Breast Cancer Detection Using Interferometric MUSIC: Experimental and Numerical Assessment

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Purpose: In microwave breast cancer detection it is often beneficial to arrange sensors in close proximity to the breast. The resulting coupling generally changes the antenna response. As an a priori characterization of the radio-frequency system becomes difficult, this can lead to severe degradation of the detection efficacy. The purpose of this paper is to demonstrate advantages of adopting an Interferometric Multiple Signal Classification (I-MUSIC) approach due to its limited dependence from a priori information on the antenna. The performance of I-MUSIC detection was measured in terms of Signal-to-Clutter Ratio (SCR), Signal-to-Mean Ratio (SMR) and Spatial Displacement (SD) and compared to other common linear non-coherent imaging methods, such as migration and the standard wideband MUSIC (WB-MUSIC) which also works when the antenna is not accounted for.

Methods: The data were acquired by scanning a synthetic oil-in-gelatin phantom that mimics the dielectric properties of breast tissues across the spectrum 1 – 3 GHz using a proprietary breast microwave multi-monostatic radar system. The phantom is a multi-layer structure that includes skin, adipose, fibroconnective, fibroglandular and tumor tissue with an adipose component accounting for 60% of the whole structure. The detected tumor has a diameter of 5 mm and is inserted inside a fibroglandular region with a permittivity contrast \( \varepsilon_{\text{tumor}} / \varepsilon_{\text{fibroglandular}} < 1.15 \) over the operating band. Three datasets were recorded corresponding to three antennas with different coupling mechanisms. This was done to assess the independence of the I-MUSIC method from antenna characterizations. The datasets were processed by using I-MUSIC, non-coherent migration and Wide Band MUSIC under equivalent conditions (i.e. operative bandwidth, frequency samples and scanning positions). SCR, SMR and SD figures were measured from all reconstructed images. In order to benchmark experimental results, numerical simulations of equivalent scenarios were carried out using CST Microwave Studio. The three numerical datasets were then processed following the same procedure that was designed for the experimental case.

Results: Detection results are presented for both experimental and numerical phantom and higher performance of the I-MUSIC method in comparison with the WB-MUSIC and non-coherent migration is achieved. This finding is confirmed for the three different antennas in this study. Although a delocalization effect occurs, experimental datasets show that the signal-to-clutter ratio and the signal-to-mean performance with the I-MUSIC are at least 5 and 2.3 times higher than the other methods, respectively. The numerical datasets calculated on an equivalent phantom for cross-testing confirm the improved performance of the I-MUSIC in terms of SCR and SMR. In numerical simulations the delocalization effect is dramatically reduced up to an SD value of 1.61 achieved with the I-MUSIC in combination with the antipodal Vivaldi antenna. This shows that mechanical uncertainties are the main reason for the delocalization effect in the measurements.

Conclusions: Experimental results show that the I-MUSIC generates images with signal to clutter levels higher than 5.46 dB across all working conditions and it reaches 7.84 dB in combination with the antipodal Vivaldi antenna. Numerical simulations confirm this trend and due to ideal mechanical conditions return a signal to clutter level higher than 7.61 dB. The I-MUSIC largely outperforms the methods under comparison and is able to detect a 5-mm tumor with a permittivity contrast of 1.5.
I. INTRODUCTION

Breast cancer is the most common type in women and incidences are increasing in the developing world due to extended life expectancy, increased urbanization and expansion of western lifestyles. The World Health Organization launched a key message to stress the importance of “early detection in order to improve breast cancer outcome and survival remains the cornerstone of breast cancer control”[1]. So far the only breast cancer screening method that has proved to be effective is mammography screening. However, its detection capabilities have been shown to be limited due to poor benign/malignant tissue contrast (around 10%). Between 4% and 34% of all breast cancers are missed and nearly 70% of all breast lesions turn out to be benign[2]. The exposure to low levels of ionizing radiation may reduce patient compliance with screening. On the other hand, Magnetic Resonance Imaging (MRI) offers higher performance in terms of resolution and, hence, a correct diagnosis. Typical resolution in a breast image obtained via a 3-Tesla (3-T) MRI system is about 1 mm. However, MRI is still expensive and not proven to be a practical procedure for wide screening campaigns. Ultrasound can supplement X-ray mammograms by discerning between liquid cysts or solid tumors. Alternative methods at exploratory stage are based on different tissue parameters such as elasticity, temperature and optical properties. Exploiting dielectric contrast to image relevant features in biological targets has been investigated for decades[3] but recently this research area has gained critical mass. Higher spatial resolution is achievable by using X-ray radiation, but radio frequency (RF) technology offers higher dielectric contrast between normal and diseased breast tissues, which can supplement conventional diagnostics. This can enable the reduction of false diagnosis with consequent high social impact and reduced costs to health systems. Moreover, it is based on low-power non-ionizing radiation which facilitates cost-competitive screenings. These important potential outcomes have triggered the investigation of many microwave imaging techniques, aimed at detecting, localizing and identifying tumors in breast tissues[4]. However, while the contrast between malignant and adipose breast tissues may be as large as 10, those between malignant and healthy fibroglandular tissues can be as low as 10%, in both permittivity and conductivity[5,6]. This places an important challenge in microwave breast imaging as most breast tumors appear in the fibroglandular tissues.

