Application of scintillating fiber gamma ray detectors for medical imaging

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ABSTRACT

The recently developed plastic scintillating fiber technology started the development of a new generation of high spatial and time resolution gamma ray detectors for medical imaging, such as positron emission tomagraphy (PET) and single photon emission computed tomography (SPECT). A scintillating fiber PET module consisting of two 5 x 5 x 2.5 cm^3 detector stacks made of parallel 1.0 mm diameter fiber, separated by 20 cm, each viewed by a Hamamatsu R2486 position sensitive photomultiplier was developed and tested. The time resolution of the coincidence system is 10 nsec. The spatial resolution and efficiency of this module turned out to be 2.3 mm (FWHM) and 2.0%, respectively, and independent of the location of the ^{22}Na testing source inside a sphere of 2 cm radius around the center of the two fiber stacks. The effect of gammas scattered in a 15 cm diameter water filled glass cylinder into which the ^{22}Na was immersed did not change the spatial resolution of the system.

1. INTRODUCTION

The recently developed plastic scintillating fiber technology[1] started the development of a new generation of gamma ray detectors for medical imaging, such as positron emission tomography (PET) and single photon emission computed tomography (SPECT). Both PET and SPECT are nuclear medical imaging methods which are used to measure the spatial and temporal concentration of radioactivity in a volume element of tissue in a living system.

The principle of operation of PET is the fact that positron emitting radionuclides give rise in their close vicinity to positron-electron annihilation processes resulting in the emission of two gamma rays of 0.511 MeV energy in opposite directions. This allows these two gammas to be detected in coincidence by gamma ray detectors placed on opposite sides of the patient's body or experimental object. This greatly simplifies the image reconstruction since one knows that the source is on a line between the two points defined by the detected photoelectric absorptions or Compton events produced by the gammas in the opposite detectors. The accurate time resolution of the two events allows one to use narrow coincidence windows to eliminate much of the background gamma radiation, and an accurate spatial resolution results in a powerful imaging method of diagnostic radiology.

SPECT is based on the detection and spatial mapping of gamma radiation emitted by radiopharmaceuticals labeled with specific gamma ray emitting radionuclides. A scintillation detector, equipped with an appropriate lead collimator to determine the gamma emission direction, maps the gamma ray intensity to produce a projectional image.

PET scanners are using mainly ^{11}C, ^{13}N, ^{15}O and ^{18}F positron emitting radionuclides, and the major factor limiting their spatial resolution is the distance traveled by the positron from the point of emission to the point of annihilation which is of the order of 1 mm. Another factor deteriorating the spatial resolution is the fact that the annihilation gammas are not emitted at precisely 180 degrees. The combined effect of these two factors is limiting the spatial resolution for ^{18}F, having the smallest positron energy (E_{max} = 0.63 MeV) thus the shortest positron range, to abut 2 mm in FWHM. An additional smearing of the image is caused by the scattering of gammas in the body before being detected. This effect can be significantly decreased, however, by measuring the energy of gammas, and rejecting events with less than a cut-off energy

properly chosen below 0.511 MeV. The recently developed and commercially available PET scanners have spatial resolutions of about 5-6 mm.

The spatial resolution of SPECT scanners using gamma emitting radionuclides, such as ^{99m}Tc, ^{201}Tl, and ^{123}I, is in general 5-10 mm (FWHM). But because the gamma rays are emitted directly by the radionuclides, their resolution is not limited by the first two physical factors discussed for PET systems, and the image is smeared out only because of the spatial resolution of the detector, and the scattering of the gammas in the human body. Thus with improved detectors the achievement of submillimeter resolution is possible. This will be achieved by applying very thin fibers and solid state photomultipliers (visible light photon counters: VLPC's) under development at Rockwell International Research Center, Anaheim, California.

PET has the advantage of using physiologically important radionuclides, thus making possible to synthesize biochemically important radiopharmaceuticals. On the other hand the radionuclides most frequently used in PET have short half-lives ($t_{1/2}$ = 20 min, 10 min, 2 min, and 110 min, for ^{11}C, ^{13}N, ^{15}O and ^{18}F, respectively) requiring expensive on-site accelerators for production, and complex radiochemistry laboratories for the synthesis of certain radiopharmaceuticals.

