A Feasibility Prototype of a Low-dose Stationary Tomographic Molecular Breast Imaging Camera Using 3D Position Sensitive CZT Detectors

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***Abstract*— A novel Molecular Breast Imaging system (MBI) is being developed by Kromek and University College London. MBI systems have been shown to provide excellent results in patients with dense breasts, but higher than mammography patient dose and long imaging time of 40 min impede their wide adoption. The design is based on a pair of opposing CZT detector arrays and high-density multi-pinhole (MPH) collimators. The new system combines superior energy and position resolution and depth of interaction (DOI) sensing of CZT detectors, wide range angular sampling of MPH collimators and novel demultiplexing image reconstruction techniques to deliver tomographic images with stationary detectors. The new system is expected to mitigate the MBI drawbacks without compromising diagnostic content. Following the initial evaluation, a new feasibility prototype using upgraded ASIC and new front-end electronics was designed, built, and characterised. New image reconstruction and de-noising algorithms were developed on simulated data to improve the Contrast-to-Noise (CNR) ratio and reduce the fluctuations in the correlated background noise associated with MPH image reconstruction. The collimator geometry was further optimised by including non-local means noise filtering which increased the CNR by almost a factor of 3. The current results indicate that we can achieve the dose reduction to the mammography level, reduce the minimum size of detected lesions to about 6 mm and have an additional potential for reduction of the measurement time.**

***Index Terms*—CZT, molecular breast imaging, multi-pinhole collimator, multiplexing, dose reduction.**

I. Introduction

MAMMOGRAPHY is widely used as a screening procedure for breast cancer, however, its sensitivity for lesion detection is limited in patients with dense breast tissue [1]. Molecular breast imaging (MBI) using a pair of planar detector arrays has been shown to have high sensitivity in cancer detection, even in patients with dense breast tissue [2]. However, long imaging time and radiation dose that is higher than mammography impede its widespread adoption in clinical practice. Several tomographic MBI systems have been proposed is the past [3-6], including rotating dual head systems with non-multiplexing multi-pinhole (MPH) collimators. A prototype of a low-dose MBI system offering tomographic imaging with stationary detectors is currently under development by Kromek. It is based on the use of dual opposing CZT detector arrays and densely packed MPH collimators leading to significant multiplexing (MX) in the acquired projections. The goal is to achieve at least the same sensitivity for lesion detection as planar MBI, but with effective patient dose similar to mammography. The new concept for a low-dose MBI camera and first imaging results from a proof-of-principle prototype were presented at IEEE MIC/RTSD in 2021 [7-9]. The performance of the prototype was evaluated further and the results along with further development of the image reconstruction algorithms provided a basis for designing and building an upgraded feasibility prototype. The new system is currently under evaluation including spectral and phantom measurements.

II. Methods

The proof-of-principle prototype was comprised of a 2x2 detector array of 7.3 mm thick CZT detectors with DOI capability and an MPH collimator with a large number of closely spaced pinholes providing necessary angular sampling without motion. The characterisation of the prototype along with the system simulations demonstrated the initial feasibility of the new principle for dose/time combination reduction. However, it was found that the DOI resolution and calibration stability required improvement. A new detector system using upgraded ASIC and new front-end electronics was used to design and build an upgraded feasibility prototype (see Fig.1). New image reconstruction and de-noising algorithms were developed to improve the CNR (Contrast-to-Noise) ratio and reduce the fluctuations of the correlated background noise associated with MPH image reconstruction.

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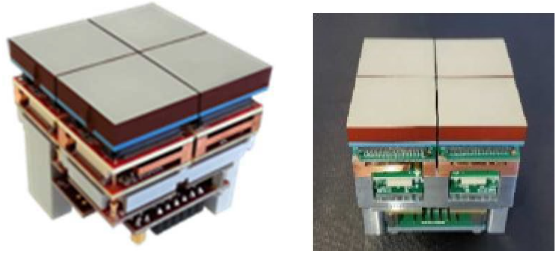

Figure 1. Detector module used in the proof-of-principle prototype (left) and the new feasibility prototype (right).