Several microwave imaging techniques were proposed[7-15] in the last two decades. These can be generally grouped into nonlinear and linear inversions. Non-linear inversions aim at retrieving the breast dielectric and conductivity profiles[7,8]. However, they are computationally intensive[9] and can suffer from convergence and reliability problems due to failure solutions[10]. Linear methods (also addressed as radar focusing) are robust and computationally effective but they only allow detection and localization of the tumor inhomogeneities[11] with the requirement for further classification steps[12]. In any case, the corresponding inverse scattering problem is very difficult as the tumor is buried within a highly inhomogeneous medium.

Imaging algorithms are only part of the picture. A crucial role is played by the RF system and in particular by the antennas. Ultra-wideband (UWB) antennas have been proposed for this application as they can offer a very large operating band, stable radiation properties and compact dimensions both with planar[13,14] and 3-D profiles[15]. In order to fully assess the achievable performance of a breast cancer detection system, the presence of the antenna must be taken into account. Ideal sources are worthy of consideration at a preliminary stage when the focus is limited to the algorithm. Recent contributions deal with more complex scenarios where antennas are considered in the numerical model[16,17]. Also a number of antenna prototypes were developed and integrated into experimental phantom-based systems[18] or under preclinical conditions[19].

Accurate near-field microwave imaging requires characterization/equalization of the antenna behavior. This can be pursued by a suitable set of measurements or numerical simulations. However, as the breast properties change from patient to patient, residual errors still remain. With uncertainty levels as high as the magnitude of the tumor scattered field, the imaging procedure robustness is dramatically endangered. In particular, dense breasts are particularly exposed to this problem as they present lower tumor/healthy-tissue contrast.

In order to mitigate the requirement for antenna characterization in complex near-field scenarios suitable detection methods should be properly devised. Interferometric-MUSIC was developed[16] to address this issue. This approach represents a multi-frequency variant of the well-known time-reversal MUSIC[20] when adapted to a multi-monostatic configuration. Theoretical details and performance assessment were addressed by Solimene et al.[21]. Preliminary numerical results on a breast scenario were reported by Ruvio et al.[22]. In this study a 2-D numerical mostly fatty breast model was derived from an MRI scan according to the Wisconsin’s repository[23]. It is shown, that the method succeeds in detecting a low-contrast tumor when the antenna response is completely neglected in the imaging procedure. As the scattered signal coming from the tumor is generally weaker than reflections from benign tissues, clutter removal procedures are an important step in the diagnostic system. Several clutter-rejection methods have been proposed in the literature. Some of them rely on filtering procedures that attempt to estimate the clutter signal at a given position and subtract it from the actual measurements[24]. Alternatively, some procedures employ a differential scheme where two sets of measurements corresponding to two different rotations of the system are subtracted from each other[25]. These procedures strongly depend on the uniformity of the skin layer, breast shape, and on antenna response. Moreover, they can also result in filtering part of the signal weakly scattered by the tumor scattered signal. Time gating procedures can exploit entropic metrics that enables identification of time windows in which signals have to be silenced[26] but rely on similar assumption about the breast. Subspace projection methods
also entails filtering the scattered signals but being based on amplitude differences do not required uniformity assumptions. Sarafianou et al. proposed a novel skin reflection removal algorithm \(^{27}\) that operates starting from a preliminary breast surface estimation which is then used to create a synthetic phantom. Finally, the field scattered by this synthetic phantom is subtracted from the actual measurements.

