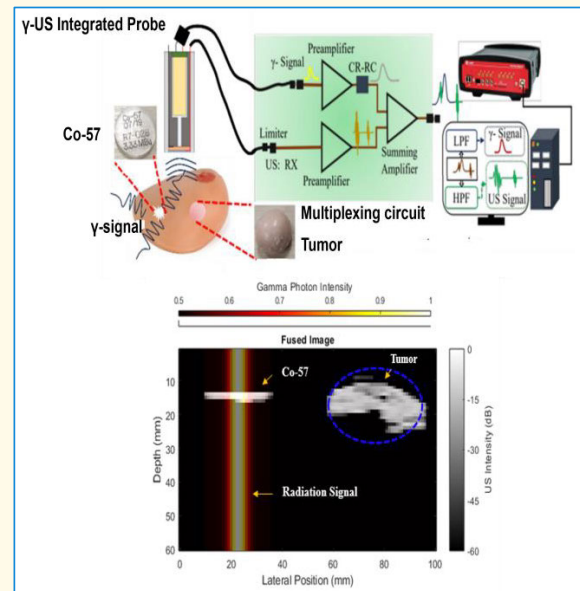


Development of an Integrated Gamma-Ultrasound (γ -US) Probe Detector and Multiplexed Readout for Simultaneous γ -US Detection

Muhammad Nasir Ullah¹, Muhammad Waqar¹, Graduate Student Member, IEEE, Yuseung Park², Gyeong Beom Kim, Kyungseok Choi, Seungjoo Jang, Chanyoung Kim³, Graduate Student Member, IEEE, Qifa Zhou⁴, Fellow, IEEE, Hojong Choi⁵, and Jung-Yeol Yeom, Member, IEEE

Abstract—This article presents the first fully integrated, highly compact single-element integrated gamma-ultrasound (γ -US) probe for simultaneous molecular and anatomical imaging. The probe integrates a PZT-5A ultrasound (US) transducer with a Ce:GAGG scintillator and silicon photomultiplier (SiPM), enabling dual-modality detection along the same line-of-sight (LOS). The tungsten-epoxy composite serves dual roles as both the US backing layer and γ collimator, making it innovative and significantly reduces size and complexity. Furthermore, a single-channel frequency-based multiplexed circuit enables real-time dual-modality data acquisition (DAQ). The γ part of the probe achieves 17.3% energy resolution (FWHM at 122 keV), 80.0 cps/MBq sensitivity, and 2.69 mm spatial resolution at 1 cm distance in air. The US component provides a 36.0% fractional bandwidth (FBW), 5.95 dB signal-to-noise ratio (SNR), 1.32 mm axial resolution, and 4.90 mm lateral resolution at 5 cm depth in water. Finally, a fused B-mode image of a breast phantom validates the probe's imaging capability. This integrated probe design offers a compact, cost-effective solution for handheld medical diagnostics and nondestructive testing (NDT), with minimal performance trade-offs compared to standalone systems.

Index Terms—Hybrid Probes, scintillator, silicon photomultiplier (SiPM), ultrasound (US) probe.



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Muhammad Nasir Ullah was with the Department of Bioengineering, Korea University, Seoul 02841, South Korea. He is now with the Department of Radiology, School of Medicine, Stanford University, Stanford, CA 94305 USA (e-mail: nasir909@stanford.edu).

Muhammad Waqar and Chanyoung Kim are with the Department of Bioengineering and the Interdisciplinary Program in Precision Public Health, Korea University, Seoul 02841, South Korea.

Yuseung Park, Gyeong Beom Kim, Kyungseok Choi, and Seungjoo Jang were with the Department of Bioengineering, Korea University, Seoul 02841, South Korea. They are now with Samsung Electronics, Suwon 16677, South Korea.

Qifa Zhou is with the Department of Biomedical Engineering, University of Southern California, Los Angeles, CA 90089 USA.

Hojong Choi is with the Department of Electronic Engineering, Gachon University, Seongnam 13120, South Korea (e-mail: hojongch@gachon.ac.kr).

Jung-Yeol Yeom is with the School of Biomedical Engineering, Department of Bioengineering and the Interdisciplinary Program in Precision Public Health, Korea University, Seoul 02841, South Korea (e-mail: jungyeol@korea.ac.kr).

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I. INTRODUCTION

IN RECENT decades, surgical interventions have been transformed by the widespread adoption of intraoperative probes [1]. They have a range of applications, from accurately localizing sentinel lymph nodes in cancer staging to guiding complex procedures such as parathyroidectomy [2]. By integrating diverse image modalities such as US [3], fluorescence Raman spectroscopy [4], radio frequency (RF) [5], and nuclear medicine (NM) using ionizing radiation [6]—intraoperative probes now provide comprehensive anatomical, physiological, and molecular information and have revolutionized the surgical management of complex pathologies.

NM probes, such as gamma (γ) and positron (β^+) probes, are extensively used in intraoperative procedures, with the γ -probe being particularly common. These probes detect ionizing radiation emitted by radiolabeled pharmaceuticals within a patient's body, providing insights at the molecular or cellular level depending on specific radiopharmaceutical used [7]. While certain probes, such as γ -cameras, provide 2-D images of regions of interest (ROI), others, simply referred to as probes, offer essential data regarding radiopharmaceutical concentration in counts per second (cps). Despite their high molecular sensitivity, γ -probes lack anatomical context,

Highlights

- **A novel fully integrated, highly compact γ -US probe along with frequency multiplexed single-readout circuitry system for simultaneous molecular and anatomical information/image acquisition.**
- **Demonstrates optimal γ (sensitivity, energy and spatial resolution, and side-shielding rate) and US (FBW, SNR, and axial/lateral resolution) performance, with fused γ -US phantom imaging.**
- **Significant advancement in compact integrated dual modality detector design and optimized cost-effective readout circuitry for intraoperative procedures and NDT.**

limiting their ability to provide integrated structural and functional information during surgery [8].

Therefore, integrating γ -probes with other modalities to obtain complementary anatomical information shows significant potential. Each modality—whether it be US [9], near-infrared (NIR) [10], or magnetic resonance imaging (MRI) [11]—has its own strengths and limitations. Integrating these modalities can improve diagnostic capabilities, although certain combinations may offer greater advantages than others. For instance, a hybrid, γ -NIR probe enables simultaneous detection of radiotracers (molecular sensitivity) and visualizing tissue perfusion and oxygenation (functional insight). However, NIR's limited penetration depth restricts its effectiveness to superficial tissues [12].

The US probes, recognized for their real-time capability, noninvasive imaging of soft tissues, exhibit a wide range of applications in disciplines such as tissue biomechanics assessment, joint inflammation diagnosis, musculoskeletal injury monitoring, and cardiovascular disease evaluation [13]. Integrating γ and US detection functionalities within a single probe offers significant advantages, allowing for the simultaneous acquisition of detailed anatomical and molecular information. This integration simplifies data acquisition (DAQ), reduces procedural complexities, and ultimately improves patient care and management [9].

Integrated handheld gamma-ultrasound (γ -US) probes have the potential to enhance both clinical and industrial applications by combining anatomical imaging with radiation dose or defect monitoring. In radiation therapy, this fusion can improve treatment precision and patient safety, while in nondestructive testing (NDT), it offers deeper insights into material properties and defects [14], [15]. Their portability and real-time imaging capabilities make them well suited for point-of-care use. However, designing a compact, lightweight, and power-efficient hybrid probe presents significant challenges. Most current systems rely on separate readout electronics for γ and US modalities, which, while providing high performance, increase system size, power consumption, and cost—limiting their clinical portability [16]. Additionally, prior approaches often depend on simulations or standalone systems that require postprocessing to align multimodal data [17]. To address integration complexity, our previous PRF-based method [9] acquired γ signals during US idle periods, reducing system demands but introducing dead time that compromised γ sensitivity.

We are developing a fully integrated handheld γ -US imaging probe for intraoperative applications and NDT. This article presents preliminary results from the first fully

integrated single-element γ -US probe and a novel single-channel frequency-based multiplexed readout circuit. The proposed design of the probe features a compact, dual-modality γ -US probe integrating a Ce:GAGG scintillator with a PZT transducer. The innovative multiplexed readout circuit leverages the frequency-dependent characteristics of US signals to enable simultaneous γ and US DAQ without dead time or sensitivity loss. The combination of the compact detector design and the multiplexed readout circuit offer a cost-effective and handheld dual-modality imaging solution.

This article is organized as follows. Section II describes the materials and methods, with five subsections covering (Section II-A) the standalone γ -probe detector design; (Section II-B) the standalone US probe detector design; (Section II-C) the integrated γ -US probe detector design; (Section II-D) the proposed frequency-based multiplexing front-end circuit; and (Section II-E) DAQ and processing. Section III outlines the experimental setup. Section IV presents a comparative analysis of the standalone and integrated probes performance, including imaging feasibility and the evaluation of the proposed frequency-based multiplexing front-end readout circuit. Section V discusses the results and future research directions, and Section VI concludes this article.

