with the flow of patients through the scanner unit". A similar scanner design was kept for the first whole-body tomograph (ACTA) developed by Ledley in 1974 at the National Biomedical Research Foundation (NBRF) of the Georgetown University [10]. The water bath was obviously not compatible with a scanner designed to image the patient's trunk, and hence this part was replaced by a pre-patient attenuator shaped in such a way to reduce the photon fluence at the periphery of the field of view (now called bowtie filter, typically made in aluminum).

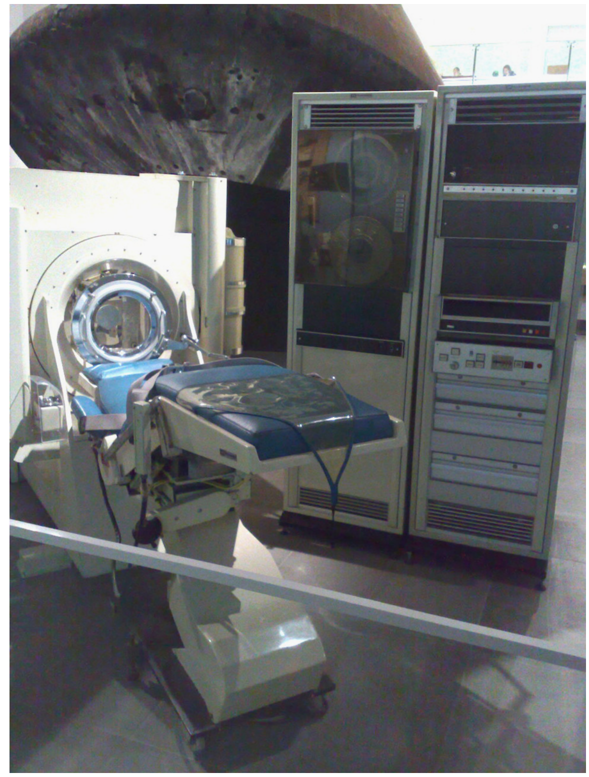

Fig. 1. The EMI Mark I scanner at the South Kensington Science Museum, London. The rubber cap used to separate the patient head from the water bag can be seen at the bore entrance. This water bag served to reduce the photon fluence at the periphery of the field of view (outside the skull's margins), thus avoiding to saturate the very low dynamic range of the NaI-based detectors.

After the introduction of the Ledley's ACTA scanner, the evolution of CT technology was marked by the pursuit of faster and faster data collection strategies by progressively increasing the number of detector elements in both transaxial (1970–1990) and then in the axial (since 1990 to date) direction. Large detector arrays had not only the advantage of shortening the scanning time and hence reducing the motion artifacts [30], but also the important advantage of increasing considerably the thermal efficiency of the X-ray tubes by enlarging the collimation width (initially only in the transverse direction). The big stability and uniformity of response required for large detector arrays, especially when using Filtered Back-projection as reconstruction method, was the main reason why early third-generation scanners started to employ pressurized xenon-gas chambers instead of scintillators coupled to photomultipliers [23]. Because the efficiency of a detection element in a multi-cell Xenon proportional chamber is mainly dependent on the gas pressure, which is intrinsically uniform throughout the entire chamber, these kinds of detectors exhibited very high uniformity thus reducing image artifacts such as ring-shaped artifacts. On the other hand, quantum detection efficiency (QDE) was low as compared to solid-state detectors. This is why since the '80s manufactures started to build CT detector arrays made of scintillating finger crystals coupled to photodiodes.

2.3. QDE, decay time and image quality

QDE is defined as the normalized integral of the energy dependent quantum efficiency, η, over the whole energy spectrum

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