Preclinical testing requires the realization of accurate experimental phantoms with varying degree of complexity\(^{28}\). In general, the realization of synthetic materials that closely mimic the physical properties of various human tissues becomes very difficult when a large operating frequency range is required.

In this paper, the achievable performance of the I-MUSIC in combination with a subspace-based anti-clutter technique is experimentally evaluated with a 2-D inhomogeneous breast phantom that was manufactured with oil-in-gelatin emulsions. These tissue-mimicking materials are able to reproduce the electrical properties of different normal and malignant breast tissues\(^{29}\). Different dielectric properties can be obtained by varying the percentages of a 50% kerosene - 50% safflower oil solution to simulate the electric behavior of healthy and diseased breast tissues. Experimental data are obtained by using three different types of antennas (i.e. a planar monopole, a semi-folded monopole and an antipodal Vivaldi antenna). Data are then processed to detect a 5-mm tumor included within a fibroglandular region. Moreover, experimental results were cross-referenced against equivalent numerical model simulations. The achievable performance obtained by I-MUSIC is then compared with two other methods in the literature: the non-coherent migration\(^{30}\), which is a particular version of beamforming, and the Wide Band MUSIC\(^{31}\) which belongs to the class of spectral estimation algorithms as I-MUSIC. The aim of this paper is to highlight the enhanced detection performance achieved by using I-MUSIC when no a priori antenna characterization is given. To the best of our knowledge, this paper is the first to analyze and quantify through experimental and numerical data the detection capabilities of I-MUSIC by using three non-characterized antennas with different antenna/tissues coupling mechanisms.

II. PHANTOM AND MEASUREMENT SETUP

Breast phantoms based on oil-in-gelatin emulsions can reproduce the electrical properties of various normal and malignant breast tissues. An essential property of these materials is the capability to create heterogeneous anthropomorphic structures with long-term stability of mechanical and electromagnetic properties. Moreover, due to their gelatinous consistence, these materials are convenient for relatively easy and inexpensive manufacturability as well as being characterized for the Radio Frequency range of interest in this study. For these reasons oil-in-gelatin materials were
selected to manufacture a heterogeneous 2-D phantom (Fig. 1a) including skin, adipose, fibroconnective, fibroglandular and tumor tissues. 2-D phantoms (if not 1-D) represent a convenient and common choice for a faster and fair system assessment\textsuperscript{14,32,33,34}. In particular, phantom simplification enables a more detailed investigation when non-ideal antennas are used\textsuperscript{17}.

Different tissues were realized by properly mixing the 50\% kerosene - 50\% safflower oil solution with a formaldehyde-based emulsion. The following oil-percentages were used to make the tissues: 80\%-oil concentration for the adipose tissue, 40\%-oil for fibroconnective tissue, 30\%-oil for fibroglandular, and skin tissues, 20\%-oil for the tumor, respectively\textsuperscript{28}. The phantom has an overall diameter of 114 mm including the 2-mm thick skin layer. The fibroconnective and fibroglandular regions have 68 mm and 20 mm diameters, respectively. The multi-layer structure was realized with a multi-stage procedure: first the adipose tissue was poured into a mould where a rod was introduced to make room for the fibroconnective / fibroglandular / tumor structure. After the adipose tissue had gelled, the same procedure was repeated for the other internal layers. Finally, a 2-mm thick skin tissue was attached to the adipose tissue. Based on such adipose composition, this breast model can be categorized into heterogeneous mix (31-84\% adipose)\textsuperscript{6}. In order to create a challenging detection scenario, the 5-mm diameter tumor was asymmetrically located inside the fibrogranular region.

The dispersive behavior of skin, adipose, fibro-connective, fibro-glandular and tumor properties was measured over a large frequency range spanning from 1 to 4 GHz using a coaxial probe (Fig. 2a). As can be seen, the real part of the dielectric permittivity is reasonably close to the average values measured by Laezbnik \textit{et al.}\textsuperscript{5}. The tissues realized for this study present a conductivity contrast between benign and tumor tissues which is lower than to the average values measured by Laezbnik \textit{et al.}\textsuperscript{6} and it is not displayed. This choice requires more robustness from the detection system and it only relies on the permittivity contrast. In particular, the overall dielectric contrast between the tumor and the fibroglandular tissue is no greater than 1.5:1 which can be considered relatively low. The phantom was also imaged using MRI in order to assess its compliance from a morphologic perspective as well. The MR images were acquired on a 3T system (Achieva, Philips, Netherlands) using a high resolution 3D T2-weighted imaging sequence with a spatial resolution of 0.7 x 0.7 x 0.7 mm\textsuperscript{3}. From Fig. 1b it can be appreciated that the desired layering was actually achieved.

