3D parallel-detection microwave tomography for clinical breast imaging

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A biomedical microwave tomography system with 3D-imaging capabilities has been constructed and translated to the clinic. Updates to the hardware and reconfiguration of the electronic-network layouts in a more compartmentalized construct have streamlined system packaging. Upgrades to the data acquisition and microwave components have increased data-acquisition speeds and improved system performance. By incorporating analog-to-digital boards that accommodate the linear amplification and dynamic-range coverage our system requires, a complete set of data (for a fixed array position at a single frequency) is now acquired in 5.8 s. Replacement of key components (e.g., switches and power dividers) by devices with improved operational bandwidths has enhanced system response over a wider frequency range. High-integrity, low-power signals are routinely measured down to -130 dBm for frequencies ranging from 500 to 2300 MHz. Adequate inter-channel isolation has been maintained, and a dynamic range >110 dB has been achieved for the full operating frequency range (500–2900 MHz). For our primary band of interest, the associated measurement deviations are less than 0.33% and 0.5° for signal amplitude and phase values, respectively. A modified monopole antenna array (composed of two interwoven eight-element sub-arrays), in conjunction with an updated motion-control system capable of independently moving the sub-arrays to various in-plane and cross-plane positions within the illumination chamber, has been configured in the new design for full volumetric data acquisition. Signal-to-noise ratios (SNRs) are more than adequate for all transmit/receive antenna pairs over the full frequency range and for the variety of in-plane and cross-plane configurations. For proximal receivers, in-plane SNRs greater than 80 dB are observed up to 2900 MHz, while cross-plane SNRs greater than 80 dB are seen for 6 cm sub-array spacing (for frequencies up to 1500 MHz). We demonstrate accurate recovery of 3D dielectric property distributions for breast-like phantoms with tumor inclusions utilizing both the in-plane and new cross-plane data.

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I. INTRODUCTION

Breast cancer is a major health problem in the United States (US) and is the second leading cause of cancer deaths among American women.¹ American Cancer Society estimates for 2014 suggest 232 760 newly diagnosed invasive breast-cancer cases with approximately 40 000 associated disease-related deaths.¹ Early detection improves an individual's overall prognosis; however, the conventional, clinically accepted screening and diagnostic modalities have important weaknesses. For instance, x-ray mammography is specific but its sensitivity, especially in the dense breast, is relatively poor.² Ultrasound is only available for distinguishing liquid-filled cysts from tumors.³ Contrast-enhanced magnetic resonance imaging (MRI) has been used diagnostically but suffers from a high false positive rate and is expensive.^{4,5} From a spectrum of emerging technologies, microwave

imaging is particularly intriguing because of the potential dielectric property (DP) contrast that exists between healthy and abnormal breast tissue.^{6–9}

Biomedical microwave imaging has been of interest in the academic community for at least three decades as a result of tissue-specific DPs and associated variations caused by physiological aberrations, induced changes associated with treatments such as hyperthermia, and neoplastic transformations indicative of different cancers.^{10–13} The earliest efforts applied simple-imaging algorithms such as diffraction back projection and Born and Rytov approximations¹⁴ as they required only modest computational resources, which were not available to levels that are currently more routine.^{15,16} Work by Jofre *et al.*¹⁷ represented important and innovative first steps but ultimately fell short of clinical translation—mainly because of an inability to image large high-contrast heterogeneous targets commonly encountered in biomedical imaging.¹⁸

More recent investigations have focused on tomographic^{19–23} and radar^{24–27} approaches. Similar to other emerging imaging-technologies, initial efforts focused

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on numerical simulations as a means of studying the underlying problems that ultimately inform hardware designs without the large expense, time and risk associated with actual implementation. Unfortunately, the majority of these undertakings faced difficulties translating concepts into viable experimental or clinical systems,^{28–30} and many studies have remained primarily within the confines of numerical simulations. Two systems successfully solved the under-appreciated challenge of multipath signals either using algorithmic processing³¹ or suppression through attenuation in a lossy coupling-medium.³²

Problems with mutual coupling of adjacent antenna-array elements have also been reported.³³ Work by Franza *et al.*³⁴ and Meaney *et al.*^{35,36} have extensively studied and addressed this issue. Other confounding problems have arisen from inadequate channel-to-channel isolation within the microwave electronics,³⁷ and limited system dynamic range for imaging dense breast-tissue.^{38,39} Until recently, the signal-detection dynamic ranges for most commercial network analyzers were limited to roughly -100 dBm, which is insufficient for some systems. At Dartmouth College (Hanover, NH, USA), we solved the signal-isolation and dynamic-range issues by fabricating custom microwave electronics, and overcame the multipath and mutual-coupling challenges through a synergistic combination of a lossy coupling-medium and a novel-coupling compensation-scheme.⁴⁰

This paper discusses a third generation of the original imaging system described by Meaney *et al.*,⁴¹ which affords important hardware advances and simplifications that have improved system performance. Sections II–III describe the implemented electronic-network changes and system component advances. The illumination tank, antenna array, and motion-control hardware have also been reconfigured into a low-profile ergonomic module that allows independent movement of antenna-array subsets to arbitrary heights within the tank, thereby enabling acquisition of both in-plane (IP) and cross-plane (XP) transmissions. We present data on current system performance and assess the new XP data in terms of measurement sensitivity. Optimal system-operating parameters are defined using signal-to-noise ratio (SNR) metrics resulting from the analysis. Utilizing this information, reconstructed image sets of a breast-shaped phantom with a tumor-like inclusion are provided, demonstrating the system's operation using the XP data and highlighting its 3D-imaging capabilities for clinical use.

