Title
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Positron Emission Mammography Imaging

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Abstract

This paper examines current trends in Positron Emission Mammography (PEM) instrumentation and the performance tradeoffs inherent in them. The most common geometry is a pair of parallel planes of detector modules. They subtend a larger solid angle around the breast than conventional PET cameras, and so have both higher efficiency and lower cost. Extensions to this geometry include encircling the breast, measuring the depth of interaction (DOI), and dual-modality imaging (PEM and x-ray mammography, as well as PEM and x-ray guided biopsy). The ultimate utility of PEM may not be decided by instrument performance, but by biological and medical factors, such as the patient to patient variation in radiotracer uptake or the as yet undetermined role of PEM in breast cancer diagnosis and treatment.

Keywords: Positron Emission Mammography; Breast Cancer; PEM Camera Design and Optimization

1. Introduction

A number of dedicated PET cameras optimized to image the breast have been proposed or constructed [1-9]. These cameras, commonly known as Positron Emission Mammography or PEM cameras, restrict the field of view to a single breast, and have higher performance and lower cost than a conventional PET camera. By placing the detectors close to the breast, the PEM geometry subtends more solid angle around the breast than a conventional PET camera. In addition, gamma rays emitted in the breast have to pass through at most one attenuation length (~10 cm) of tissue in the PEM geometry, but may have to travel through as much as four attenuation lengths of tissue in a conventional PET camera. These two factors significantly increase the sensitivity (the detected coincident event rate per unit activity in the field of view) in the PEM geometry. This paper reviews PEM camera design and the tradeoffs that affect their imaging performance, as well as non-instrumental issues that affect them.

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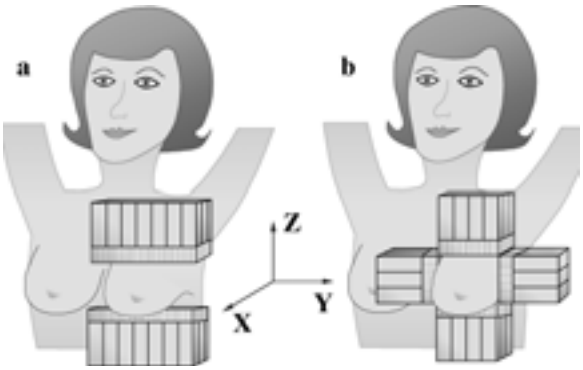

Figure 1. The most common PEM camera geometry is the parallel plane geometry shown in a). A variant rectangular geometry is shown in b).

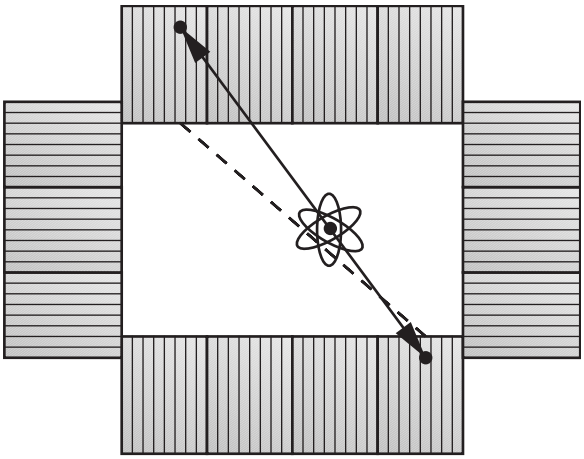

Figure 2. If the interaction position of gamma rays that penetrate into the detector module is assigned to the front face of the detector element, mis-positioning errors occur as the line connecting these points does not go through the source (the dotted line). If the interaction depth in the detector is measured, then the position is no longer assigned to the front face and the mis-positioning error is eliminated (solid line).