The system \textit{antenna + phantom} was immersed into a coupling medium which presents properties equivalent to the expected adipose tissue ($\varepsilon_r$\textsuperscript{coupling\_medium} = $\varepsilon_r$\textsuperscript{adipose} = 12). The adipose component is in fact dominant in the breast tissue distribution. The coupling medium was realized with the same 80\%-oil emulsion used to make the adipose tissue in the phantom but without gelling agents. The medium improves coupling from the antenna to the phantom and also enables antenna miniaturization. With the phantom placed on a turntable, frequency-domain scattering measurements were made over 360° in 10° steps (Fig. 2b). Preliminary results based on measurements carried out with this phantom were presented with a planar monopole used as sensor\textsuperscript{33}.

### III. The Antennas

As antennas are generally characterized in terms of far-field behavior, the choice of best performing antenna for this particular imaging/sensing application is not obvious. On one hand, monopole antennas present very stable behavior and are resilient to detuning when close to human tissue. On the other hand, properly oriented Vivaldi antennas provide greater coupling to adjacent human tissue. Figure 3 shows prototypes of the three antennas\textsuperscript{36} and used for this comparative analysis. The three antennas were optimized to operate with a minimum 10-dB return loss across the 100\% bandwidth ranging from 1 to 3 GHz when immersed into a coupling medium with $\varepsilon_r$ = 12. The dielectric substrate used in all three designs is 1.58-mm thick Taconic CER10 material. Figure 3a shows a printed planar monopole. The bottom edge of the radiating element was beveled to extend a good impedance match over a larger frequency range. The front view and a section of the semi-folded monopole are depicted in Figure 3b. This antenna consists of a printed part and an external 0.2-mm thick brass element. It is worth noticing that the front area of the planar monopole is reduced by 39\% when the antenna is modified into a semi-folded geometry. Finally, Figure 1c shows the printed antipodal Vivaldi antenna. The tapering of the two arms is based on spline curves whose generating points were determined with an Efficient Global Optimization (EGO) algorithm. The antennas under test were placed against the multilayer breast phantom so to have the printed radiating element lying on the $xz$-plane and centrally placed against the phantom according to the reference axis in Figure 1a. In particular, the Vivaldi antenna is positioned in such a way that the edge corresponding to the aperture of the arms and lying on the $x$ axis is centrally placed against the multilayer system. The planar monopole, the semi-folded monopole and the Vivaldi antenna present very different coupling/radiation mechanisms. The planar monopole tends to couple energy in an omnidirectional pattern in its azimuth plane, whereas the semi-folded monopole can convey more power in one direction (towards the phantom). Finally, the Vivaldi antenna combines good matching and directive stable coupling/radiation pattern across a large frequency range.

### IV. The Imaging Procedure

The field scattered by the phantom is collected by a TX/RX antenna over $N$ scanning positions ($r_{01}, r_{02}, ..., r_{0N}$) taken uniformly around it at $N_f$ frequency bins. The data of a complete scan can be accordingly arranged in the $N \times N_f$ scattering matrix $S = [S^1 ... S^N]$, where $S^k$ is the column vector of data collected over the observation positions at $k$-th frequency. These data include the tumor signal but also strong clutter components generated from antenna’s internal
matrix over the singular function corresponding to the highest singular value. Due to reflections in the antenna and from the skin layer, it can be considered true that the first singular function mainly contains clutter contribution. This choice can be considered conservative but realistic at the same time. Consequently, the de-cluttered data matrix is obtained as

$$S_d = \sum_{k=2}^{P} \lambda_k u^k v^{kH}$$  \hspace{1cm} (1)

With the clutter removed, the scattering data vectors (i.e., the columns of $S_d$) can be approximated at the $n$-th working frequency by

$$S_n^H = H_n(f_n)A(f_n)H_T(f_n)b(f_n)$$  \hspace{1cm} (2)