II. HARDWARE DESIGN

The current design retains the parallel-detection scheme introduced in the previous generation,²⁰ with modifications made to the system's radio-frequency switching matrix and local-oscillator (LO) power-division networks. Additionally, the 16-bit analog-to-digital (A/D) conversion board and variable gain amplifier combination employed in the previous system has been replaced by a 24-bit A/D board (that covers the desired dynamic range at all power-level settings). Alone, this component accommodates the linear amplification that our system requires at all frequencies, eliminating time-delays associated with the previous amplification process (including

repeated sampling of the data for individual transmit/receive pairs at different gains and sampling times). The new design incorporates an updated motion-control system capable of independently moving complementary antenna sub-arrays (SAs) within the illumination chamber, allowing collection of both IP and XP signals (whereas the previous system was limited to IP positions only).

The third-generation imaging-system advancements structure into four areas: (a) microwave electronics, (b) liquid-coupled breast interface, (c) motion-control system, and (d) data acquisition and processing, and are detailed in Sections II A–II D, respectively.

A. Microwave electronics

Our first- and second-generation systems successfully employed a signal heterodyning technique for acquiring amplitude and phase measurements. This has been re-implemented in the current design, in addition to the replacement of key components (e.g., power dividers (PDs) and A/D boards) by newer, cost-effective, and commercially available devices, with increased operational bandwidths. This upgrade improved system response over a wider frequency range and increased data-acquisition speeds. Changes to the radio-frequency (rf) and LO network layouts have significantly reduced the system's overall size, allowing construction of an enclosed modularized structure where the electronics, data-acquisition (DAQ) components, power supplies, and liquid-coupled breast interface have been compartmentalized, as pictured in Figure 1.

1. Microwave source and local oscillator network

An Agilent ESG-D series 4432B digital rf signal generator (Palo Alto, California, USA) serves as the system's microwave-signal source. With a frequency range of 250 kHz to 3000 MHz and calibrated variable power-output ranging from $+10$ to -136 dBm, it offers customizable test functions with an adjustable offset or intermediate frequency (IF). Multi-tone capabilities are available and through software, adjustments to nearly every aspect of the digital signal or signal-operating environment are possible.

Compared to the previous system's signal source, which yielded sideband and carrier suppression to -60 dBc over the operational frequency range, the current source yields a modulated signal with sideband and carrier suppression to -80 dBc over the same range. Additionally, coherent LO and I/Q signal outputs are located on the rear of the generator for signal demodulation and lock-in amplifier-based I and Q recovery, where I and Q represent the in-phase and quadrature components of the detected microwave signals, respectively.

The LO signal from the source generator is fed to a PD network consisting of one two-way power divider (Pulsar Microwave Corporation PS2-22-450/10S, Clifton, New Jersey, USA) and two eight-way power dividers (Pulsar Microwave Corporation PS2-22-450/50S), resulting in 16 coherent reference signals, one for each channel of the parallel-detection scheme (Figure 2). The PD's frequency range is 500–6000 MHz with a maximum voltage standing-wave ratio (VSWR) of 1.40. Over our system's operating frequency range

FIG. 1. Photographs of (a) the system's modular configuration highlighting four key compartments: 1 – the patient liquid-coupled breast interface, 2 – the microwave electronic networks, 3 – the DAQ components, 4 – the power supplies, (b) the illumination chamber and coupling-medium reservoir (shown with antennas connected to the electronic networks through a bulkhead partition, and (c) the system covered with a customized patient table and breast aperture as it appears in the clinic.

(500–3000 MHz), the insertion losses are less than 0.6 and 1.5 dB above the theoretical 3.0 and 9.0 dB for the two- and eight-way power dividers, respectively. Additionally, the minimum output channel isolation is 18 and 20 dB for these components.⁴²

Fixed coaxial attenuators placed between the PD stages suppress insertion loss mismatch. To boost the LO network-based output signal strength to levels required for driving the transceiver module mixers (Minicircuits ADE-30), an amplification stage (M/A-COM MAAM020350-A2, Lowell, Massachusetts, USA) has been configured into the network. The mixer's power requirements vary as a function of frequency, and unlike the rf signal's variable-power control, the LO sig-

nal (provided by a reference port on the back of the source generator) lacks adjustable-power control (i.e., the LO signal is output at a constant power-level). To meet the mixer's variable drive-power requirement, a transistor-transistor logic (TTL) controlled digital attenuator (M/A-COM AT-213) has been integrated into the network. Input power-levels have been optimized in an effort to minimize the mixer's conversion loss without increasing higher order harmonics.

