

Detecting Intracranial Hemorrhage via Wearable Microwave Imaging: Ex Vivo Human Study

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Abstract — This article presents an experimental validation of a wearable microwave imaging (MWI) headband for the detection of intracranial hemorrhage. The system employs a dual-ring array of 32 multiplexed antennas via a MEMS switching matrix, acquiring 128 bistatic measurements per scan. Through a differential imaging protocol, common-mode clutter suppression and symmetric artifact filtering isolate the target scattering. The data are then reconstructed using Delay-Multiply-and-Sum (DMAS) beamforming. The system was evaluated on three human cadaver heads with simulated hemorrhagic targets of 15 mL and 30 mL, representing established operative thresholds. Spatial co-registration with optical photographs yielded a mean localization error of 14.7 mm, with four of six scenarios below 20 mm. To the authors' knowledge, this represents one of the few experimental validations of a wearable MWI system on human cadavers, advancing MWI toward clinical translation.

Keywords — beamforming, cadaver, DMAS, intracranial hemorrhage, microwave imaging, wearable.

I. INTRODUCTION

Intracranial imaging is essential for diagnosing neurological emergencies, such as stroke, traumatic brain injury, and intracranial hemorrhage, where rapid assessment dictates clinical outcomes. While Computed Tomography (CT) and Magnetic Resonance Imaging (MRI) remain the clinical gold standards, their reliance on hospital-based infrastructure, ionizing radiation (CT), and lengthy acquisition times (MRI) preclude continuous point-of-care or pre-hospital monitoring. These limitations necessitate the development of portable, rapid, and safe diagnostic alternatives.

Microwave imaging (MWI) has emerged as a promising non-ionizing modality that maps the dielectric properties of biological tissues. Because hemorrhage and ischemia alter local tissue water content, they generate measurable dielectric contrasts against healthy brain parenchyma [1], [2]. Over the past two decades, various MWI systems have been proposed for brain diagnostics, including helmet-like arrays and tomographic platforms [3], [4], [5], [6]. Clinical evaluation of microwave-based intracranial hemorrhage detection has also been reported [7]. Alongside hardware advancements, reconstruction techniques have evolved from radar-based beamforming and inverse-scattering algorithms to data-driven machine learning approaches [8], [9], [10].

Despite these advancements, experimental validation remains largely confined to numerical simulations, simplified phantoms, and *ex vivo* animal models [11], [3], [5], [4], [6], which cannot reproduce the electromagnetic attenuation and anatomical complexity of the human skull. Validation under realistic anatomical conditions is therefore a critical prerequisite for clinical translation.

To bridge this translational gap, this paper presents a preliminary feasibility assessment of a wearable, 32-element microwave imaging headband designed for rapid intracranial monitoring. The system employs a differential imaging protocol to isolate pathological scattering from static anatomical clutter. The device is evaluated on human cadaveric heads, demonstrating detection and localization of simulated acute intracranial hemorrhages within a realistic anatomical environment.

II. SYSTEM DESIGN

A. Antenna Array Design

The antenna array comprises 32 elements arranged in two vertically stacked rings of 16 antennas each, integrated into a wearable headband. The two rings are separated axially by a distance of 25 mm and are azimuthally aligned around the circumference of the head.

Each transceiver antenna block houses two vertically adjacent elements, for a total of 16 blocks around the headband. The individual elements are modified slotted bowtie patch antennas fabricated on a 0.9 mm thick FR4 substrate, identical to the design in [12]. Antenna-to-head coupling is ensured by a graphite-loaded RTK silicone matching layer fabricated following the procedure in [13]. For mechanical robustness, the 16 blocks are interconnected via flexible meshes housing the coaxial feed lines to the RF front-end.

B. RF Front-End and Signal Acquisition

The RF signal chain comprises a LibreVNA Vector Network Analyzer (VNA) and a custom MEMS switching matrix multiplexing the 32 elements to the VNA's two ports. All elements support both transmission and reception. Two complementary multiplexer banks route even- and odd-indexed

antennas within each ring to separate VNA ports, yielding $8 \times 8 = 64$ bistatic ($S_{ij}, i \neq j$) measurements per ring. TX/RX roles are complementary across rings: within each block, the upper element transmits while the lower receives, or vice versa. Combining both rings yields 128 bistatic measurements per scan, providing spatial diversity at two axial heights.