A. *Characterisation and upgrade of prototype*

The imaging performance of the proof-of-principle prototype was characterised using several “activity-painted” [10] phantoms, including a large volume uniform distribution. The position resolution was studied using a 250 µm pencil beam collimator with a 57Co source mounted on a motorised XY stage and illuminating detector from the side. Based on these results and further image reconstruction simulation studies, it was determined that the dose/time reduction quantification requires upgrading the detector system to improve the DOI precision to < 1 mm from the previously reported 1.2 mm. In addition, the cathode peak position stability should be improved to < 1% over the 24 h time scale to ensure the long-term stability of per-pixel DOI calibration.

The above requirements were fulfilled with an upgraded version of the previously used D-Matrix gamma imager technology. The new system includes a revised version of the ASIC and fully redesigned readout and FE electronics to incorporate the cathode readout within the board layout of the module and to improve the detector noise performance. The thickness on the new detectors was reduced from 7.3 mm to the more mainstream 5 mm. The change was based on the simulation study which confirmed that with our tomographic image reconstruction the improvement in CNR and lesion detection sensitivity is virtually the same within this range of thickness.

The DOI resolution was measured with the same way but with a 500 µm collimator. The spot size was estimated using GEANT4 [11] simulations by measuring the FWHM of the detected photon position distribution in the detector. The FWHM of the measured beam profile is approximated as convolution of a Gaussian and a boxcar function representing the beam profile. The DOI is calculated as by deconvolving of the two functions:

$$\int_{-a}^{a} e^{-\frac{(x-x\prime)^2}{\sigma^2}}\, dx' \sim\sigma \left(\mathrm{erf}\left(\frac{a-x}{\sigma}\right)+erf\left(\frac{a+x}{\sigma}\right)\right)$$

B. *Optimisation of collimator design and manufacturing*

The 3D printed tungsten collimator for the proof-ofprinciple prototype was found to have a high number of small defects on the collimator surface: chippings and spikes across the surface and significant irregularities at the edges of the square hole apertures, as shown in Fig.2 (left). Such defects led to significant variations in the effective size and shape of the pinholes which will have a prominent adverse effect on the reconstructed image. A set of new collimators was printedusing an alternative supplier, which provided much better printing quality as shown in Fig.2 (right).

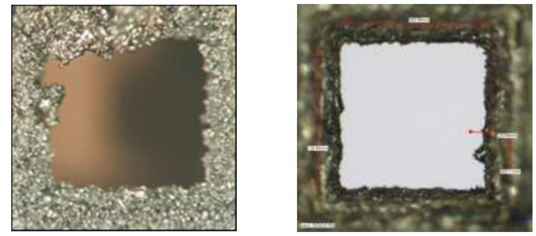

Figure 2. 3D printed tungsten collimator hole picture from the first proof-of principle prototype (left) and new feasibility prototype (right).

C. *Optimisation of image reconstruction and noise filtering*

The optimisation of the image reconstruction performance is based on a simulation model comprised of an 8x8 array of 7.3 mm thick CZT detectors. The phantom is a 6 cm thick uniform layer with two sets of 7 spherical lesions with 4-10 mm diameter as shown in Fig.3. The injected activity used in the model was 150 MBq (half of the lowest dose typically used in clinical practice [2]) with target-to-background ratio (TBR) of 10 and 2.5-min measurement time (quarter of the time currently used in clinical practice). The collimator parameters studies in the optimisation were pinhole size, opening angle, and separation distance. Non-local noise filtering and a relaxation scheme were applied to the results. More details about the optimisation process can be found in [12].