2. Camera Design

2.1. Geometry

The most common PEM geometry is the parallel plane geometry shown in Figure 1a. It consists of a pair of parallel plane detector heads placed above and below a single breast. The size of the plane varies, but is usually less than 8 in. square so that it can fit within an x-ray mammography unit. Smaller areas than this are common, as most of the existing PEM cameras are effectively prototypes, and so have reduced their area in order to minimize cost and complexity. The spacing between detector planes is variable, allowing mild breast compression. Compression serves several purposes. It reduces motion artifacts, it spreads out the breast (making it easier to resolve separate structures) and it thins the breast (which improves the contrast as it reduces the amount of normal tissue in the field of view).

2.2. Photomultiplier Tubes

Recent advances in scintillator array and position sensitive photomultiplier tube (PSPMT) technologies have simplified construction of PEM camera detectors. PSPMTs can economically and accurately read out arrays of optically isolated scintillator pixels. These devices generally produce four analog output signals. The sum of the four signals is proportional to the total energy deposited into the scintillator crystal array, while the ratio of provides the 2-dimensional center of gravity of the optical signal and thus the position of the crystal that the interaction took place in. While each anode of a multi-anode PMT could be coupled one-to-one to a scintillator crystal, it is more common to use an external resistor network to effectively turn them into PSPMTs.

2.3. Scintillators

The spatial resolution of the camera is largely determined by the in-plane size of the scintillator crystal. To identify 5 mm diameter tumors, high spatial resolution (~2 mm) is required, so the scintillator crystals are usually between 2 mm and 4 mm square. Thick (~3 attenuation length) scintillator crystals are desired in order to obtain high efficiency. Scintillator crystals with high density and high atomic number (such as LSO, BGO, GSO, or LuAP) are also desired, although cameras have also been constructed using NaI:Tl scintillator [4].

However, deep crystals lead to an inherent degradation because the object to be imaged is close to the detector modules. As Figure 2 shows, many gamma rays penetrate a significant distance into the detectors before they interact and are detected. If the interaction depth within a detector element is not

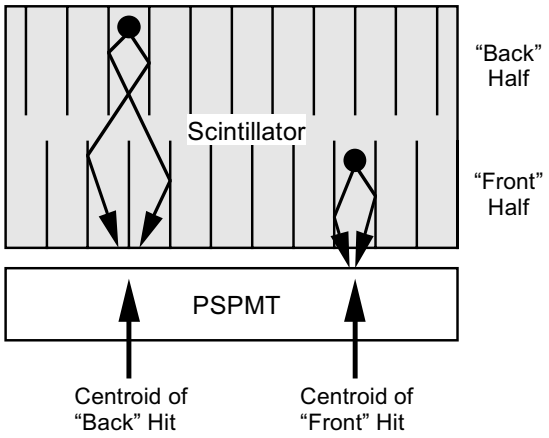

Figure 3. A method for determining the depth of interaction. Interactions that occur in the "front" section of the scintillator produce light whose centroid (at the PSPMT entrance) is at the center of the scintillator pixel. Interactions that occur in the "back" section of the scintillator block produce light whose centroid is at the edge of the scintillator pixel (*i.e.*, offset by half of a crystal width).

measured (as with conventional PET detector modules), then the interaction position is assigned to the front face of the detector element that the interaction occurs in. A line joining the two such assigned points may not pass through the actual source position, resulting in mis-positioning errors and degrading the spatial resolution.

Reducing the thickness of the detectors can reduce this distortion. However, this decreases the fraction of emitted gamma rays that interact in the detectors, reducing the single gamma ray detection efficiency. As two detections are required for a PEM event, the overall detection efficiency is the square of the single gamma ray detection efficiency, so a camera with 1 attenuation length of detector will have an efficiency that is less than half that of a camera with 3 attenuation lengths of detector.