Each S_{ij} spans 0.5–3.0 GHz over 101 frequency points (25 MHz step). Prior to cadaver imaging, the system was Open-Short-Load (OSL) calibrated up to each antenna feeder point.

III. EX-VIVO EXPERIMENT

A. Specimens and Target Preparation

Three fresh-frozen human cadaver heads, preserved by immersion in an alcohol-based fixative solution, were obtained from the Trecchi Human Laboratory (Cremona, Italy). Specimens were from adult donors with no documented neurological pathology. Surgical access was achieved via craniotomy performed by experienced medical staff. The antenna headband was secured to each specimen using an elastic retention band, with its center height aligned with the maximum head circumference plane as defined in the ANSUR II anthropometric database [14].

To simulate an acute intracranial hemorrhage, the balloon of a 2-way 30 mL Foley catheter (Fig. 1a) was introduced through the surgical access into the intracranial space (Fig. 1b) and infused with a blood-mimicking fluid. For each specimen, two target volumes were tested sequentially at the exact same insertion site: 15 mL and 30 mL. These specific volumes were selected to accurately reflect the critical clinical thresholds for operative intervention established in neurosurgical guidelines [15], [16], [17]. The antenna headband array was not repositioned between acquisitions, ensuring precise differential subtraction and spatial co-registration across all scans.

B. Measurement Protocol

The experiments were conducted in May 2025. For each specimen, a differential imaging protocol was applied: (i) a *baseline scan* of the craniotomized specimen prior to target insertion, (ii) insertion of the simulated hemorrhagic target, and (iii) a *target-present scan* acquired immediately after placement. The craniotomies were pre-existing from prior anatomical procedures; the intracranial structures remained intact. The differential signal, obtained by subtracting the baseline from the target-present scan, isolates the target scattering while suppressing static clutter from the skull, skin, and surrounding tissues. Two target volumes were tested per specimen (15 mL and 30 mL), yielding six experimental scenarios.

For optical calibration, a retro-reflective target with six reference points was placed adjacent to the specimen. An optical camera captured photographs at each protocol stage. A planar homography, computed from reference-point correspondences, mapped image coordinates to the physical frame; the same transform was applied to dual retro-reflective

markers on each antenna block to derive antenna positions for beamforming. This pipeline also enabled co-registration of reconstructed images with optical photographs for ground truth estimation (Section IV-B). Thermal stability was monitored via a calibrated thermal camera (40 mK sensitivity) to ensure accurate differential background cancellation.

C. Signal Preprocessing

Despite differential subtraction, residual clutter from imperfect cancellation, antenna coupling, and structural reflections persists. Each channel corresponds to one bistatic antenna pair (i, j) . A three-stage preprocessing pipeline is applied: (1) the mean over frequency is subtracted from each channel to remove the DC offset; (2) the per-frequency average across all channels is subtracted from each channel, suppressing common-mode components from antenna coupling and shared anatomical reflections; (3) each channel is subtracted from its diametrically opposite counterpart, removing symmetric artifacts while preserving asymmetric scattering from localized targets.

IV. IMAGING AND RESULTS

A. Beamforming Algorithm

Image reconstruction is performed using the Delay-Multiply-and-Sum (DMAS) beamforming algorithm [18]. Let $S_i(f)$ represent the preprocessed signal for the i -th bistatic channel at frequency f , where $i \in \{1, \dots, M\}$. For focal point $\mathbf{r}$, the phase-compensated signal is $Y_i(\mathbf{r}, f) = S_i(f) \cdot e^{-j2\pi f \tau_i(\mathbf{r})}$, where $\tau_i(\mathbf{r})$ is the two-way delay assuming a homogeneous medium with relative permittivity $\epsilon_r \approx 15$, estimated from the expected dielectric properties of the heterogeneous propagation path through skull bone and preserved cerebral tissue.

The DMAS image intensity at each focal point is obtained by summing the pairwise cross-correlations of the phase-compensated signals over frequency:

$$I_{\text{DMAS}}(\mathbf{r}) = \sum_{i=1}^{M-1} \sum_{j=i+1}^M \text{Re} \left\{ \sum_f Y_i(\mathbf{r}, f) \cdot Y_j^*(\mathbf{r}, f) \right\} \quad (1)$$

Pairwise multiplication rejects clutter—uncorrelated signals cancel while coherent returns reinforce—yielding sharper focal spots than delay-and-sum.