2. Transmitting and receiving channels

Due to the system's single-channel transmission/parallel-detection design, fast switches and well-matched electronics

FIG. 2. A schematic of the imaging system highlighting its microwave signal source and local-oscillator network layouts.

FIG. 3. A schematic of an individual transceiver unit, including the electronic components (amplifiers, switches, and a mixer) and their layout, as well as the rf, LO, Antenna, and A/D board system connection ports.

(with high inter-channel isolation) are required to allow each of the 16 channels and their associated antennas to transmit microwave signals independently and sequentially. The bi-level 16-channel switching matrix consists of five broadband single-pole, four-throw (SP4T) switches (M/A-COM SW311), such that individual antennas can be selected to transmit microwave signals while all non-active channels receive the radiated signal information in parallel. The independent transceiver unit for each channel is the core building-block of the microwave-imaging array; it allows antenna elements to operate as either a transmitter or receiver.

Figure 3 shows a schematic of a transceiver unit, including amplifier chambers, single-pole, single-throw (SPST) and single-pole, double-throw (SPDT) switches (M/A-COM SW311 and SW313), and a mixer (Minicircuits ADE-30). The SPDT switch allows the transceiver to operate in either transmit or receive mode while the SPST adds isolation. The two amplifiers (M/A-COM MAAM020350-A2) positioned at the rf and LO ports compensate for insertion losses from the rf switching matrix and the LO PD networks, respectively. By placing two low-noise amplifiers near the beginning of the receiver component cascade (between the SPDT switch and the mixer), we compensate for the anticipated tissue attenuation while maintaining a low system-noise level. Additionally, coplanar waveguide is used for transmission in the circuit layout to further minimize transmission losses and undesired signal modes.

B. Liquid-breast interface

The system's liquid-breast interface consists of a 16-element monopole-antenna array housed in an illumination chamber filled with a coupling medium. The cylindrical tank's breast-aperture incorporates custom-positioning inserts tailored to a wide-range of breast diameters. The lossy (glycerin/water) coupling-medium acts primarily as a matching fluid to limit scatter at the breast boundary,⁴³ but offers the added benefit of reducing multipath signals and debilitating mutual coupling.³² Transferring liquid into and out of the

chamber uses an automated pumping system (Baldor Industrial custom DC isolated pump, Fort Smith, Arizona, USA).

In free-space, the monopole antenna has a narrow bandwidth, a circularly symmetric beam pattern and the potential to excite surface currents along its outer conductor.^{44,45} However, when operating in the lossy coupling-medium its bandwidth broadens due to the resistive loading.⁴⁶ The antennas maintain return losses better than 10 dB over the operating frequency range and the attenuating nature of the fluid reduces surface-current excitation. Despite the monopole's low efficiency, this radiator is fully capable of propagating signals across a breast using our system.

The monopole is made of semi-ridged coaxial cable 2.3 mm in diameter. Forming the antenna's active region requires removing the outer conductor's top 3.4 cm. A protective Delrin sleeve of equal length covers this exposure, and a 4.4 mm diameter stainless-steel rod (SS303) covers the remaining feedline to ensure mechanical rigidity. This casing provides a surface that slides through hydraulic seals in the tank base during antenna motion. The 16 vertically-oriented monopole antennas are positioned in a circular array and organized into two interleaved SA (with alternating elements rigidly connected to independent mounting plates). This dual-mounting arrangement enables our system to collect XP volumetric signal measurements used in 3D image reconstruction.

A series of hydraulic valves facilitates the filling and draining of liquid from the tank. A low-profile optical sensor (Madison OPT 4299-2, Bradford, CT, USA) installed in the top of the tank automatically terminates the pumping procedure when the tank is full. Accurate bath levels are efficiently and readily achieved using the automated pump's variable speed, which also permits fine adjustments to the fluid level remotely. Incorporating custom designed seals (made of a Teflon outer shell and two internal O-rings) into the base-plate has prevented liquid leakage through the tank-base antenna entry-sites. Each seal has an opening in its center that allows the protected feedline to enter the tank. A custom threaded fixture holds the seal in place and compresses the internal O-rings, creating a liquid-tight closure. Figure 4 shows photographs of the antenna array-mounting plates and seals as they appear within the complete-imaging platform, as well as a perspective view of the sub-arrays configured in and out-of-plane with one another.

C. Antenna motion-control system

Antenna-array movement is driven by four rotary motors (Parker Hannifin S57-83, Cleveland Ohio, USA) mounted inline to a companion linear actuator (Parker Hannifin ET32 803-4960A-269). Wired to each complex is a motion-controlling amplifier. Their connection to the system's computer uses an RS-232 serial interface, and includes a micro-controller to regulate motor movement. With a resolution of 0.0254 mm at 25 000 steps per revolution,⁴⁷ the motors can obtain a maximum velocity of 396 mm/s. The high torque achieved by the four assemblies is required to drive the actuators with a force sufficient to overcome resistance from the compressed O-ring seals, allowing translation of the two antenna SAs to different heights within the illumination tank.