SPECT has the practical advantage of using readily available radionuclides that are routinely employed in hospitals and nuclear medical centers. The half-lifes of these isotopes are sufficiently long that they can be produced in nuclear reactors or accelerator facilities and shipped to the hospitals or medical diagnostic centers.

2. APPLICATION OF SCINTILLATING FIBER DETECTORS FOR PET SCANNERS

Recent advances in the improvement of the spatial resolution, time resolution and scanner sensitivity of PET and SPECT devices made them possibly the most powerful methods of non-invasive cardiological and neurological diagnosis, and unique tools for investigation of changes in human physiology, biochemistry and cellular functions[2]. The PET technique was also used for the quantitative assessment of fluorine in bone[3], and SPECT was shown to have an increasing role in the medical imaging of cancer.[4]

Previous studies carried out by us [5-8] showed that the highest spatial resolution, time resolution and a relatively high scanner sensitivity can be achieved by applying plastic scintillating fibers coupled to position sensitive photomultipliers, a new detector technology recently developed for high-energy accelerator physics such as the Superconducting Super Collider experiments, and space-based gamma ray astronomy.

In a previous paper[6] we discussed the Monte Carlo modeling of a PET detector constructed of scintillating fibers of 0.5 x 0.5 mm^2 cross section. The simulations showed that a detector of this sort would have a spatial resolution better than the positron smearing and the angular deviation inherent to PET technology. In our most recent paper[7] we described the construction and testing of scintillating fiber PET camera modules using different fiber diameters and fiber stack configurations. The PET scanners are usually consisting of a circular array of detectors (PET camera). Constructing two small opposite sections and use it as a module of this curricular camera, one can test the spatial resolution and efficiency of the PET system.

The best results have been achieved by the following PET module:

Two 5 x 5 x 2.5 cm^3 detector stacks made of parallel 1.0 mm diameter scintillating fibers (Fig. 1), each viewed by a Hamamatsu R2486 position sensitive photomultiplier. The two stacks were separated by 20 cm and connected in coincidence forming a PET module (Fig. 2).

The scintillating fibers are made of polystyrene scintillator (refractive index n = 1.59) doped with Butyl-PBD and d-POPOP producing λ =420 nm wave length photons, with a low refractive index (n=1.41) cladding of 30 μm thickness. The electronic readout system is the same as used in our previous experiments[5]. The time resolution of the coincidence system is 10 nsec. The fibers were coated with a very thin light absorbing paint to avoid cross talk between the fibers.

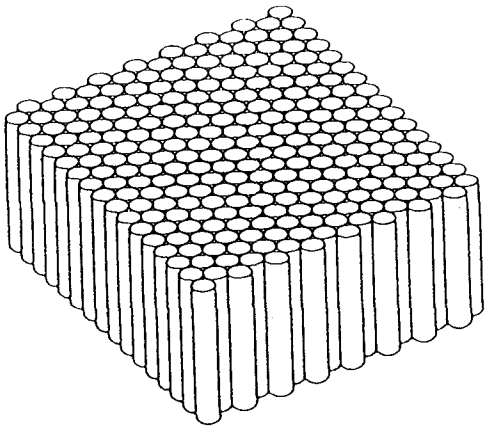

Fig. 1 Schematic view of the 5 x 5 x 2.5 cm^3 fiber stack made of parallel scintillating fibers

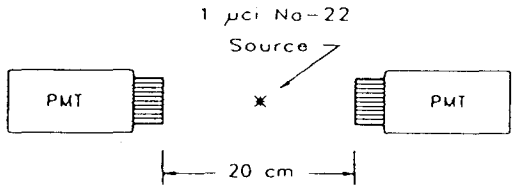

Fig. 2 Schematic view of the PET module made of two parallel scintillating fiber stacks

An approximately 1mci ^{22}Na source of 0.5 mm diameter was positioned in the center of the two stacks (Fig.2). This PET module permits the 2-dimensional reconstruction of the image. A 3-dimensional image reconstruction of an abut 5 cm deep section of an object can be achieved with a set of identical modules forming a circular camera. A deeper section of the object can be imaged either by moving it relatively to the circular PET camera, or constructing several adjacent circles of PET modules leading to a larger depth of the PET camera. The long living ^{22}Na radioisotope ($t_{1/2} \sim 2.6$ year) was selected for this test because it emits positrons of similar energy spectrum as ^{18}F (E_{max} = 0.55 and 0.63 Mev, for ^{22}Na and ^{18}F, respectively).