Signals from both rings are reconstructed over a three-dimensional ellipsoidal volume spanning the vertical extent of the antenna array, conforming to each head's cross-section. Axial slices with optimal focal quality were selected for evaluation.

B. Visual Validation

Detection performance was evaluated both visually, via co-registration of reconstructed images with optical photographs (Figs. 1c and 2), and quantitatively, by computing the localization error Δ (Table 1). Using the calibration pipeline of Section III, photographs were rectified into the physical coordinate system. The ground truth position was

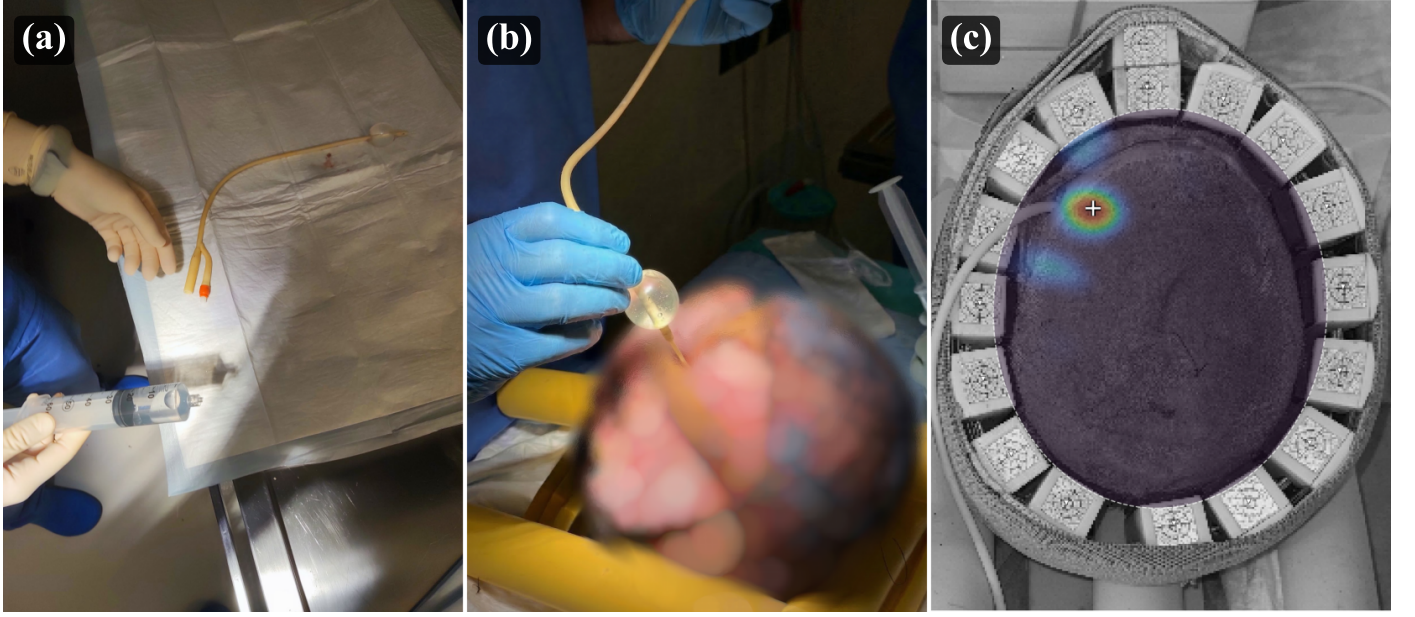


Fig. 1. Experimental setup and representative imaging result. (a) Two-way Foley catheter with balloon inflated to 30 mL using a graduated syringe, simulating an intracranial hemorrhagic target. (b) Insertion of the inflated balloon through the craniotomy access into the intracranial space of a cadaveric specimen. (c) Reconstructed MWI image overlaid on the rectified optical photograph of Specimen 2 (30 mL target); the white crosshair indicates the optically estimated ground truth position.