II. MATERIALS AND METHODS

To assess the performance of the integrated γ -US probe detector, custom standalone γ and US probe detectors were developed as benchmarks. The following sections detail the design, components, fabrication methods, and testing protocols for each standalone probe to ensure accurate and reliable performance comparisons with the integrated probe.

A. Standalone Gamma (γ)-Probe Detector

In this study, a custom-designed standalone γ -probe detector was developed as a benchmark for comparison with the γ -component of the integrated γ -US probe. The detector integrates a $2 \times 2 \times 15$ mm³ cerium-doped Gd₃Al₂Ga₃O₁₂ (Ce:GAGG) crystal scintillator [18], [19], [20], mechanically polished, wrapped in Teflon reflective tape, and optically coupled to a 2×2 mm² silicon photomultiplier (SiPM) (Hamamatsu S13360-2050VE) using optical epoxy. Additionally, a cylindrical tungsten collimator (12 mm outer diameter, 30 mm length) with a 1.3 mm diameter, 15 mm deep bore was used to achieve a 3.3 mm field-of-view (FOV) at a 1 cm distance. The spatial resolution and detection sensitivity are trade-offs to consider, and this FOV was selected to optimize

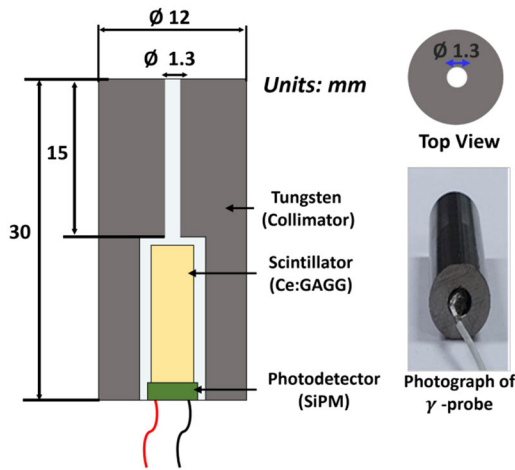


Fig. 1. Schematic, design specifications, and a photograph of the custom standalone γ -probe detector are presented. The probe integrates a Ce:GAGG scintillator coupled to a SiPM for γ -photon detection, while tungsten collimator ensures directional sensitivity.

geometric efficiency while maintaining resolution as confirmed by ^{57}Co point-source measurements [see Section IV-D]. This design ensures sufficient sensitivity for low-activity tracers (e.g., ^{57}Co or $^{99\text{m}}\text{Tc}$) without compromising target discrimination. Tungsten was selected for its high atomic number, density, stopping power, and cost-effectiveness compared to alternatives such as lead [21]. The targeted γ -energy of this probe is to detect low-to moderate γ -photons, that is, $^{99\text{m}}\text{Tc}$ (140 keV, used in clinical settings) [22] and ^{57}Co (122 keV, used in this study). Fig. 1 shows a schematic with detailed specifications and a photograph of the custom-designed standalone γ -probe.

B. Standalone US-Probe Detector

Like the γ -probe, a standalone US probe detector was custom-designed and developed as a benchmark for assessing the US component of the integrated γ -US probe detector. The US probe detector used a 0.2 mm thick PZT-5 A piezoceramic element to achieve a 10 MHz center frequency. A 0.1 mm-thick EPOTEK-301 matching layer was applied to the front face, and a tungsten-epoxy compound was used as a backing material. The tungsten is an ideal candidate for backing materials in US due to its higher acoustic impedance and density, which effectively dampen backward waves and signal reflections [23], [24]. The final US probe detector has a cylindrical shape with an outer diameter of 12 mm, and a height of 30 mm. Fig. 2 shows a schematic of the US probe detector with design specifications, along with an actual photograph.

C. Integrated γ -US Probe Detector

In this proof-of-concept study, we developed the first fully integrated γ -US probe detector [25] shown in Fig. 3. To reduce size and weight, we used a dual-purpose tungsten component that serves as both the γ collimator and US backing material, two of the most space-intensive parts. This compact design maintains a shared line of sight (LOS)

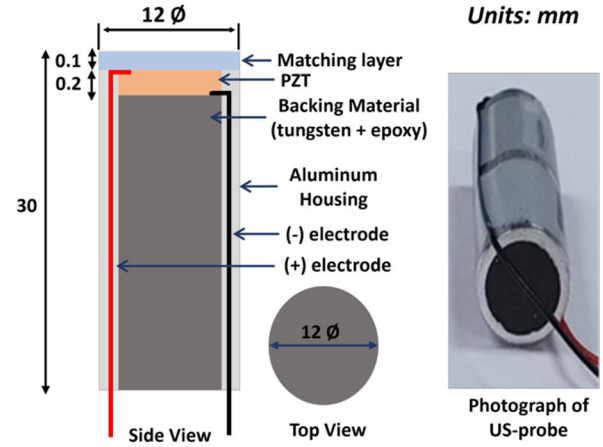


Fig. 2. Schematic, design specifications, and a photograph of custom standalone US probe detector are shown. The probe features a PZT-5 A transducer with 0.2 mm thickness and 0.1 mm matching layer. The overall dimensions are 30 mm in height and 12 mm in diameter.

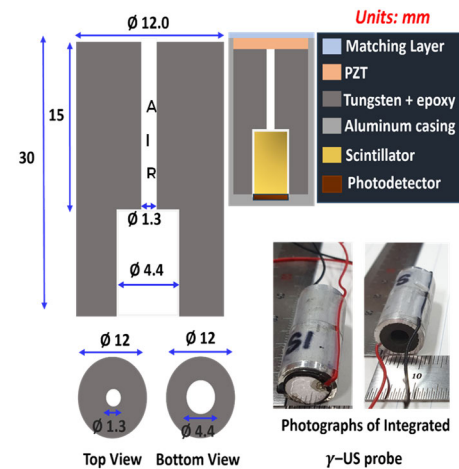


Fig. 3. Schematic, design specifications, and a photograph of the custom-designed integrated γ -US probe detector.

for both modalities without compromising performance. The specifications of the γ -component of the integrated probe are kept the same as discussed in Section II-A (1.3 and 4.4 mm holes in tungsten-epoxy layer for γ collimation and scintillator-SiPM assembly, respectively) with targeted low-to-moderate γ energies, that is, $^{99\text{m}}\text{Tc}$ (140 keV) or ^{57}Co (122 keV). The integrated probe has the same transducer specification as discussed Section II-B, that is, a 10 MHz PZT transducer with similar matching and backing layers to offer an axial resolution (~ 1 mm) with optimal penetration (3–5 cm), ideal for superficial imaging like breast tumor or lymph node localization. Its compact size favors high-frequency use for better resolution in short-range applications. Furthermore, the white space in Fig. 3 shows air or free space to provide a smooth collimation path for γ photons.

D. Frequency-Based Multiplexing Front-End Readout Circuit

In a previous study [9], we used a switching readout method to share a single ADC channel between γ and US

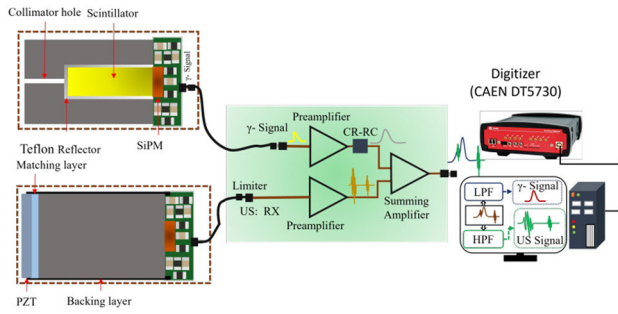


Fig. 4. Schematic of the frequency-based multiplexed readout circuit is shown, detailing all components from the detectors to postprocessing for both the γ and US probes. The raw signals are first preprocessed and then combined via a summing amplifier, which multiplexes the γ and US signals. The resulting signal is digitized through a single channel of the digitizer and transferred to the DAQ system for postprocessing.

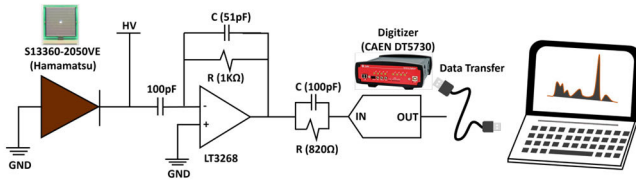


Fig. 5. Schematic of the front-end readout circuit used for γ -probe and γ -component of the integrated γ -US probe detector signal acquisition. The circuit includes CSA and PZC stages for processing the SiPM output signal.