D. *Phantom measurements*

For the experimental verification of the imaging performance, a series of measurements were carried out using phantoms filled with 99mTc solution together with the prototype detector array described above. The phantoms served as representations of a single spherical lesion over a homogeneous tissue background (see Fig.4). Two phantoms were measured separately with a tumour-background ratio of 2 and 3, and then the data files were combined before reconstruction. The measurement times were set in a way that corresponds with the number of decays expected during a typical MBI patient measurement of acquisition time ranging from 24 seconds to 20 minutes. A typical reconstructed image is shown in Fig.5. Following the reconstruction, the contrast-to-noise ratio values were measured on all images. As a

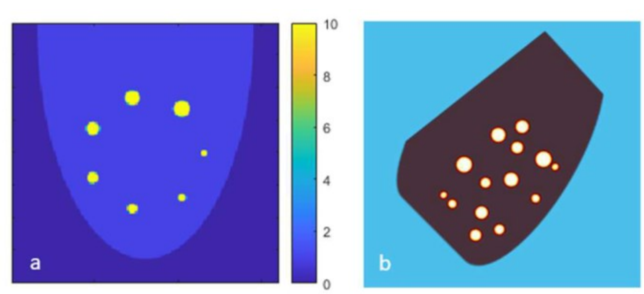

Figure 3. Phantom with two layers of 7 spherical inserts; a) transaxial (x-z) plane, b) maximum-intensity projection at an oblique angle, showing all spheres.

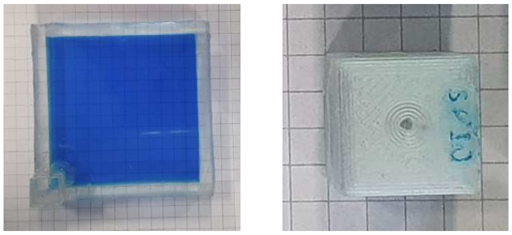

Figure 4. Homogeneous box phantom (left) with 30 x 30 x 20 mm cavity and spherical phantom (right) with a cavity of 6 mm in diameter.

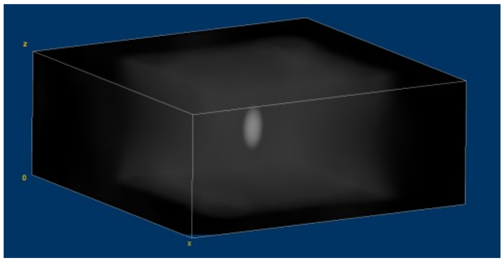

Figure 5. Volumetric view of the reconstructed image based on the combined measurements of the two phantoms.

reference, simulated 2D parallel-hole data was obtained with the same phantom configurations using a previously validated Monte Carlo transport simulation.

III. Results

The The spectral performance of the new prototype was characterised using collimated ^{241}Am and ^{57}Co sources. This was compared to the performance of the previous system using the same CZT crystal, as shown in Fig.6. The average FWHM energy resolution improved from 3.75% (7.56%) to 2.72% (4.01%) for 122 keV (59.5 keV). The lower noise floor of the system allowed us to decrease the minimum energy pixel threshold from ~45 keV to ~10 keV.

The DOI resolution was measured at six depth positions. The FWHM of the beam spot calculated from the GEANT4 simulations was 0.94 mm. The deconvolved DOI resolution is shown in Table 1 and was found to be less than 1 mm at most depth positions.

Achieving the CNR of ~15 was obtained with collimator hole size of 1.75 mm, opening angle of 90° and inter-aperture

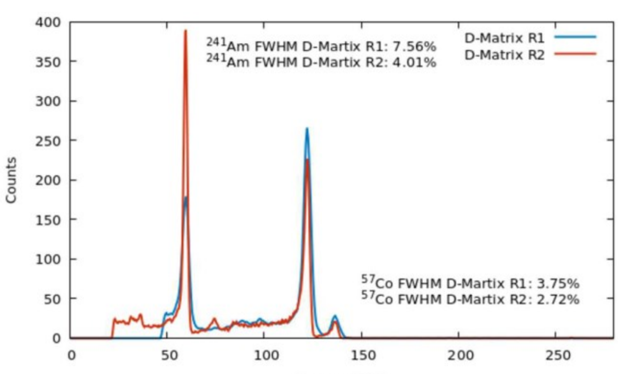

Figure 6. Comparison between the raw 57Co spectra measured with previous D-Matrix R1 (blue line) and new R2 (red line) system.