The 2-dimensional spatial resolution recently obtained using the above described PET module is 2.3 mm (FWHM) (Fig. 3), and independent of the location of the ^{22}Na source inside a sphere of 2 cm radius around the center point between the two stacks. The efficiency of the PET module is 2.0% also independent of the location of the source inside the same sphere. These new results on the spatial resolution and efficiency of the system are in good agreement with our previously published values measured only in the center point between the two stacks[7].

We tested also the effect of the scattering of annihilation gammas in the human body in respect of the spatial resolution. The ^{22}Na source was immersed into the center of a 15 cm tall cylindrical glass container of 15 cm diameter and 0.3 cm wall thickness, filled with water simulating the scattering effect of human tissues. The 2-dimensional spatial resolution of this system turned out to be 2.3 mm (FWHM), the same value which was found without the water absorber, and again independent of the location of the source inside the same sphere.

The negligibly small effect of the scattering of gammas in the absorber could be explained by the fact that only a very small fraction of these gammas falls into the close vicinity of the less than 3 mm diameter circle of the image of the 0.5 mm diameter source, thus leaving the width of this peak in the coincidence count rate practically unchanged and increasing only the relative background level all over the detector surface. The small effect of scattered gammas on the spatial resolution is also due to the small acceptance angle of the PET module. Monte Carlo simulations are under way to clarify these effects, and permit the optimization of the design of larger circular PET scanners for imaging human-size objects, with particular emphasis on the improvement of the sensitivity of the system.

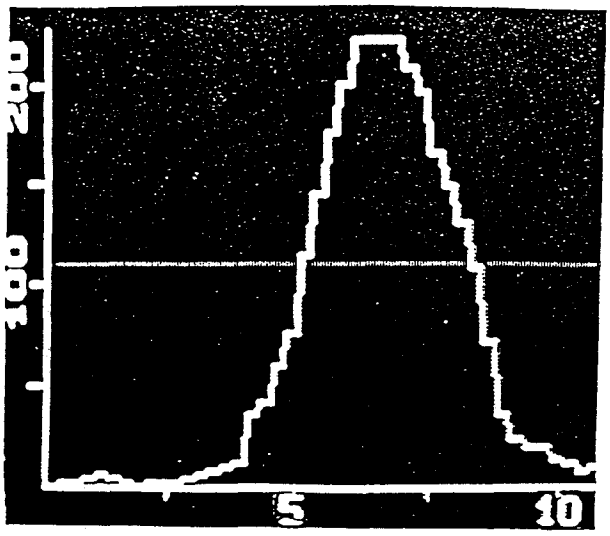

Fig. 4 Count rate vs. position (mm) along the center of the image of the 0.5 mm diameter 1 μci Na-22 source

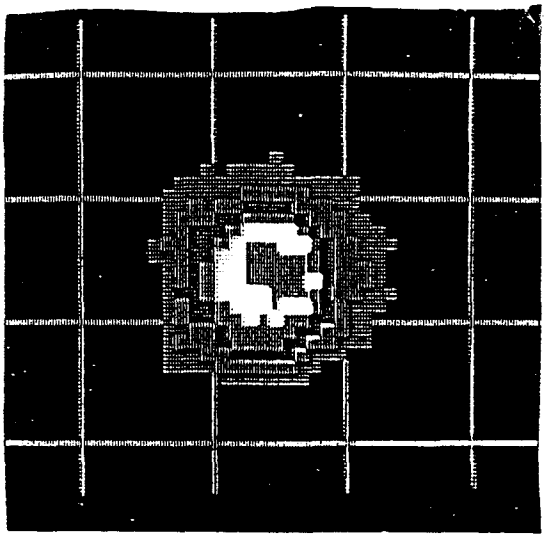

Fig. 5 Image of the 0.5 mm diameter 1 μci Na-22 source. Each square is 2.5 x 2.5 mm^2

3. REFERENCES

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