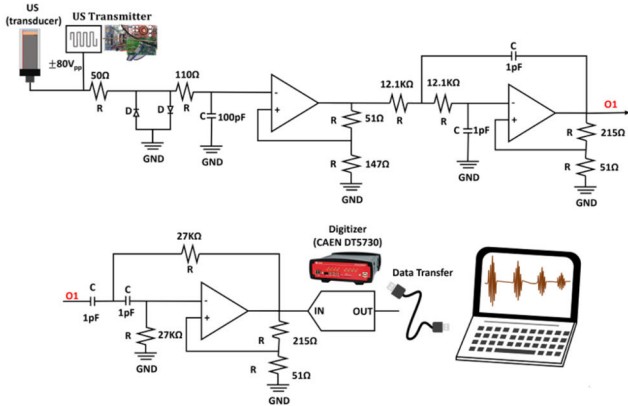


Fig. 6. Schematic of the front-end readout circuit for the US probe and the US component of the integrated γ -US probe detector. An active BPF with a 1–15 MHz bandwidth was implemented for signal acquisition.

signals, but it introduced γ dead time. Here, we present a new frequency-based multiplexing approach that enables true simultaneous acquisition on a single channel by exploiting the inherent frequency separation between γ and US signals, eliminating delay and dead time (Fig. 4). Signals (γ and US) from integrated probe (Fig. 3) are simultaneously acquired and preprocessed (Figs. 5 and 6) and combined using a summing amplifier before being fed to the ADC for real-time analysis.

E. DAQ and Processing

Custom-designed electronics were used to acquire data from the standalone γ -probe, US probe, and integrated γ -US probe. The DAQ can be divided into three main steps: front-end

readout, signal processing, and post-processing. Dedicated circuits were developed and optimized for each signal type— γ , and US—to ensure accurate and efficient DAQ.

1) *γ -Probe Detector Front-End Readout Circuitry*: A custom-designed charge-sensitive amplifier (CSA) and pole-zero cancellation (PZC) circuit were implemented to process signals from the γ -probe detector using a Ce:GAGG scintillator and SiPM. The event rate is primarily determined by the radioactive source's activity and system factors such as detectors and probe efficiency. When γ -photons interact with the scintillator, visible light is produced and detected by the SiPM, which converts it into an electrical signal. The resulting pulse rate is thus proportional to the source's activity. We benchmarked count-rate handling with a readily available ^{57}Co source that is often used in place of $^{99\text{m}}\text{Tc}$ used in clinical setting (0.0355 MBq at time of experiment) to validate the multiplexed readout and γ -US detection under controlled, nonsaturating conditions aligned with intraoperative γ -probe targets (≈ 103 – 105 c/s, geometry/efficiency dependent). For clinical workflows using $^{99\text{m}}\text{Tc}$ (10–100 MBq, protocol-dependent) as a detector probe, photon fluxes may be higher, but our readout electronics (100-MHz field-programmable gate array (FPGA), 1- μs shaping) is designed to function with negligible count loss at clinically relevant activities. The CSA converts the charge generated by the SiPM upon detecting γ -photons into a voltage signal, ensuring high gain and low noise to accurately capture the rapid, low-amplitude pulses characteristic of the SiPM output. To refine the signal, the PZC circuit compensates for the exponential decay of the SiPM pulse, minimizing undershoot and improving baseline stability [26]. This is achieved with a resistor-capacitor (RC) network that cancels the pole introduced by the CSA, resulting in a more accurate and stable output for subsequent processing. The schematic circuit for pre-processing the γ -probe detector signals, including the CSA, PZC, and digitizer, is shown in Fig. 5.

2) *US-Probe Detector Front-End Readout Circuitry*: Like the γ -probe detector readout, custom-designed electronics were used for the US probe detector's raw receive (RX) signal acquisition and preprocessing. An active bandpass filter (BPF) circuit with a bandwidth of 1–15 MHz was implemented for signal conditioning. Additional RX circuit design can be found in [9]. For this proof-of-concept study, we used a commercial MD1213DB1 demo board. The board has a logic-controlled input interface and supports high-speed pulses, with an output stage capable of delivering up to ± 100 V with peak currents of ± 2.0 A. To generate an 80 V_{pp} signal, a commercial dc–dc boost converter was used. For the MD1213DB1 input logic control, a FPGA was configured to provide 3-cycle square wave pulses with a central frequency (f_c) of 10 MHz and pulse repetition frequency (PRF) of 5 KHz. A circuit schematic is shown in Fig. 6.

3) *DAQ and Analysis*: In this study, we used a commercial digitizer (CAEN DT5730) with a 14-bit resolution and a sampling rate of 500 MS/s. This digitizer facilitates communication and data transfer to the DAQ system via USB and optical links. Acquired data was stored in the DAQ system and analyzed using MATLAB, where key parameters were

extracted, quantified, and visualized. The γ -probe detector and its γ -US counterpart were evaluated based on energy resolution (FWHM). Energy histograms were analyzed by fitting exponential (Compton peak at the beginning of the fitting), Gaussian (photopeak), and linear functions (straight ending of the curve), with FWHM calculated using (1), where σ represents the Gaussian standard deviation. This method provided a precise assessment of the probe's performance

$$\text{FWHM} = 2\sqrt{2\ln(2)} \times \sigma. \quad (1)$$

In terms of US performance, both the SNR and fractional bandwidth (FBW) were measured for detectors of both the US probe and the integrated γ -US probe. The SNR was calculated using (2), where S represents the average signal intensity and N represents the average noise intensity level

$$\text{SNR}_{\text{dB}} = 20\log\left(\frac{S}{N}\right). \quad (2)$$

The FBW was determined using (3), where f_c shows the center frequency (geometric mean for logarithmic responses, and arithmetic mean for narrowband responses), f_L , and f_H denote, lower frequency, and higher frequency at -6 dB points, respectively [27]

$$\text{FBW}_{-6\text{dB}} = \left(\frac{f_H - f_L}{f_c}\right) \times 100. \quad (3)$$

The axial resolution of both the US component of the integrated probe and the standalone US probe was determined by measuring the length of the received (RX) echo pulse. This echo pulse length was measured from the -6 dB width of the envelope, and the axial resolution was calculated accordingly using (4), providing a realistic estimate of system performance [27]

$$\text{Axial Resolution} = \frac{c \times t}{2} \quad (4)$$

where t represents the echo length (time in μs) of the R_X echo pulse and c is the speed of sound in the medium. Finally, we computed lateral resolution of both the standalone US and US-component of integrated probes from the FWHM [see (1)] of the US signals at various lateral positions.

In terms of frequency-based multiplexed signal readout postprocessing, critical information such as detected γ -photon energy and count rate based on the energy window for ^{57}Co at 122 keV photopeak for γ -probe and the US echo signal amplitude and time difference between consecutive echoes, the multiplexed γ -US signal required to be demultiplexed and separated into two separate signals (γ and US).

To demultiplex the raw γ -US signals acquired using a single-channel readout circuit, low-pass, and BPFs (LPF and BPF) were implemented in MATLAB. The LPF was set with a cutoff frequency of 1 MHz to extract the γ -signal, whereas the BPF with a 1–15 MHz range effectively isolated the US signal (US transducer $f_c = 10$ MHz) from the multiplexed signal. Fig. 7(a) illustrates an example of the raw multiplexed signal containing both γ and US components, while Fig. 7(b) and (c) depicts the separated γ and US signals following LPF and BPF, respectively. Subsequently, the data was segmented into 1-s intervals using embedded timestamps. For the γ signal,

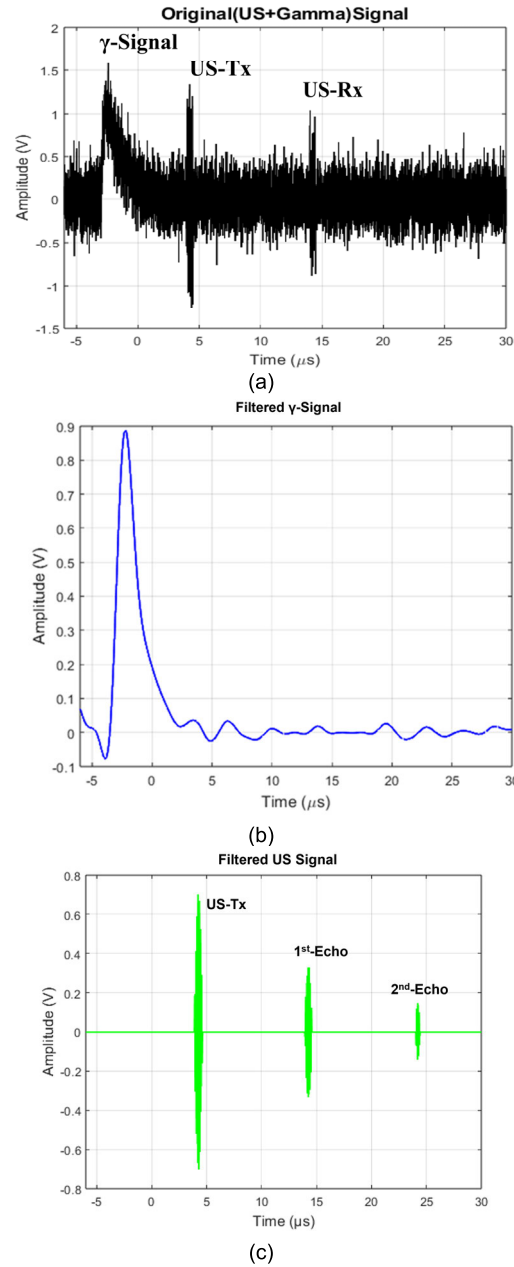


Fig. 7. (a) Raw data acquired using the proposed frequency-based multiplexed method for simultaneous γ and US signal readout via a single channel. (b) and (c) Separated γ and US signals, respectively, obtained by an LPF and BPF to the raw multiplexed data. The results show the accurate decoding of both modalities using simple postprocessing, validating the effectiveness of the proposed single-channel readout method.

a defined energy window of 100–133 keV (approximately ± 18 keV) was then applied to isolate the photo-peak counts, to evaluate CPS. US performance was evaluated by computing SNR, FBW, lateral, and axial resolution.