FIG. 4. Photographs of (a) the two sub-array mounting plates and associated antennae, (b) the antennas entering the illumination tank from the bottom of the imaging apparatus through the base-plate holes and O-rings seal, (c) the sub-arrays configured in-plane, and (d) the sub-arrays configured in a cross-plane position.

The four actuator-motor complexes are symmetrically located underneath the illumination chamber, and attach to two independent sub-array mounting plates (see Figure 4). Opposing pairs control the same antenna-mounting plate. In this design, the 16-element antenna array is composed of two interwoven eight-element SAs, where neighboring antennas are associated with the complementary SA. The ability to move the SAs independently to various IP and XP orientations within the tank has increased the number of independent measurements sites, facilitating 3D volumetric data collection.

1. 2D movement strategy

Two-dimensional imaging utilizes only IP data. For a given imaging plane, both antenna SAs reside at the same vertical location during signal transmission and detection. Each of the 16 elements individually transmits microwave signals while the 15 complementary non-transmitting elements receive the transmitted signals in parallel, resulting in 240 measurements for a single frequency (and is repeated for multiple frequencies). For a typical 2D seven-plane imaging procedure, the system collects amplitude and phase measurements from plane 1 (P1), located at the top of the imaging tank, down to plane 7 (P7). The distance between imaging planes depends on patient breast size, and is system-operator selected to achieve full coverage of the breast while submerged in the illumination chamber.

2. 3D movement strategy

Three-dimensional imaging acquires both IP and XP data. The start of the 3D data acquisition sequence is similar to the 2D case, both SAs begin at P1 and amplitude and phase measurements are collected. Then, SA2 retracts to P2 while SA1 remains in P1, followed by measurement collection for this XP configuration. Next, SA2 moves to P3 (while SA1 remains in P1) and amplitude and phase measurements are again collected. This process continues until SA2 has reached and recorded data for all seven initial cross-plane positions. It then returns to P1 while SA1 simultaneously moves to P2.

Data acquisition then occurs at the next set of image-plane combinations. The iterative process continues until SA1 has reached the final-imaging plane and data have been collected for all n^2 (7^2 in this case) combinations of SA1 and SA2 positions.

D. Data acquisition and processing

The entire system is automated and controlled by a single personal computer (PC) through a PXI Embedded Controller (National Instruments (NI) PXI-8176 Austin, Texas, USA) utilizing a GPIB Ethernet Controller connection (NI PXI-8231). An 8-Slot 3U Chassis with Universal Power Supply (NI PXI-1042) controls the rf signal generator and the power supplies, and an RS-232 serial port connection (in conjunction with the micro-controller) facilitates antenna-array motion. A Digital I/O Module (NI PXI-6508) allows logic control of the switching matrix and adjustments to the digital attenuator.

Signal measurements are mixed with the reference LO signal and the modulated, low-pass filtered IF output is sampled in parallel by two eight-channel 24-bit Dynamic Signal Acquisition Boards (NI PXI-6508) at a sampling rate of 102.4 kS/s.⁴⁸ Amplitude and phase measurements extracted from the IF signal using a software-based lock-in amplifier technique serve as inputs to the reconstruction algorithm. The acquisition time for a complete set of data (for a fixed position at a single frequency) is 5.8 s. Overall system operation, including signal conditioning, A/D conversion, switching-matrix control, and antenna motion is coordinated using a custom LabVIEW (NI) program. A user-friendly graphical user interface (GUI) allows system operation from a PC located outside the clinical examination room for improved patient privacy.

1. Imaging algorithm

This section includes a brief explanation of the image reconstruction process, which uses algorithms identical to those previously reported; it will not be covered in detail here. Electric-field amplitude and phase measurements

are initially normalized according to a standard calibration procedure²⁰ for offline image reconstruction. A 3D iterative Gauss-Newton algorithm, utilizing the finite-difference time-domain (FDTD) method as the computational engine for calculating the forward solutions at each iteration is described thoroughly in Ref. 49. The most unique aspect of the algorithm is the log transformation, which makes the inverse problem more linear and is described in Refs. 50 and 51. Phase unwrapping of the measured and computed field values is required as part of the log-transformation process; these techniques are described in Refs. 40 and 52. More recent results by Grzegorzczak *et al.*⁵³ have confirmed that the algorithm does not converge to local minima, as reported for other techniques.^{54–56} In all cases, the initial electrical property values for our iterative reconstruction process were the same as those of the coupling medium.

III. RESULTS

Reconfiguring the microwave electronics, DAQ and motion-control components have increased the system's data-acquisition speed and measurement capabilities. In this section, data are presented to assess the system's performance in terms of (a) sensitivity, (b) channel-to-channel isolation, and (c) measurement repeatability. An analysis of the system's (d) IP and (e) XP measurement sensitivity is included and optimal operational-parameters are identified using SNR metrics. Reconstructed images of a breast-shaped phantom with a tumor-like inclusion are shown to demonstrate system-imaging capabilities using the new XP data.

A. System sensitivity

System sensitivity and absolute error were evaluated by assessing each channel using an Agilent ESG-D series 4432B digital rf signal generator. An important goal of the system redesign was maintenance of linear output as a function of input power. For this analysis, each transmitting antenna-channel was directly connected to the external rf source (with an inline 10 dB attenuator) and signal phase and amplitude measurements were acquired (from 500 to 2900 MHz in 300 MHz increments) over a range of calibrated input power-levels. Figure 5 shows the resulting data.