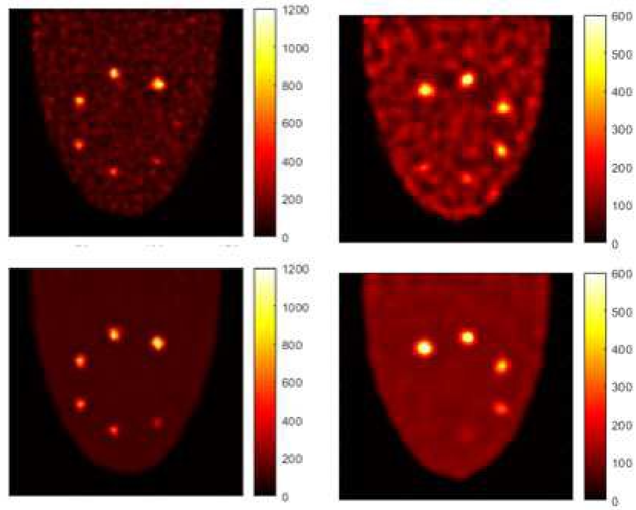

Figure 7. Reconstructed images from the simulation study from close to the detector (left) and middle (right) planes without (top) and with (bottom) NLM noise filtering.

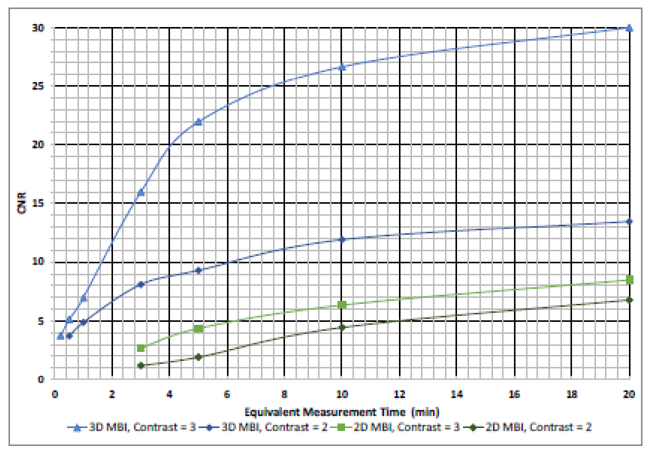

Figure 8. CNR values for different tumour-background ratios (contrast = 2 and 3) as the function of measurement time. Blue lines represent the values measured with the multi-pinhole 3D MBI prototype, while green lines show simulated 2D parallel-hole data.

spacing of 10 mm. Fig.7 shows reconstructed images from the middle and edge (close to the detectors) planes of the simulated phantom, without and with NLM-filtering. The dose/time combination in the simulations was 8 times lower comparing to the current clinical practice.

The quantitative evaluation of the ^{99m}Tc phantom measurements provided CNR data for the 3D MBI system that we could directly compare with the parallel hole performance. Fig.8 shows this comparison with different measurement times and tumour-background ratios.

IV. Discussion & Conclusions

A new feasibility prototype of a stationary tomographic MBI system has been manufactured and characterised. The collimator geometry was further optimised in the simulation study by including NLM noise filtering. The results show an increase in the CNR by almost a factor of 3. Our previous results [7-9] indicated that we can achieve the goal of dose reduction to the mammography level. The current results with

NLM filtering suggest that further significant reduction in the dose/time combination is possible with a minimum size of detected lesions of 5-6 mm. The CNR-based comparison of the new 3D and the conventional 2D imaging suggests a strong improvement in imaging performance in terms of lesion detectability that could be potentially translated into a reduction of measurement time or patient dose comparable to the expectations from the simulation studies.