III. INTEGRATED γ -US PROBE DETECTOR PERFORMANCE EVALUATION EXPERIMENTAL SETUP AND FREQUENCY-BASED SIGNAL MULTIPLEXING READOUT ASSESSMENT

The performance evaluation of the proposed γ -US integrated detector was conducted through a series of experiments,

benchmarked against the standalone γ and US probe detectors. For the γ -component in the integrated γ -US detector, key parameters assessed including sensitivity, spatial resolution, side shielding of the collimator, and energy resolution. The US component was evaluated based on SNR, FBW, axial resolution, and lateral resolution, with results compared to those of the standalone US probe.

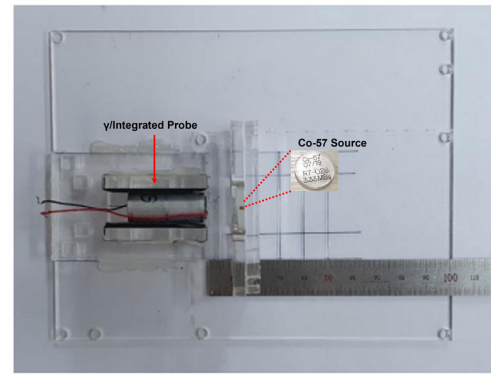
The performance of the proposed frequency-based multiplexed γ -US signal readout, data were acquired using both dedicated γ and US circuits (Figs. 5 and 6) and the integrated single-channel multiplexed circuit. In terms of γ -probe performance, two fundamental key parameters, namely sensitivity, and energy resolution, were compared with those obtained from the dedicated γ -probe readout, whereas the US-probe parameters such as SNR and FBW were assessed. Furthermore, the proposed frequency-based multiplexing method was compared to our previously reported switching-based readout method [9], highlighting improvements in continuous acquisition without any dead time.

A. γ -Component Performance Experimental Setup

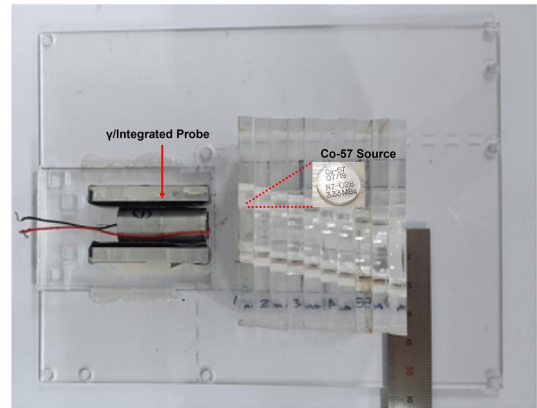
All γ -probe detector experiments were performed in the air without any scattering medium using a ^{57}Co (0.0355 MBq, M1-760, Eckert and Ziegler Isotope Products) sealed radiation point source.

The sensitivity of both the γ -probe and the γ -component of the integrated γ -US probe was assessed using a ^{57}Co radiation source placed 1 cm away from the detector along the axial direction. The probe remained fixed while the source was moved in 1 cm increments up to 5 cm. At each distance, the count rate measured in CPS was recorded over a 10-min interval [Fig. 8(a)]. Similarly, the spatial resolution was evaluated using the same setup as the sensitivity experiment. The ^{57}Co source was moved radially ± 8 cm from the center of the collimator hole of the γ -detector, with the axial distance fixed at 1 cm and a step size of 0.5 cm. At each radial step, the CPS was recorded over a 3-min interval, and data were plotted against the radial step positions to find spatial resolution [Fig. 8(b)]. Furthermore, the collimator shielding tests were performed by positioning the ^{57}Co source 7 cm away from each side of both the γ -probe and the integrated γ -US probe detector. Counting rates (CPS) were recorded for three different sides for each type of probe. Data was saved for each position over 20 min [Fig. 8(c)]. Lastly, for energy resolution measurement of both the γ -probe and the integrated γ -US probe detectors, the ^{57}Co source was positioned at the maximum sensitivity position of each probe. Approximately 100 000 γ -photons counts were recorded.

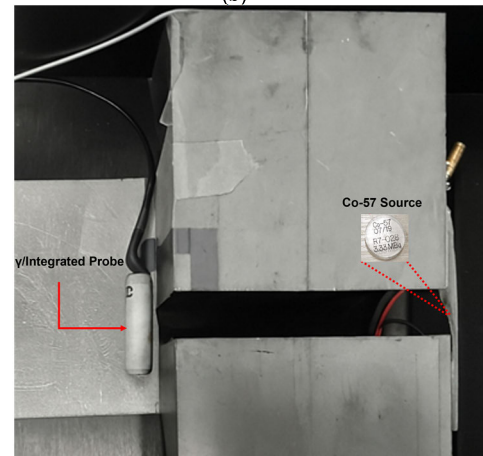
To generate the energy spectra, voltage pulses from individual γ -ray interactions were recorded over a fixed acquisition period, with each pulse's peak voltage corresponding to the energy deposited by an incident photon. These peak voltages were extracted during postprocessing to construct a histogram, where the x -axis represents pulse height (in Volts) and the y -axis indicates the event count per energy bin. An energy window corresponding to the photopeak energy (e.g., 122 keV for ^{57}Co) was applied to exclude events outside the desired



(a)



(b)



(c)

Fig. 8. Photograph of the performance evaluation setup for the γ -probe and the γ -component of the integrated γ -US probe. (a) Sensitivity was measured by fixing the probe and moving the ^{57}Co source axially. (b) Spatial resolution was measured by moving the source radially and (c) side shielding was assessed by irradiating the side of the probe with the ^{57}Co source.

range, thereby enhancing spectral resolution and reducing background noise.

B. US-Component Performance Assessment Experimental Setup

The key performance parameters such as axial resolution, lateral resolution, FBW, and SNR of the US probe and its

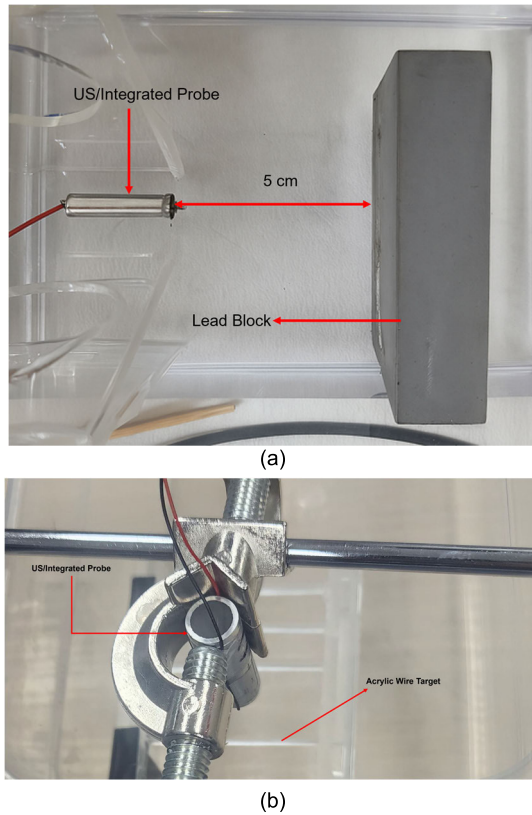


Fig. 9. Photograph of the performance evaluation setup for the US probe and the integrated γ -US probe detectors, illustrating the axial (a) with a lead block target placed at 5 cm from the probe and (b) lateral resolution experimental setup with acrylic wire target with a diameter of 1 mm was placed at a depth of 5 cm from the probe.

counterpart in the integrated probe were assessed experimentally. For axial resolution, FBW, and SNR, the standalone probe and the integrated γ -US probe detectors were submerged in water, with a lead block placed at 5 cm from the probe, whereas an acrylic rod of 1 mm diameter used as the target material for calculating the lateral resolution of the probes.

The axial resolution for the US probe and the integrated γ -US probe detectors were measured using the experimental setup shown in Fig. 9(a). Similarly, Fig. 9(b) shows the experimental setup for calculating the lateral resolution of the standalone US probe detector and its counterpart in the integrated probe detector. The acrylic target is fixed at a depth of 5 cm in water, and the probe was moved in 0.5 mm steps along the axial direction up to 5 cm from the center of the FOV. The data were plotted against each position and FWHM was computed.