If the depth of interaction (DOI) is measured (leaving the detector 3 attenuation lengths thick), the line joining the two measured interaction positions will pass through the actual source position, and the high detection efficiency will be maintained with no mis-positioning errors. Two different schemes have been employed to measure the DOI. The first, shown in Figure 3, determines whether the interaction occurred in the front or back half of the scintillator block [1]. Quasi-separate arrays of crystals are cut in the front and back halves of a single block of scintillator material, with the front and back arrays offset by a half crystal width. Light that is produced by interactions in the "front" half is confined to the crystal that it was produced in, and so its center of gravity (as measured at the entrance to the PSPMT) is at the center of the crystal. Light that is produced by interactions in "back" crystals is transmitted to the PSPMT by two "front" crystals, and so its center of gravity is at the boundary between two "front" crystals. Another method is to place a photodetector at each end of each scintillator crystal. The sum of the two photodetector signals measures the energy and the ratio provides the DOI measurement [10].

3. Non-Instrumentation Issues

Some of the critical limitations to PEM have nothing to do with camera design. For example, there is considerable interest in detecting small (3 mm diameter and below) tumors. However, small tumors contain low absolute amounts of activity and so may be very difficult to observe above the background activity level. Assuming a 1 mCi injection into a 70 kg patient and a 3:1 tumor to normal tissue uptake ratio (a typical value for fluoro-deoxyglucose, which is the most commonly used radiotracer for breast cancer), the expected activity concentration is 14 nCi/cc in normal tissue and 43 nCi/cc in tumors. This implies that during a 10 minute acquisition time there would be only 13,000 annihilations in a 3 mm diameter tumor, as compared to 500,000 in a 1 cm diameter tumor and 310 million annihilations from ~1000 cc of normal breast tissue. Thus, imaging small tumors will be difficult because the volume (and hence number of annihilations) scales as the cube of the tumor diameter.

In addition, there is significant patient to patient variation in the tumor activity concentration. The cause for this is not understood — a recent study searched for correlations between the tumor SUV (standard uptake value, which is effectively a measure of the tumor to normal tissue ratio) for fluoro-deoxyglucose and over a dozen different histological and pathological measures of tumor characteristics (*e.g.*, size, grade, vascularity, estrogen and progesterone receptor status, mitotic figure, etc.) and either weak or no correlation was observed with

each measure [11]. Thus, it is possible that an impeccably designed PEM camera will be unable to image a breast cancer tumor merely because the tumor, for unknown reasons, has a low tracer uptake.

Finally, the exact role of PEM in clinical diagnosis and treatment is uncertain. It is likely to be too expensive to replace x-ray mammography for routine screening, and while its diagnostic accuracy is similar to that of biopsy (PET has <10% false negative and false positive fractions for >1 cm diameter tumors, while biopsy has a false negative fraction of ~10% and a false positive fraction of 0%), a new technology must have superior (not merely comparable) performance to replace an existing technology. PEM's limited field of view also limits its utility for staging (determining how far the cancer has spread) or treatment follow-up. While the exact role is yet undetermined, I feel that there are likely to be valuable clinical uses for PEM, such as routine screening for the >10% of women for which x-ray mammography is unsuitable (*e.g.*, those with mammographically dense breasts, implants, or scars from previous biopsy or surgery) or as a non-invasive adjunct to biopsy. I further believe that it is important to develop PEM cameras and perform clinical trials with them. As with any new technology, the optimal role for PEM is not likely to be recognized until PEM studies have been performed and a body of clinical information obtained. PEM will only be accepted if it provides information that changes the course of treatment for the patient or their expected outcome.

4. Conclusions

PEM offers significantly higher sensitivity for tumors in the breast than conventional PET cameras, mainly because of significantly increased solid angle coverage and reduced attenuation in the patient. Several design features, notably increased solid angle coverage due to encircling the breast and detector modules that measure the depth of interaction, could be implemented in PEM cameras to improve their performance. Finally, there are significant limitations due to non-instrumental effects, such as the absolute amount of radiotracer that is absorbed by the tumor and the uncertain niche for PEM in clinical diagnosis and treatment.

Acknowledgments

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