Similar to the previous system, phase measurement errors are the limiting factor in detecting reliable low-power signals. From Figure 5, the lowest detectable signal with phase RMS errors less than 1° is roughly -130 dBm for frequencies below 2300 MHz, and degrades to -110 dBm over the higher frequency band. Overall, the absolute amplitude and phase errors increased from average values of 0.33% and less than 0.5° at higher input power-levels to greater than 1% and 1° as the input signal's power-level decreased towards the noise floor, respectively. These values are absolute (amplitude and phase RMS) errors calculated for each discrete frequency (from 500 MHz to 2900 MHz in 300 MHz increments). They were also averaged over the system's optimal (i.e., input-power spectrum-span with phase error is less than 1°) frequency and overall operational frequency bands, re-

spectively. RMS phase error represents the variation between the recovered phase values of a reference signal supplied directly to the DAQ and the measured signal that traveled the system's receiver component cascade to the DAQ. Similarly, absolute amplitude error is the difference between the system-based and reference signals' amplitude. Absolute errors were determined by subtracting the DAQ-acquired reference values from the system-based measurements, and the averaged absolute errors are the mean values of these absolute errors (amplitude and phase RMS, respectively) calculated over a given frequency range.

B. Channel-to-channel isolation

By assessing signal leakage and coupling between various system pathways, a channel-isolation evaluation was permissible. For this analysis, the antenna ports of transceiver units 1 and 13 (relative-receiving channel 12) were directly connected using a short coaxial cable. Relative-receiving channels are numbered sequentially from 1 to 15 (starting immediately after the transmitting antenna) based on the transmitting channel's array position. $50\ \Omega$ matched loads terminated the antenna ports of all remaining transceiver units. With channel 1 active, this setup directly routed the transmitted signals to relative-receiving channel 12. Using the system's DAQ for data acquisition, signal measurement collection for all non-transmitting channels occurred simultaneously. All potential ground paths in the dc power electronics and feedline configurations were isolated, and the only remaining multipath signals resulted from leakage through the rf and LO networks.

Results shown in Figure 6(a) reveal interesting behavior in channels connected to the same SP4T switch in the rf switching network (i.e., relative-receiving channels 1, 2, and 3, respectively). Elevated signal-levels were detected at all three non-transmitting channels associated with the same SP4T switch, especially above 1300 MHz. By reducing the size of the switch-matrix network, we cut the number of matrix layers from three to two (relative to our second-generation system). In the new design, the leakage is only associated with channels that share a SP4T switch in the rf network and decreases as the distance between the receiving and transmitting switches increases.

By incorporating additional attenuating elements in the rf network, our new system achieves inter-channel isolation similar to that of the previous generation. Figure 6(b) shows the corresponding measurements collected by the DAQ under the same test setup after incorporating the additional attenuation, which increased the system's dynamic range by roughly 20 dB, and lowered the noise floor to approximately -115 dB.

C. Measurement repeatability

To evaluate the measured data's quality, a breast-shaped phantom composed of an 86:14 glycerin/water mixture ($\epsilon_r = 15.8$, $\sigma = 0.98$ S/m), including a high-contrast tumor-like inclusion made of a 50:50 glycerin/water mixture ($\epsilon_r = 53.1$, $\sigma = 1.43$ S/m), was repeatedly interrogated while sub-

FIG. 5. System (a) amplitude sensitivity response and (b) phase error as a function of input power.

merged in an 80:20 glycerin/water coupling fluid ($\epsilon_r = 22.1$, $\sigma = 1.25$ S/m). The dielectric properties listed here were measured independently with a slim-form dielectric probe (Agilent 85070E Probe Kit, Santa Clara, California, USA), and represent those typical of the breast over our frequency band of interest (in this case 1300 MHz).

Datasets consisting of 10, 20, 50, and 100 DAQ-acquired amplitude and phase measurements consecutively extracted from the same microwave signal transmissions were used to evaluate measurement fluctuations. The system was powered-up for approximately 1 h prior to the experiments as part of the operational protocol to allow the signal source, DAQ, electronic-network components and cables to reach their steady state operating temperatures. According to manufacturer specifications, the rf source signal amplitude and phase can vary up to 0.06 dB and 0.2° with temperature fluctuations in the ambient environment, respectively. Based on these precautions, system measurement repeatability is min-

imally influenced by variations in the room temperature where the experiments were conducted. For frequencies up to 2300 MHz, the averaged absolute error for each receiving channel (i.e., the average value of the absolute errors for each of the four sample sizes) was 0.4% and less than 1° for the measured amplitude and phase values, respectively. At higher-operating frequencies, repeatability performance degraded, with the amplitude and phase errors reaching 0.8% and greater than 2° , respectively. The larger errors associated with high-frequency operation are primarily due to increased signal attenuation in the lossy coupling-medium.