C. Imaging Validation of Integrated γ -US Probe Detector

In this study, we demonstrated the feasibility of acquiring fused γ -US image using our proposed integrated γ -US probe detector. As mentioned before, ^{57}Co was used to simulate a tumor with localized uptake of γ -emitting tracers, mimicking clinical scenario. To create a realistic tissue environment, a commercial breast phantom (VK-10 221, gaMPT Ultrasonic Solutions) was modified by carefully opening and filling it with US gel. The phantom contained two targets positioned

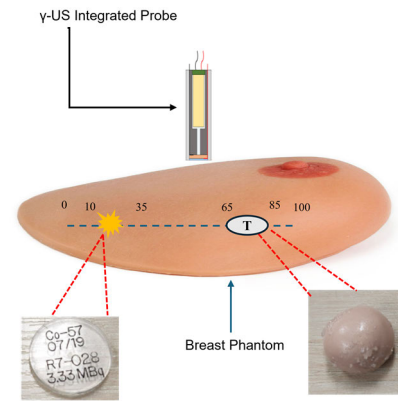


Fig. 10. Schematic of the experimental setup for assessing the imaging performance of the integrated γ -US probe detector. Simultaneous acquisition of γ -photons and US signals was conducted by scanning the breast phantom containing a ^{57}Co radiation source and a tumor target with a 1 mm step size.

at 10 mm depth: a 25.4 mm diameter ^{57}Co radiation source, and 20 mm diameter tumor-mimicking object. The source was located laterally at 17.7 mm (spanning 10–35.4 mm), and the tumor mimic at 75.4 mm (spanning 65.4–85.4 mm), maintaining a 30 mm separation for clear target distinction in the B-mode image. The probe was scanned laterally in 1 mm steps across a 100 mm path, simultaneously acquiring γ -photon counts and US signals at each position. The experimental setup is illustrated on Fig. 10.

IV. RESULTS

To comprehensively evaluate the proposed integrated detector, its γ and US components were individually compared against standalone γ and US probe detectors.

A. γ -Probe and γ -Component of Integrated γ -US Probe Result

The feasibility of the γ -component in the integrated probe was validated by comparing energy resolution, sensitivity, spatial resolution, and shielding rate with those of the standalone γ -probe by placing the ^{57}Co at 1 cm from the probe in air.

The energy resolution (FWHM) showed minimal difference among the integrated probe and a standalone γ -probe, with values of 17.3% and 16.9%, respectively. Energy histograms of γ -probe (left) integrated γ -US probe detector (right) are shown in Fig. 11(a). Similarly, the integrated probe achieved a sensitivity of 80.0 c/s/MBq compared to 90.0 c/s/MBq given by the standalone γ -probe as shown in Fig. 11(b). The spatial resolution for the γ -probe and integrated probe detectors were measured at 2.32 mm FWHM and 2.69 mm FWHM, respectively [Fig. 11(c)]. Finally, the integrated probe showed a side shielding rate of 95.9% as compared to 97.1% of the standalone γ probe. Tables I and II summarize the γ performance of both the standalone γ and integrated probes.

B. Comparative Results of the US-Component of the Integrated γ -US Probe With the Standalone US-Probe

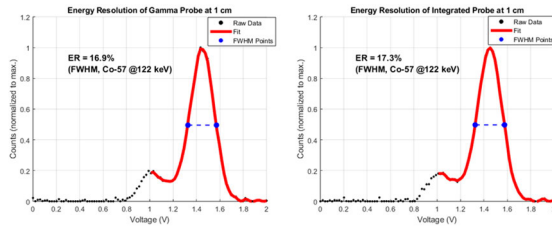
The FBW was determined by acquiring echo signals and applying (3). The standalone US probe achieved an FBW of

TABLE I

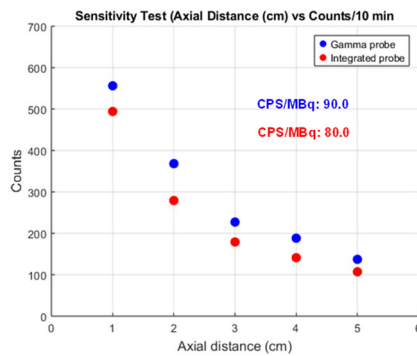
SIDE-SHIELDING (20-MIN COUNTS) AND ON-AXIS SENSITIVITY (10-MIN COUNTS, CPS/MBQ) FOR THE γ -PROBE AND INTEGRATED γ -US PROBE (^{57}Co)

SIDE SHIELDING RATE RESULTS OF γ AND INTEGRATED PROBE							
	Counts/20 min	Side Shielding Rate (%)					
γ -Probe	44	97.1					
Integrated Probe	62	95.9					
No Collimator	1513	0.0					

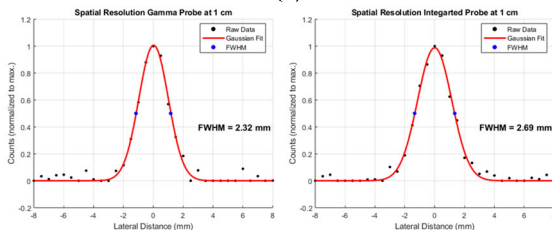
SENSITIVITY TEST RESULTS OF γ AND INTEGRATED PROBE							
Counts/10 min	Distance (cm)	1	2	3	4	5	CPS/MBq
γ -Probe		556	368	227	188	137	90.0
Integrated Probe		494	279	179	141	107	80.0



(a)



(b)



(c)

Fig. 11. (a) Energy Spectrum: Left plot shows energy distribution for standalone γ -probe detector; right plot for integrated γ -US probe detector. (b) Sensitivity Plot: Graph illustrating the sensitivity of γ -probe and the integrated γ -US probe detectors. (c) Spatial Resolution Plot: Depiction of spatial resolution performance; left plot for standalone γ -probe detector, and right plot for integrated γ -US probe detector.

TABLE II

PERFORMANCE OF THE INTEGRATED γ -US PROBE VS. STANDALONE PROBES: γ -COMPONENT (ENERGY RESOLUTION (FWHM, %), SENSITIVITY (CPS/MBQ), SPATIAL RESOLUTION (mm), SIDE-SHIELDING (%)) AND US-COMPONENT (SNR (dB), -6 dB FBW (%), LATERAL/AXIAL RESOLUTION (mm) AT 5 cm)

γ -COMPONENT OF INTEGRATED PROBE PERFORMANCE AND COMPARISON WITH STANDALONE γ -PROBE		
Performance Parameter	Standalone γ -Probe	Integrated Probe
Energy Resolution (FHM,%)	16.9	17.3
Sensitivity (CPS/MBq)	90.0	80.0
Spatial Resolution (mm)	2.32	2.69
Side Shielding Rate (%)	97.1	95.9

US-COMPONENT OF INTEGRATED PROBE PERFORMANCE AND COMPARISON WITH STANDALONE US-PROBE		
Performance Parameter	Standalone US-Probe	Integrated Probe
SNR (dB)	6.49	5.95
FBW (-6 dB bandwidth, %)	47.8	36.0
Lateral Resolution at 5 cm (FWHM, mm)	4.48	4.90

AXIAL RESOLUTION AT 5 cm (mm)			
Probe Type	Echo length (μ s)	c (m/s) in water	Axial resolution (mm)
US- Probe	148	1500	1.11
Integrated Probe	176	1500	1.32

47.8% at a f_c of 9.78 MHz, while the proposed integrated γ -US probe detector demonstrated an FBW of 36.0% at a f_c of 9.51 MHz [Fig. 12(a)]. The observed SNR for the standalone US probe and the integrated γ -US probe detectors were 6.49 and 5.95 dB, respectively (Table II). The lateral resolution of standalone US probes and the US component within the integrated γ -US probe was determined to be 4.48 and 4.90 mm (FWHM), respectively, at 5 cm. Fig. 12(b) illustrates the lateral resolution plot of both the standalone US and integrated γ -US probe detectors. Finally, the axial resolution was computed from the echo length. The echo length in the received echo signal from the standalone US probe and integrated γ -US probe were 148 and 176 μ s, respectively. Our results showed an axial resolution of 1.32 mm for the

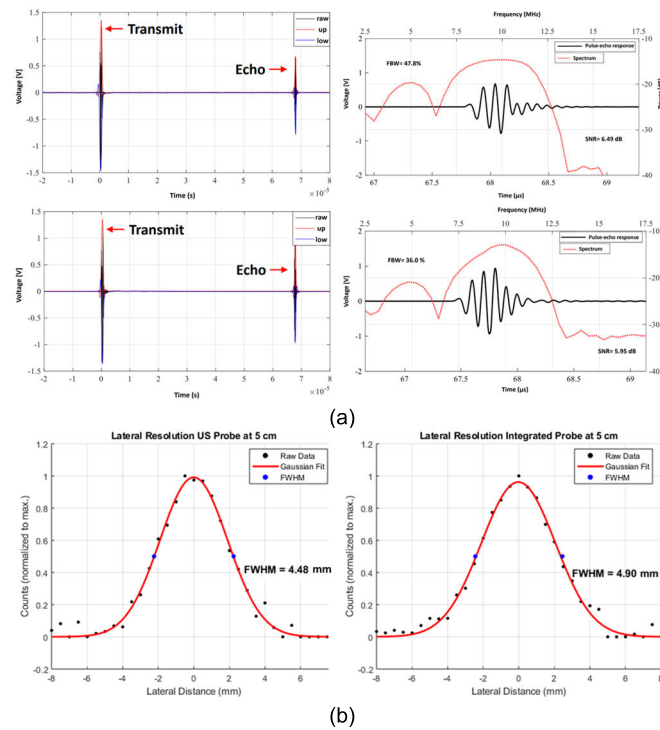


Fig. 12. (a) Top plot shows the frequency bandwidth (FBW) and SNR of the standalone US probe, while the bottom plot presents the FBW and SNR of the integrated γ -US probe detector. (b) Lateral resolution plot comparing the US probe (left) and the integrated γ -US probe detector (right).