In eq. (2), $H^T(f_n)$ and $H^B(f_n)$ are the TX/RX antenna responses, respectively. $b(f_n)$ is the vector of the scattering coefficients and $A(f_n)$ is the propagator which links the scatterers (located at $(r_1, r_2, ..., r_M)$) to the scattered field data. In particular, since a multi-monostatic configuration is considered, the $l$-th column of $A(f_n)$ is expressed as

$$A_l(r_l; f_n) = [G^2(r_{l1}, r_l; f_n), ..., G^2(r_{lN}, r_l; f_n)]^T$$  \hspace{1cm} (3)

where $G(\cdot; f_n)$ is the Green’s function relative to the breast host medium. Some remarks are now appropriate. First, it is noted that in eq. (2) mutual scattering between different tumors was neglected. This is actually a reasonable assumption because the breast is a highly lossy medium. Second, a multi-monostatic configuration was adopted. This choice is justified for the necessity to develop an imaging method that does not rely on the antenna response. In a multi-monostatic configuration, $H^T(f_n)$ and $H^B(f_n)$ can be considered nearly constant across the different scanning positions. However, antenna responses, $H^T(f_n)$ and $H^B(f_n)$, depend on frequency as well. Hence, as they are actually considered unknown in eq. (2), data at different frequencies cannot be coherently combined while forming the images. However, data can be separately processed at each single frequency and the corresponding reconstructions can be then suitably combined. To this end, the I-MUSIC is proposed. First, for each adopted frequency, the correlation matrix is obtained as

$$R(f_n) = S_n^H S_n^H$$  \hspace{1cm} (4)

Since $R(f_n)$ refers to a single snapshot and multiple views are not exploited, such a matrix is rank deficient with rank equal to one. Thus the pseudospectrum can be conveniently built without singular spectrum computation as

$$\Phi^n (\lambda_k; f_n) = 1/\|P^n\lambda_k(f_n)\|^2$$  \hspace{1cm} (5)

where $P^n = I - Q^n = I - R(f_n)/\|S_n^H(f_n)\|^2$ is a projector onto the noise subspace and
$A_k(f_n) = A^R(r_k; f_n) / ||A^R(r_k; f_n)||$ is the steering vector evaluated at the trial positions $r_k$ within the spatial domain $D$ under investigation. It is also remarked that, as no a priori information regarding the phantom is given, the steering vector is constructed by employing the Green’s function of an equivalent homogeneous breast medium. This is done by assigning the dielectric permittivity of the coupling medium to the equivalent breast medium in the Green’s function expression.

Rank deficiency of $R(f_n)$ entails some limitations on the achievable performance and uniqueness problems, especially with a low number of scanning positions$^{31}$. In order to overcome these limitations, pseudospectra obtained in different frequencies can be incoherently combined as suggested by Yavuz et al.$^{31}$. By doing so, the so-called wideband MUSIC is obtained

$$\Phi_{WB-MUSIC}(r_k) = 1 / \sum_{n=1}^{N_f} ||P^n A_k(f_n)||^2$$

In this contribution, single frequency pseudospectra are combined in an interferometric arrangement as done by Ruvio et al.$^{22}$. The multi-frequency pseudospectrum is overall built as

$$\Phi_{M-MUSIC}(r_k) = 1 / \prod_{n=1}^{N_f} ||P^n A_k(f_n)||^2$$

Hence, it consists in multiplying pixel by pixel the pseudospectra obtained at different frequencies. Both methods are herein compared. As beamforming methods are largely employed in this kind of application, it is interesting to consider also an algorithm taken from this class in the comparison. To this end, the following Non-coherent Migration (N-M)$^{11}$ approach is also taken into account

$$\Phi_{N-M}(r_k) = \sum_{n=1}^{N_f} ||Q^n A_k(f_n)||^2$$

In appendix it is shown that (8) actually represents a particular beamforming algorithm where the time integration windowing is not imposed. This corresponds to the fact that no information about the antenna is available. In particular, the travel time within the antenna, which is required to set the time integration window, is not known.