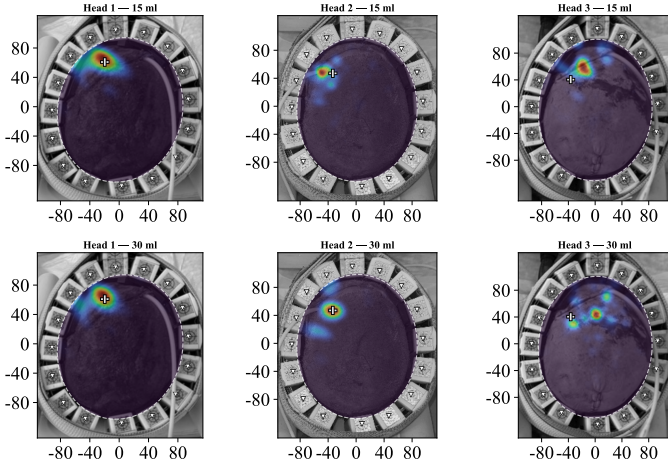


Fig. 2. Differential MWI reconstructions overlaid on rectified optical photographs for all six scenarios (rows: specimens; columns: target volumes). The turbo-coloured heatmap shows the reconstructed DMAS intensity within the imaging domain (dashed ellipse). Antenna positions are shown as white triangles; white crosses indicate optically estimated ground truth positions. Axes in mm relative to head centre.

estimated as the centroid of the surgical access site in the rectified photographs; given this indirect estimation, reported localization errors should be interpreted as approximate. The localization error $\Delta = \|\mathbf{r}_{\text{peak}} - \mathbf{r}_{\text{GT}}\|$ is the Euclidean distance between the reconstructed peak and the estimated target position. The Signal-to-Clutter Ratio (SCR) quantifies anomaly contrast relative to background clutter.

Table 1. Quantitative localization results. Δx and Δy : displacement of the reconstructed peak from the optically estimated ground truth position. Δ : Euclidean localization error. SCR: Signal-to-Clutter Ratio.

Sp.	Vol. (mL)	Δx (mm)	Δy (mm)	Δ (mm)	SCR (dB)
1	15	-2.7	-3.3	4.3	2.6
	30	-2.6	-3.3	4.2	3.7
2	15	-14.9	-0.5	14.9	5.4
	30	-1.0	1.3	1.7	5.5
3	15	18.8	-17.1	25.4	4.6
	30	37.7	-3.0	37.8	2.4

C. Detection Results

Table 1 summarizes the localization results across all six scenarios. The mean localization error is $\Delta = 14.7$ mm, with four of six scenarios yielding $\Delta < 20$ mm.

Specimens 1 and 2 show Δ from 1.7 mm to 14.9 mm with positive SCR in all four cases (Fig. 2). Specimen 3 yields the largest errors ($\Delta = 25.4$ mm and 37.8 mm), with competing focal spots indicating incomplete clutter suppression for this geometry.

V. DISCUSSION AND CONCLUSION

The unified imaging protocol, combining DMAS beamforming with common-mode and symmetric artifact suppression, achieved a mean localization error of 14.7 mm across six experimental scenarios on three human cadaveric heads (Table 1), with four of six scenarios below 20 mm. The degraded performance on Specimen 3 (both volumes) may be attributed to the specific craniotomy geometry and target

depth, which introduced stronger anatomical clutter that the preprocessing pipeline could not fully suppress.

Several limitations must be acknowledged. The hardware remains a prototype; antenna integration, switching matrix, and calibration are not yet optimized for clinical use. The sample size of three specimens precludes statistical generalization. Ground truth positions were optically approximated from the surgical access site, introducing spatial uncertainties. Finally, the dielectric properties of preserved tissue differ from *in vivo* conditions, potentially affecting scattering intensities.

Despite these limitations, the results demonstrate that wearable MWI-based hemorrhage detection is feasible under realistic anatomical conditions. To address hardware limitations, a second cadaver experiment with an industrialized prototype featuring improved antenna-skin coupling and sensor-based automatic antenna localization is planned, alongside an expanded specimen cohort to support statistical validation. The current differential protocol requires a patient-specific baseline scan, limiting applicability in acute clinical scenarios. To overcome this, we are investigating deep-learning-based reconstruction that operates on absolute (non-differential) data, eliminating baseline dependence. Complementary approaches include contralateral hemisphere comparison and model-based synthetic reference generation. Ground truth accuracy will be improved through CT-based target co-registration, and future *in vivo* trials will address tissue-property discrepancies inherent to preserved specimens. More accurate propagation models, including layered tissue models and physics-informed data-driven approaches, are also being developed to overcome the simplified homogeneous assumption.

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