TABLE III

PERFORMANCE COMPARISON OF γ AND US-PROBE READOUT METHODS—DEDICATED, SWITCHING [14], AND FREQUENCY-MULTIPLEXED—SHOWING γ : ENERGY RESOLUTION (FWHM, %) AND SENSITIVITY (CPS/MBq); US: SNR (dB) AND -6 dB FBW (%)

PERFORMANCE COMPARISON OF γ AND US-PROBE READOUT METHODS			
γ -PROBE	Dedicated Readout	Switching Method [14]	Frequency Based Multiplexed Readout
Energy Resolution (FWHM, %)	19.2	20.7	19.5
Sensitivity (CPS/MBq)	90.0	72.5	89.7
US-PROBE	Dedicated Readout	Switching Method [14]	Frequency Based Multiplexed Readout
SNR (dB)	14.95	10.68	22.98
FBW (-6 dB, %)	36	32.0	32.0

US component of the integrated γ -US probe against 1.11 mm for the standalone US probe, respectively, at a target depth of 5 cm from the probe. Table II summarizes the results of the US performance of both the standalone and integrated probes.

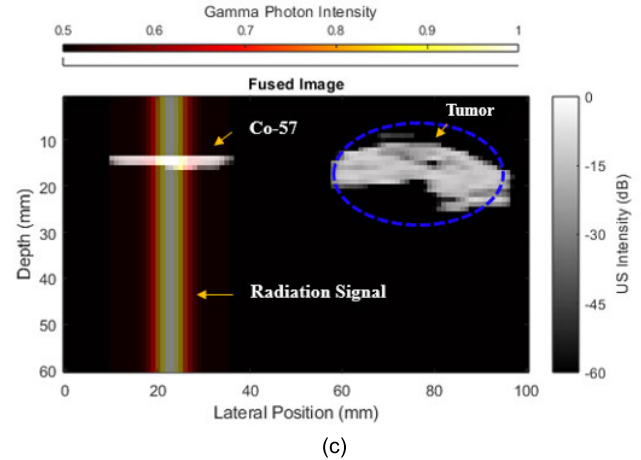
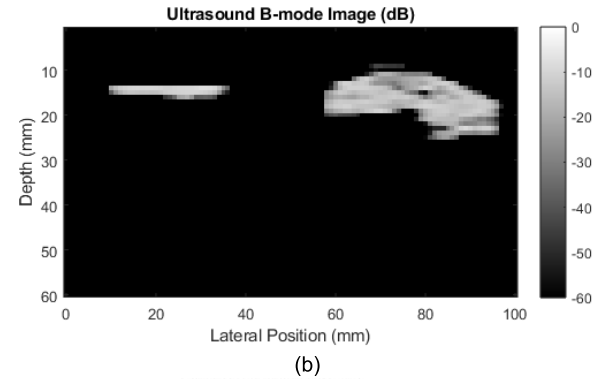
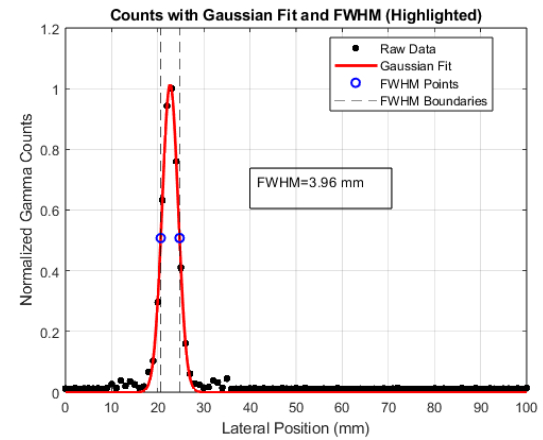


Fig. 13. (a) γ -intensity profile reveals the radiation source (^{57}Co). (b) B-scan US image shows both targets (^{57}Co) source and a round tumor mimicking structure. (c) Fused image combines γ -photon and US data, clearly differentiating the radiation source from the tumor.

C. Frequency-Based Multiplexed-Readout Method Assessment

The FWHM of the proposed frequency-based multiplexing readout method was assessed and compared with previous switching method [9]. The dedicated readout and switching methods yielded energy resolutions of 19.2% and 20.7%, respectively, while the proposed multiplexing method achieved an energy resolution of 19.5%. Similarly, regarding CPS performance, negligible differences were observed between the dedicated readout (90.0 CPS/MBq) and the proposed frequency-based multiplexed readout method (89.7 CPS/MBq). The switching method [9] on the other hand

showed a significant drop in CPS, introduced by PRF of the US transmission—higher PRF shows reduced γ counts. At 1 kHz PRF, the switching method showed $\approx 18\%$ CPS drop (72.5 CPS/MBq versus 90 CPS/MBq). Table III provides a summary of results.

To evaluate the US signal performance of the proposed frequency-based multiplexed readout, two key performance parameters of the US probe, including the SNR and FBW at -6 dB. Performance was compared across three front-end readout configurations: 1) a custom-designed dedicated US readout circuit; 2) a previously reported switching method [9]; and 3) the proposed frequency-based multiplexed method. The dedicated circuit achieved an FBW of 35.7%, the multiplexed method 33.3%, and the switching method 32.0%. In terms of SNR, the proposed multiplexed approach outperformed the others, achieving 22.98 dB, compared to 14.95 dB for the dedicated circuit and 10.68 dB for the switching method. A detailed summary is provided in Table III.

D. Imaging Feasibility Validation

The B-mode US image and the γ -photon profile were acquired using the integrated γ -US probe detector (Fig. 13). The γ -photon profile [Fig. 13(a)] reveals only the radiation source, whereas the B-mode image [Fig. 13(b)] displays both targets but does not provide information about the γ -photons, whereas the fused image [Fig. 13(c)], which combines γ -photon and US data, clearly differentiates the radiation source from the tumor-mimicking target and allows for the measurement of the thickness of both materials. This demonstrates the complementary abilities of the integrated γ -US probe detector in providing both anatomical and molecular information.

V. DISCUSSION

This proof-of-concept study successfully demonstrates the feasibility of a handheld integrated γ -US probe detector, combining γ and US modalities. The proposed integrated probe is designed for low-to-moderate targeted γ energies (^{99m}Tc at 140 keV or ^{57}Co at 122 keV) along with integration of a 10 MHz PZT transducer, which could be well-suited for high-resolution and short-ranging imaging medical applications such as intraoperative procedures (breast tumors or lymph node localization) [6] and nonmedical (NDT [16]). Performance comparisons with standalone γ and US probes show that integration of these two modalities can be achieved with minimal degradation, thus validating the practical potential of this integrated technology.

Hybrid imaging and nonimaging probes, combining different modalities, have gained considerable attention for their potential to enhance diagnostic and therapeutic outcomes. For instance, hybrid PET/CT [28] and PET/MRI [11] systems have revolutionized cancer diagnostics by providing both metabolic and anatomical information, improving tumor detection and characterization. Similarly, hybrid γ -optical probes have been developed for real-time sentinel lymph node detection during surgery [29]. These multimodal systems leverages complementary strengths of each modality, such as the high sensitivity of γ detection for functional imaging and the superior spatial resolution of US for anatomical imaging [30].

The γ -detection performance of the integrated γ -US probe was closely compared to that of a standalone γ -probe by placing the ^{57}Co at 1 cm from the probe. Energy resolution (FWHM) of a radiation detector is crucial for intraoperative probes and NDT because it enhances the precision of tumor localization and defect identification in materials by improving contrast and suppressing signals from scattered radiation, thereby reducing the risk of false positives in healthy tissues. Table II shows energy resolution results for both integrated and standalone detectors [17.3% versus 16.9% (FWHM)]. Similarly, sensitivity directly impacts the administered radiation dose, count rate, and scan time of a γ detector. Table I shows that the integrated probe had a sensitivity of 80.0 c/s/MBq as compared to 90.0 c/s/MBq. Spatial resolution is a very critical performance measure as it increases the precision of a radiation source localization, improves detection accuracy, and ensures effective identification of small or closely spaced targets. The integrated probe achieved a spatial resolution of 2.69 mm, compared to 2.32 mm for the standalone γ -probe—indicating only minor trade-offs in performance. Furthermore, the collimator's shielding rate signifies the probe's capacity to block γ -photons from entering its sides, thus minimizing background noise and false positive counts. The integrated probe showed a side-shielding rate of 95.9% as compared to 97.1% of the standalone γ probe.