D. In-plane measurement sensitivity

Signal amplitudes and the corresponding noise floor for each channel were collected over the operational bandwidth when the antenna SAs were located in the same plane. The resulting SNRs for each receiving channel are plotted in

FIG. 6. Plots of measured signal amplitudes (dB) with channel 1 directly connected to channel 13 (relative receiver 12), with $50\ \Omega$ loads terminating all other channels (a) prior to introducing additional attenuating components in the rf switch matrix and (b) after incorporating additional attenuating components into the rf switch matrix.

Figure 7 and listed in Table I. SNRs greater than 40 dB were observed at all receiving channels for frequencies up to 1500 MHz, and degraded as frequency and distance between the transmitting and receiving elements increased. In the array's circular arrangement, the positions of relative-receiving channels 7, 8, and 9 are furthest from the transmitting element; consequently, signals received at these locations experienced the largest amount of attenuation. These channels had SNRs greater than 30 dB up to 1900 MHz, while elements located closer to the transmitting antenna were capable of achieving SNRs greater than 80 dB for frequencies as high as 2900 MHz.

E. Cross-plane measurement sensitivity

Large transmitter-receiver separation distances and high-frequency operation contribute directly to increased signal

FIG. 7. SNR values plotted for each receiving channel of the parallel-detection scheme.

attenuation. Hence, cross-plane data collection requires assessing antennas on the SA complementary to the transmitting element, as their distance relative to the transmitter increases during the XP motion procedure. Consequently, SNRs for XP relative-receiving channels 9, 11, and 13 were evaluated for vertical separation distances from 1 to 11 cm, in which case distances from the XP transmitter decreased as a function of increasing channel number. Tables II–IV summarize the resulting data for channels 9, 11, and 13, respectively. For relative receiver 13, SNRs greater than 40 dB were identified for XP separation distances up to 10 cm (at all frequencies).

As anticipated, the SNRs for relative-receiving channels 11 and 9 degraded due to their location on the XP array. For channel 11, SNRs greater than 35 dB were found for XP separation distances up to 9 cm (for frequencies as high as 1500 MHz); moreover, SNRs greater than 40 dB were recorded for XP separation distances up to 6 cm over the same frequency range. As expected, relative-receiving channel 9 was the limiting array element, although SNRs greater than 35 dB were observed for XP separation distances up to 11 cm (for frequencies as high as 1100 MHz) and 40 dB for XP separation distances up to 6 cm (for frequencies as high as 1300 MHz).

F. Cross-plane data and image reconstruction sensitivity

Three-dimensional image reconstruction used IP and XP data. Since XP data are influenced by sub-array separation distance, as these distances increase, transmitted signals (measured by out-of-plane receiving elements) experience progressively greater attenuation due to longer transmit-receive travel-path lengths through the lossy medium. A longer XP separation distance also leads to non-optimal receiver alignment within the transmitting element's radiation beam. Thus, the quality of measurement data collected

TABLE I. Tabulated SNRs (dB) for each relative-receiving channel (Rx#^a) as a function of frequency (MHz).

	Rx1	Rx2	Rx3	Rx4	Rx5	Rx6	Rx7	Rx8	Rx9	Rx10	Rx11	Rx12	Rx13	Rx14	Rx15
500 MHz	114	107	111	100	93	96	83	89	84	88	87	91	103	111	124
700 MHz	109	113	100	109	90	89	86	93	82	87	96	107	106	114	119
900 MHz	109	104	87	92	89	84	78	76	81	91	84	91	106	104	116
1100 MHz	97	101	76	95	90	80	79	71	65	75	99	100	107	119	125
1300 MHz	85	90	65	80	84	70	69	58	54	66	84	84	95	116	124
1500 MHz	79	80	54	70	67	64	60	48	43	58	67	84	92	107	139
1700 MHz	69	70	41	65	63	49	42	44	36	58	58	84	80	119	118
1900 MHz	65	58	30	47	43	23	32	32	31	35	39	61	70	90	106
2100 MHz	65	49	21	38	29	31	23	29	20	32	40	52	68	104	122
2300 MHz	57	44	15	28	19	30	31	16	16	33	29	55	71	89	105
2500 MHz	55	40	14	20	12	21	20	10	12	20	22	37	55	77	99
2700 MHz	49	27	14	13	9	−11	15	8	10	27	18	35	42	69	90
2900 MHz	40	12	14	15	14	11	17	11	2	21	16	9	30	58	84

^aRx# is the channel number for each relative-receiving channel of the system's parallel-detection scheme.

at larger XP separation distances decreases, and its inclusion into image reconstruction can degrade the final images (as a consequence of using data with low SNRs).