DOI	Measured FWHM	Deconvolved FWHM
1.0 mm	1.17 mm	0.93 mm
1.5 mm	1.20 mm	0.97 mm
2.0 mm	1.19 mm	0.98 mm
2.5 mm	1.24 mm	1.02 mm
3.5 mm	1.17 mm	0.93 mm

Table 1. DOI resolution (FWHM) at different depths.

REFERENCES

[1] D. J. Rhodes, C. B. Hruska, A. L. Conners, C. L. Tortorelli, R. W. Maxwell, K. N. Jones, A. Y. Toledano, M. K. O'Connor, "Molecular breast imaging at reduced radiation dose for supplemental screening in mammographically dense breasts", AJR Am J Roentgenol, vol. 204, no. 2, pp. 241-251, Feb. 2015.

[2] C. B. Hruska, "Molecular breast imaging for screening in dense breasts: state of the art and future directions", AJR Am J Roentgenol, vol. 208, no. 2, pp. 275-283, Feb. 2017.

[3] J. P. Shah, S. D. Mann, R. L. McKinley, M. P. Tornai, “Implementation and CT sampling characterization of a third-generation SPECT-CT system for dedicated breast imaging”, J. of Medical Imaging, vol. 4, no. 3, p. 033502, Jul. 2017.

[4] J. van Roosmalen, F. J. Beekman, M. C. Goorden, “Comparison of fanbeam, slit-slat and multi-pinhole collimators for molecular breast tomosynthesis”, Phys. Med. Biol., vol. 63, no. 10, pp. 105009-105020, May 2018.

[5] J. J. James, S. L. Tennant, “Contrast-enhanced spectral mammography (CESM)”, Clin. Radiol., vol. 73, no. 8, pp. 715-723, Aug. 2018.

[6] J. van Roosmalen, M. C. Goorden, F. J. Beekman, “Molecular breast tomosynthesis with scanning focus multi-pinhole cameras”, Phys. Med. Biol., vol. 61, no. 15, pp. 5508-28, Jul. 2016.

[7] B. F. Hutton, K. Erlandsson, A. Wirth, I. Baistow, K. Thielemans, A. Cherlin, “Design of an ultra-low-dose, stationary, tomographic Molecular Breast Imaging system”, Proc IEEE NSS/MIC 2021.

[8] A. Cherlin, A. Wirth, K. Erlandsson, I. Baistow, K. Thielemans, B. F. Hutton, “A New Concept for a Low-Dose Stationary Tomographic Molecular Breast Imaging Camera Using 3D Position Sensitive CZT Detectors”, Proc IEEE NSS/MIC 2021.

[9] K. Erlandsson, A. Wirth, K. Thielemans, I. Baistow, A. Cherlin, B. F. Hutton, “Challenges in optimization of a stationary tomographic molecular breast imaging system”, Proc IEEE NSS/MIC 2021.

[10] A. Forgacs, P. Kallos-Balogh, F. Nagy, A. K. Krizsan, I. Garai, L. Tron, “Activity painting: PET images of freely defined activity distributions applying a novel phantom technique”, PLoS ONE., vol. 14, no. 1, e0207658, Jan. 2019.

[11] S. Jan et al., "GATE: a simulation toolkit for PET and SPECT," Phys. Med. Biol., vol. 49, no. 19, pp. 4543-4561, Oct. 2004.

[12] K. Erlandsson, A. Wirth, K. Thielemans, I. Baistow, A. Cherlin, B. F. Hutton, “Optimization of a Stationary Tomographic MBI System Including Non-Local Means Filtering”, IEEE NSS MIC 2022.