Similarly, the US component was benchmarked against a standalone US probe, both using non-focused transducers. The integrated probe achieved an SNR of 5.95 versus 6.49 dB for the standalone unit. Axial and lateral resolutions at a 5 cm depth were 1.32 and 4.90 mm (FWHM), respectively, compared to 1.11 and 4.48 mm for the standalone probe. These results confirm that the integrated γ -US detector maintains competitive performance with minimal degradation, validating its suitability for practical applications. While resolution could be further enhanced using a focused US probe, this study primarily aimed to demonstrate feasibility and performance comparability.

The proposed γ -US frequency-based multiplexing method enables true simultaneous acquisition of γ and US signals through a single channel by summing them with a simple OP-AMP, eliminating the need for multiple ADCs and reducing hardware complexity, size, and cost. It achieves an energy resolution of 19.5%, compared to 19.2% from a dedicated γ readout—and maintains CPS performance. The US component experiences only a slight reduction in bandwidth (32% versus 36% at -6 dB) with minimal effect on signal quality, while delivering a significantly improved SNR of 23 dB (versus 15 dB) due to effective noise suppression via bandpass filtering. Unlike switching-based methods, which suffer CPS loss at high PRFs, this approach supports simultaneous acquisition without degradation. Though it requires postprocessing (LPF/BPF), these steps are easily managed in software, making the method a robust, efficient, and cost-effective solution for integrated γ -US imaging.

The slight performance degradation observed in the integrated γ -US probe is primarily due to the use of a dual-functional tungsten-epoxy compound as both the US backing material and the γ -photon collimator. Accommodating the collimator hole, scintillator, and SiPM required

a reduction in the active backing area. Additionally, while the PZT crystal and matching layer introduced minimal γ -photon attenuation, they demonstrated effective integration by supporting both modalities along a shared LOS. Although the impact on γ -detection performance was minor, further optimization could involve using a ring-type PZT transducer, as proposed in [31]. However, this design introduces new challenges, such as dead zones in US signal reception due to the central hole and the need for postalignment and calibration to co-register data, given the differing LOS between the US and γ components.

Despite the noted challenges, this study highlights the strong potential of the integrated γ -US probe detector. Utilizing a shared LOS for both modalities simplify the design and enhances usability. While the single-element transducer was appropriate for this proof-of-concept, it is suboptimal for clinical imaging. Future work will focus on developing a multielement integrated γ -US probe to improve imaging performance. Furthermore, the compact design and efficient operation of both the probe and the frequency-based multiplexed readout system make the platform well suited for intraoperative procedures such as sentinel lymph node biopsy and tumor localization [32]. These applications benefit from the simultaneous acquisition of real-time anatomical and functional data, overcoming the limitations of larger conventional systems. Additionally, the probe's versatility extends to NDT in industrial settings, demonstrating its broader applicability beyond clinical use [33].

However, the study also highlights certain limitations. Performance validation was confined to controlled experimental settings, necessitating further validation through comprehensive clinical extensive studies. Also, the lateral resolution of the integrated probe is very low, that is, 4.48 mm for standalone probe and 4.90 mm for the integrated probe at 5 cm. This is due to a broad unfocused beam and does not apply a lens or beamforming. Future work will involve fabricating a focused US transducer to leverage its advantages: first, improved lateral resolution by concentrating acoustic energy into a narrow beam at a defined focal point, enabling more precise differentiation of deep tissue structures; second, enhanced SNR for better image quality and penetration; and third, suppression of off-axis scattering and side lobes to reduce clutter and improve target localization. This transducer will then be integrated with the gamma (γ) system, followed by clinical applicability studies using clinically relevant phantoms.

This proof-of-concept study lays a strong foundation for future efforts to develop an integrated γ -US hybrid modality. By demonstrating that minimal performance degradation occurs despite the integration of two modalities, this study paves the way for more sophisticated multimodal probe solutions. The next steps will involve transitioning from a single-element prototype to a fully functional imaging ensemble with an array of integrated γ -US elements, ultimately aiming to enhance diagnostic capabilities and expand the practical applications of this technology.

In summary, the integrated γ -US probe detector and single-channel multiplexed readout represent a significant advancement in multimodal imaging for both medical

and industrial applications. The frequency-based multiplexing approach enables simultaneous acquisition of γ and US signals using shared hardware, reducing system complexity and size. While the use of a low-bandwidth amplifier aids signal separation, it may limit γ resolution at high count rates; however, experimental results showed no significant interference between modalities. The proposed 10 MHz integrated γ -US probe offers high spatial resolution with limited penetration (~ 4 – 5 cm), making it ideal for superficial imaging. Future work will explore alternative frequency schemes and adaptive strategies to broaden the system's capabilities. This study not only confirms the feasibility of integrated γ -US detection but also lays the groundwork for next-generation intraoperative and NDT solutions, marking a key milestone in dual-modality probe development.

VI. CONCLUSION

The development of the integrated γ -US probe detector marks a significant advancement in multimodal detector technology. This proof-of-concept study demonstrates that combining γ and US modalities into a single handheld device is feasible with minimal performance trade-offs. The probe's compact form factor and efficient operation are particularly advantageous for intraoperative applications such as sentinel lymph node biopsies and tumor localization, as well as NDT in industrial settings. Despite minor performance degradation and the need for further validation, this integrated probe lays a strong foundation for future innovations in multimodal imaging technology, promising enhanced diagnostic accuracy and versatility.

REFERENCES

- [1] A. Gonzalez-Montoro et al., "Nuclear-medicine probes: Where we are and where we are going," *Med. Phys.*, vol. 49, no. 7, pp. 4372–4390, 2022.
- [2] A. Wong, J. C. Y. Wong, P. U. Pandey, and S. M. Wiseman, "Novel techniques for intraoperative parathyroid gland identification: A comprehensive review," *Expert Rev. Endocrinol. Metabolism*, vol. 15, no. 6, pp. 439–457, Nov. 2020.
- [3] L. Dixon, A. Lim, M. Grech-Sollars, D. Nandi, and S. Camp, "Intraoperative ultrasound in brain tumor surgery: A review and implementation guide," *Neurosurgical Rev.*, vol. 45, no. 4, pp. 2503–2515, Aug. 2022.
- [4] L. J. Lauwerends et al., "The complementary value of intraoperative fluorescence imaging and Raman spectroscopy for cancer surgery: Combining the incompatibles," *Eur. J. Nucl. Med. Mol. Imag.*, vol. 49, no. 7, pp. 2364–2376, Jun. 2022.
- [5] V. J. Reid, J. S. Falk, A. M. Police, C. A. Ridgeway, L. L. Cadena, and S. P. Povoski, "Minimizing re-excision after breast conserving surgery—A review of radiofrequency spectroscopy for real-time, intraoperative margin assessment," *Expert Rev. Med. Devices*, vol. 18, no. 11, pp. 1057–1068, Nov. 2021.
- [6] F. W. B. van Leeuwen and A. Chiti, "Intraoperative molecular imaging—A means to strengthen the connection between nuclear medicine and surgical disciplines," *Eur. J. Nucl. Med. Mol. Imag.*, vol. 51, no. 10, pp. 2853–2855, Aug. 2024.
- [7] M. S. De Feo, V. Frantellizzi, L. De Sio, A. Farcomeni, G. De Vincentis, and R. Pani, "Role of PET gamma detection in radioguided surgery: A systematic review," *Clin. Transl. Imag.*, vol. 11, no. 6, pp. 567–585, Apr. 2023.
- [8] S. Kaviani et al., "Design and development of a dedicated portable gamma camera system for intra-operative imaging," *Phys. Medica*, vol. 32, no. 7, pp. 889–897, Jul. 2016.
- [9] M. N. Ullah et al., "Simultaneous acquisition of ultrasound and gamma signals with a single-channel readout," *Sensors*, vol. 21, no. 4, p. 1048, Feb. 2021.