To evaluate the effect of incorporating XP data into 3D image reconstruction, a breast-shaped phantom filled with an 86:14 glycerin/water solution ($\epsilon_r = 14.1$, $\sigma = 1.05$ S/m at 1500 MHz) including a tumor-like inclusion composed of a 15:85 gelatin/water mixture ($\epsilon_r = 48.88$, $\sigma = 1.74$ S/m at 1500 MHz) was submerged in an offset position and imaged in an 80:20 glycerin/water coupling fluid ($\epsilon_r = 19.8$, $\sigma = 1.38$ S/m at 1500 MHz), as shown in Figure 8. Data were collected from nine consecutive tomographic-imaging plane permutations using the XP sub-array motion strategy outlined in Sec. II C 2. The resulting measurements were used to reconstruct three (permittivity and conductivity) image sets incorporating 3-XP, 6-XP, and 9-XP of data (appearing in Figures 9–11, respectively). The XP number specifies the range of successive out-of-plane data selected from the complete set of microwave measurements and served as the post-process data-sorting criteria used to reconstruct the images shown in the figures (i.e., the data-sorting criteria were the amount of XP-acquired data introduced into image reconstruction). Higher XP numbers indicate more out-of-plane data were used during image recovery. For a given XP number (n), all IP and n XP sorted data combinations were simultaneously incorporated into the image inversion process to generate 3D volumetric images.

TABLE II. Tabulated SNRs (dB) for relative-receiving channel 9 as a function of cross-plane (XP^a) separation distance (cm).

	XP1	XP2	XP3	XP4	XP5	XP6	XP7	XP8	XP9	XP10	XP11
1100 MHz	55	53	52	52	50	48	44	42	40	38	35
1300 MHz	51	50	47	45	43	40	37	34	32	27	22
1500 MHz	44	42	39	37	34	31	28	26	23	21	18
1700 MHz	31	29	27	25	23	21	19	18	17	14	12
1900 MHz	21	17	15	12	9	6	3	<1	<1	<1	<1
2100 MHz	19	15	13	10	7	3	1	<1	<1	<1	<1
2300 MHz	13	7	3	<1	<1	<1	<1	<1	<1	<1	<1

^aXP is the cross-plane sub-array separation distance in (cm).TABLE III. Tabulated SNRs (dB) for relative-receiving channel 11 as a function of cross-plane (XP^a) separation distance (cm).

	XP1	XP2	XP3	XP4	XP5	XP6	XP7	XP8	XP9	XP10	XP11
1100 MHz	55	54	54	54	54	53	53	52	52	50	48
1300 MHz	53	52	51	49	47	45	43	41	38	36	34
1500 MHz	51	47	46	46	44	40	39	37	35	34	29
1700 MHz	40	38	35	34	34	32	31	29	27	23	16
1900 MHz	30	28	27	24	22	19	17	15	13	10	6
2100 MHz	20	14	12	11	12	12	12	12	8	2	<1
2300 MHz	19	17	14	12	10	8	6	4	2	1	<1

^aXP is the cross-plane sub-array separation distance in (cm).TABLE IV. Tabulated SNRs (dB) for relative-receiving channel 13 as a function of cross-plane (XP^a) separation distance (cm).

	XP1	XP2	XP3	XP4	XP5	XP6	XP7	XP8	XP9	XP10	XP11
1100 MHz	85	87	81	79	77	74	71	68	63	57	55
1300 MHz	90	90	88	86	83	78	73	69	64	61	58
1500 MHz	90	89	89	89	87	82	76	70	65	56	50
1700 MHz	91	88	89	88	85	78	70	61	57	52	46
1900 MHz	91	85	87	84	78	70	63	56	53	43	44
2100 MHz	86	80	81	76	70	63	55	49	47	47	47
2300 MHz	78	75	73	70	67	58	49	47	45	43	38

^aXP is the cross-plane sub-array separation distance in (cm).

FIG. 8. Photograph of (a) the breast phantom with irregular shaped inclusion placed in imaging tank and (b) a top-down view showing the breast phantom and tumor inclusion.

FIG. 9. 1500 MHz (a) permittivity and (b) conductivity images (of the phantom shown in Figure 8) incorporating 3-XP of data in the 3D reconstruction algorithm.

These XP distances were selected based on the range of their minimum SNRs—3-XP (SNR of 39 dB), 6-XP (SNR of 31 dB), and 9-XP (SNR of 23). Each image set incorporated a greater amount of data but also more measurements influenced by noise (i.e., data collected at larger XP separation distances) into its reconstruction, and was formed with our 3D reconstruction algorithm under the same initial property estimates and IP microwave data. The only difference between the three configurations is the amount of XP data included into their post-process reconstruction. The dielectric property contrast between the breast and tumor region is evident in all images. For the 9-XP permittivity and conductivity images (Figures 11(a) and 11(b)), artifacts begin to appear at the breast phantom's top surface and edges. These features are visible in the 6-XP permittivity and conductivity images (Figures 10(a) and 10(b)) to a lesser extent. The diminished artifacts presumably result from the incorporation of data less impacted by signal noise, as indicated by the higher SNR. The 3-XP images (Figures 9(a) and 9(b)) are nearly artifact free. Additionally, the value at the center of the inclusion is overestimated in both the 6- and 9-XP images, especially the conductivity (2.5 S/m and 3.4 S/m for the 6-XP and 9-XP, respectively), when compared to the 3-XP value of 1.71 S/m. Although the tumor-like inclusion was arbitrarily positioned in the breast phantom, making a location error analysis imprecise, its cylindrical shape was accurately recovered in the 3-XP permittivity and conductivity images. The inclusion's shape progressively became less accurate as the amount of incorporated XP data increased; in fact, its shape becomes increasingly spherical as more XP data are included in the reconstruction algorithm.