- [10] H. G. Kang, H.-Y. Lee, K. M. Kim, S.-H. Song, G. C. Hong, and S. J. Hong, "A feasibility study of an integrated NIR/gamma/visible imaging system for endoscopic sentinel lymph node mapping," *Med. Phys.*, vol. 44, no. 1, pp. 227–239, Jan. 2017.
- [11] K. Kulesz, "GAMMA-MRI: Towards a novel clinical imaging modality using long-lived aligned gamma emitters," Dept. Nucl. Corpuscular Phys., Fac. Sci., Univ. Geneva, Geneva, Switzerland, Tech. Rep. CERN-THESIS-2023-197, Sep. 2023.
- [12] A. M. Almarhaby et al., "Characterisation of a near-infrared (NIR) fluorescence imaging systems intended for hybrid gamma-NIR fluorescence image guided surgery," *J. Instrum.*, vol. 14, no. 7, Jul. 2019, Art. no. P07007.
- [13] R. De Luca, L. Forzoni, F. Gelli, and J. Bamber, "An educational overview of ultrasound probe types and their fields of application," *Arch. Acoust.*, vol. 2021, pp. 3–15, Mar. 2021.
- [14] T. Wendler et al., "Real-time fusion of ultrasound and gamma probe for navigated localization of liver metastases," in *Proc. Int. Conf. Med. Image Comput. Comput.-Assist. Intervent.*, 2007, pp. 252–260.
- [15] S. K. Dwivedi, M. Vishwakarma, and P. A. Soni, "Advances and researches on non destructive testing: A review," *Mater. Today, Proc.*, vol. 5, no. 2, pp. 3690–3698, 2018.
- [16] J. Provost et al., "Simultaneous positron emission tomography and ultrafast ultrasound for hybrid molecular, anatomical and functional imaging," *Nature Biomed. Eng.*, vol. 2, no. 2, pp. 85–94, Feb. 2018.
- [17] C. Polito et al., "Dual-modality imaging with a ultrasound-gamma device for oncology," *Radiat. Phys. Chem.*, vol. 147, pp. 77–84, Jun. 2018.
- [18] C. Park, M. N. Ullah, C. Kim, S. Cho, and J.-Y. Yeom, "Investigation of optical properties of ceramic ce: GAGG by high temperature annealing," *J. Korean Phys. Soc.*, vol. 75, no. 12, pp. 962–967, Dec. 2019.
- [19] M. N. Ullah, J. H. Park, E. Pratiwi, G. B. Kim, and J.-Y. Yeom, "Wavelength discrimination (WLD) detector optimization for time-of-flight positron emission tomography with depth of interaction information," *Nucl. Instrum. Methods Phys. Res. A, Accel. Spectrom. Detect. Assoc. Equip.*, vol. 982, Dec. 2020, Art. no. 164498.
- [20] M. N. Ullah, E. Pratiwi, J. H. Park, K. Lee, H. Choi, and J.-Y. Yeom, "Wavelength discrimination (WLD) TOF-PET detector with DOI information," *Phys. Med. Biol.*, vol. 65, no. 5, Feb. 2020, Art. no. 055003.
- [21] A. N. H. Al-Ajili, S. M. Aman Allah, R. A. Abad, and A. M. O. Al-Bayti, "Stopping power for tungsten," *Tikrit J. Pure Sci.*, vol. 21, no. 1, pp. 82–86, Feb. 2023.
- [22] A. Duatti, "Review on 99mTc radiopharmaceuticals with emphasis on new advancements," *Nucl. Med. Biol.*, vol. 92, pp. 202–216, Jan. 2021.
- [23] V. T. Rathod, "A review of acoustic impedance matching techniques for piezoelectric sensors and transducers," *Sensors*, vol. 20, no. 14, p. 4051, Jul. 2020.
- [24] V. M. do Nascimento, V. L. da Silveira Nantes Button, J. M. Maia, E. T. Costa, and E. J. V. Oliveira, "Influence of backing and matching layers in ultrasound transducer performance," *Proc. SPIE*, vol. 5035, pp. 86–96, May 2003.
- [25] J. Y. Yeom, M. N. Ullah, and C. H. Kim, "Handheld ultrasound and nuclear medicine fusion imaging device," U.S. Patent 11 937 978 B2, Mar. 26, 2024.
- [26] S. H. Hesari, M. A. Haque, and N. McFarlane, "A comprehensive survey of readout strategies for SiPMs used in nuclear imaging systems," *Photonics*, vol. 8, no. 7, p. 266, 2021.
- [27] T. L. Szabo, *Diagnostic Ultrasound Imaging: Inside Out*. New York, NY, USA: Academic, 2013.
- [28] C. Peştean, A. Pavel, and D. Piciu, "The role of SPECT/CT and PET/CT hybrid imaging in the management of ectopic thyroid carcinoma—A systematic review," *Diagnostics*, vol. 14, no. 13, p. 1369, Jun. 2024.
- [29] A. L. Farnworth and S. L. Bugby, "Intraoperative gamma cameras: A review of development in the last decade and future outlook," *J. Imag.*, vol. 9, no. 5, p. 102, May 2023.
- [30] M. Wu and J. Shu, "Multimodal molecular imaging: Current status and future directions," *Contrast Media Mol. Imag.*, vol. 2018, pp. 1–12, Jun. 2018.
- [31] C.-C. Shih, C.-C. Huang, Q. Zhou, and K. K. Shung, "High-resolution acoustic-radiation-force-impulse imaging for assessing corneal sclerosis," *IEEE Trans. Med. Imag.*, vol. 32, no. 7, pp. 1316–1324, Jul. 2013.
- [32] D. Verver et al., "Gamma probe and ultrasound-guided fine needle aspiration cytology of the sentinel node (GULF) trial," *Eur. J. Nucl. Med. Mol. Imag.*, vol. 45, no. 11, pp. 1926–1933, Oct. 2018.
- [33] M. I. Silva, E. Malitckii, T. G. Santos, and P. Vilaça, "Review of conventional and advanced non-destructive testing techniques for detection and characterization of small-scale defects," *Prog. Mater. Sci.*, vol. 138, Sep. 2023, Art. no. 101155.



Muhammad Nasir Ullah received the B.S. degree in electronic engineering from International Islamic University, Islamabad (IIUI), Islamabad, Pakistan, in 2012, and the Integrated M.S./Ph.D. degree in bio-convergence engineering from Korea University, Seoul, Republic of Korea, in 2020.

He is currently working as a Physical Science Research Scientist with the Department of Radiology, Stanford University, Stanford, CA, USA.



Muhammad Waqar (Graduate Student Member, IEEE) received the B.S. degree in electronic engineering from the International Islamic University, Islamabad (IIUI), Islamabad, Pakistan, in 2012, and the M.S. degree in biomedical engineering from the National University of Sciences and Technology (NUST), Islamabad, in 2016. He is currently pursuing the Ph.D. degree in bioengineering with Korea University, Seoul, Republic of Korea.



Yuseung Park received the B.S. degree in biomedical engineering and the M.S. degree in bioengineering from Korea University, Seoul, Republic of Korea, in 2020 and 2022, respectively.

Currently, he is with Samsung Electronics, Suwon, South Korea.



Gyeong Beom Kim received the B.S. degree in biomedical engineering and the M.S. degree in bioengineering from Korea University, Seoul, Republic of Korea, in 2020 and 2022, respectively.

Currently, he is with Samsung Electronics, Suwon, South Korea.



Kyungseok Choi received the B.S. degree in biomedical engineering and the M.S. degree in bioengineering from Korea University, Seoul, Republic of Korea, in 2022 and 2024, respectively.

Currently, he is with Samsung Electronics, Suwon, South Korea.



Seungjoo Jang received the B.S. degree in biomedical engineering and the M.S. degree in bioengineering from Korea University, Seoul, Republic of Korea, in 2023 and 2025, respectively.

Currently, he is with Samsung Electronics, Suwon, South Korea.



Hojong Choi received the B.E. degree in electronic engineering from Kyung Hee University, Seoul, South Korea, in 2000, and the M.S. and Ph.D. degrees in electrical engineering from the University of Southern California, Los Angeles, CA, USA, in 2004, and 2012, respectively.

He is currently working as an Associate Professor with the Department of Electronic Engineering, Gachon University, Seongnam, South Korea.



Chanyoung Kim (Graduate Student Member, IEEE) received the B.S. degree in biomedical engineering from Korea University, Seoul, Republic of Korea, in 2025, where he is currently the M.S. degree with the Department of Bioengineering.



Qifa Zhou (Fellow, IEEE) received the Ph.D. degree from Xi'an Jiaotong University, Xi'an, China, in 1993.

He completed Postdoctoral training with The Hong Kong Polytechnic University, Hong Kong, and Penn State, University Park, PA, USA. He is currently a Professor with the University of Southern California, Los Angeles, CA, USA, and the Zohrab A. Kaprielian Fellow, with research focused on high-frequency ultrasound, photoacoustic, and multimodal imaging. He has

authored over 320 articles, holds more than ten patents.

Dr. Zhou is a fellow of SPIE, AIMBE, IAAM, and a Senior Member of the National Academy of Inventors. He has received multiple awards. He serves as an Associate Editor for IEEE TRANSACTIONS ON ULTRASONICS, FERROELECTRICS, AND FREQUENCY CONTROL and *BME Frontiers*.



Jung-Yeol Yeom (Member, IEEE) received the B.S. degree in nuclear engineering from Seoul National University, Seoul, South Korea, in 2001, and the M.S. and Ph.D. degrees in quantum engineering and systems science from The University of Tokyo, Tokyo, Japan, in 2003 and 2006, respectively, focusing on electronics and detectors for positron emission tomography.

Since 2015, he has been a Professor with the School of Biomedical Engineering, Korea University, Seoul. His research covers detector and instrumentation development for medical applications—including nuclear medicine, ultrasound, and multimodal imaging—as well as nonmedical fields such as radiation monitoring and homeland security.