FIG. 10. 1500 MHz (a) permittivity and (b) conductivity images (of the phantom shown in Figure 8) incorporating 6-XP of data in the 3D reconstruction algorithm.

FIG. 11. 1500 MHz (a) permittivity and (b) conductivity images (of the phantom shown in Figure 8) incorporating 9-XP of data in the 3D reconstruction algorithm.

IV. DISCUSSION

A third-generation microwave tomographic imaging system has been developed, which has a more compact form factor for clinical breast exams. Its modularity has increased electrical isolation by physically separating electrical components from the patient interface, and improved system functionality. Introducing a fixed bulkhead into the electronics compartment partition (directly underneath the wall-mounted transceiver modules) has streamlined antenna feedline connections to the system's electronic networks, and created a convenient method for removing and exchanging the system's antenna array with various prototypes.^{57,58} Having taken advantage of this linkage strategy, we are now capable of conducting multimodal MR-MT exams.⁵⁹

Upgrading the microwave electronic networks has substantially enhanced system performance. Receiver sensitivity has improved by reducing the number of network components that introduce mismatch related signal losses. Furthermore, it attains reliable low-power signal measurements (down to -130 dBm) with limited amplitude and phase measurement errors up to 2300 MHz. Adequate inter-channel isolation has been maintained, and by incorporating additional attenuating components into the rf network, a dynamic range of greater than 110 dB is achieved up to 2900 MHz. The updated 24-bit DAQ has significantly increased data-acquisition rates when compared to the previous system. Moreover, incorporating the automated fluid-pumping system has dramatically reduced coupling-liquid filling and draining times, and fine-tuning of the illumination chamber's coupling-fluid level can be made in real time (i.e., during individual patient exams).

Advancements to the motion system have increased the number of independent measurement positions available for 3D volumetric image reconstruction by acquiring both IP and XP signals. We have observed IP SNRs greater than 35 dB up to 1700 MHz for all receiving elements, while antennas located closest to the transmitter have SNRs as high as 80 dB up to 2900 MHz. Signal measurement quality is also a function of the complementary SA position. XP data acquisition of signals with SNRs as high as 35 dB are seen for SA separation distances up to 11 cm (for frequencies as high as 2300 MHz), and degrade as frequency and distance from the transmitting element increases.

Independent SA control has significantly enhanced our system's measurement potential. As we gain experience in the processing of more images with our new 3D algorithm,

TABLE V. Quantitative comparisons of the previous (2nd Generation) and current (3rd Generation) systems' hardware performance.

	2nd generation system		3rd generation system	
Measurement sensitivity (from -30 to -110 dBm)	Amplitude (% error)	Phase error (deg)	Amplitude (% error)	Phase error (deg)
	0.4	0.8	0.325	0.5
Dynamic range	~ 115 dB		~ 115 dB	
Measurement repeatability (from 500 to 2000 MHz)	Amplitude (% error)	Phase error (deg)	Amplitude (% error)	Phase error (deg)
	0.5	1	0.4	<1
Motion system capabilities	IP measurement positions		IP and XP measurement positions	
# of independent measurement positions (for 7 imaging planes)	1680 measurement (for a single frequency)		11 760 measurement (for a single frequency)	
Linear actuator specs	1 actuator 315 ft-lb of torque 2:1 gear ratio		4 paired actuators 540 ft-lb of torque 1:1 gear ratio	
Stepper motor specs	0.05 mm resolution 25 000 steps/revolution		0.0254 mm resolution 25 000 steps/revolution	
DAQ Specs	16-bit resolution		24-bit resolution	
Data-acquisition speed	8 s (for a single imaging plane and operating frequency)		5.8 s (for a single imaging plane and operating frequency)	
RF switch matrix configuration	32 channels (three levels of SP4T switches)		16 channels (two levels of SP4T switches)	
LO network configuration	32 channels (31 2-way power dividers) Note: only 16 channels were used in the final design		16 channels (one 2-way power divider in-line with two 8-way power dividers)	

we will leverage these new innovations in terms of improved dynamic range and increased amount of measurement data by including the cross-plane data clinically. A quantitative comparison of the previous (2nd Generation) and current (3rd Generation) systems' hardware performance is provided in Table V. The initial results presented here provide important guidelines with respect to how much data are necessary and what signal-quality level is expected for the new XP data-acquisition scheme.

Future designs will incorporate flash programmable amplifier drives and low-profile motion encoders, providing two-way communication between the PC and motion system. The iterative motion strategy currently employed for 3D data acquisition is a consequence of the implemented motion hardware's functionality, which lacks encoded motion capabilities. While we can certainly collect the XP volumetric data needed for 3D image reconstruction with the current design, utilizing motion encoders that relay motor position information back to the PC will permit the triggering of custom (non-iterative) motion sequences for optimized XP data acquisition, which will significantly reduce 3D clinical-imaging exam times.

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¹See www.cancer.org/cancer/breastcancer/detailedguide/breast-cancer-key-statistics for current American breast-cancer statistics.

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