V. DATA ACQUISITION AND RESULTS

The phantom was scanned radially over 360° with a 10° angular step (36 scanning positions) according to a multi-monostatic radar configuration. In all experiments, the reflections from the antenna ($S_{11}$) in the frequency domain (range 1 – 3 GHz) were recorded using a Rohde&Schwarz ZVB 8 vector network analyser, which operated both as the microwave signal generator and recording device. The antennas were positioned against the phantom by using a rigid coaxial cable and Perspex holders which are transparent in the adopted frequency interval. The phantom was scanned at room temperature with the three different antennas previously mentioned. 10 frequency samples ($N_f = 10$, according to previous notation) uniformly distributed in the frequency range 1 – 3 GHz were adopted for the detection procedures. From the different recorded scans operated by using the three antennas under assessment, reconstructions were carried out with the three non-coherent methods in Table I under equivalent conditions (i.e. $N$ and $N_f$). The reconstructions corresponding to three different detection methods used to process the measurements data are shown in Fig. 4 for all the antennas under consideration. Although the target results delocalized in all the reconstructions, the I-MUSIC method offers a better performance when compared to the other non-coherent approaches for enabling more focus and better dynamic range. Together with visual reconstructions of the detection algorithms, their performance was also measured by suitable metrics such as Signal-to-Clutter Ratio (SCR), Signal-to-Measure Ratio (SMR) and Spatial Displacement (SD). The SCR and SMR compare the maximum tumor response in the reconstructed image with the maximum and the average clutter response in the same image, respectively. Whereas, the SD measures the error in tumor localization and accounts for the difference between the tumor position as peak value in the reconstruction and as actual center position in the scanned phantom. Results are summarized in Table II and confirm the superior performance achieved by I-MUSIC in reconstructions. It is worth noting that the relatively high displacement expressed in terms of SD could be expected as a consequence of the approximated equivalent breast relative permittivity used in the steering vector formation (eq. (5)). For the experimental WB-MUSIC case, metrics could not be extracted as its dynamic range is smaller than the 3-dB scale adopted for their calculation.

If measured reconstructions show the superiority of the I-MUSIC approach in relative terms, it would be useful to estimate how measurement uncertainties alter the achievable performance respect with an ideal controlled environment. To this end, experimental measurements carried out on the oil-in-gelatin phantom were benchmarked by numerical simulations of equivalent scenarios. An equivalent 2-D phantom was numerically implemented in CST Microwave Studio and scanned by the three antennas in Section III. Following the very same procedure adopted for measurements, the phantom was scanned across the same 36 angles. Reconstructions obtained from numerical data are shown in Fig. 5. Although the absence of mechanical uncertainties in the numerical environment leads to a visible disagreement with experimental results, simulations confirm the reconstructions obtained through measurements with clear outperforming outcomes achieved with the I-MUSIC and the Vivaldi antenna. SCR, SMR and SD metrics were also calculated for the numerical case and the figures are summarized in Table III.
VI. DISCUSSION

The ultimate problem that was addressed in this study was to reduce the dependence from a priori characterization of the antenna response for microwave breast cancer radar detection. With the antenna placed in a near-field scenario and coupling with breast tissues which are differently distributed from patient to patient, the definition of its response is strongly limited. Previous attempts based on a priori information extracted from measurements or numerical simulations show partial efficacy that reflects in inaccurate diagnostics. By setting a complete both experimental and numerical assessment carried out with antennas with different coupling mechanisms, the I-MUSIC approach dramatically reduces the requirement for preliminary antenna response characterization and equalization. In fact, such information was never exploited for the detection and image reconstructions. Only two assumptions were made in this study: the first one is relative to the clutter-rejection procedure by discarding the contribution corresponding to the highest singular value; the second one regards the choice of an equivalent breast permittivity equal to \( \varepsilon_r^{\text{coupling\_medium}} \) for the formation of the reference Green’s function. Both assumptions were extensively justified in section IV.

In order to further highlight the advantages of the I-MUSIC, a comparison with two linear non-coherent methods (i.e. Wide Band MUSIC and non-coherent migration) that are commonly used in the literature. With both experimental measurements and numerical benchmark, the I-MUSIC shows higher performance due to its reduced dependence from a priori, and often unfeasible, characterization. The delocalization of the detected target from the actual position is visible in Fig. 4 and it is quantified in Table II in terms of SD. This affects all detection methods under comparison in both experimental and numerical cases, although it is emphasized in measured data sets due to minor mechanical uncertainties. This is mainly due to the mismatch between the adopted equivalent breast permittivity in the reference Green’s function and the actual

Fig. 4 Reconstructions based on experimental results. Tumor marked with yellow circle.
unknown permittivity mapping of the detection scene. The dependence of this delocalization effect from such mismatch was analyzed by Solimene et al.\textsuperscript{21} in controlled canonical scenarios and it draws the attention towards further developments of the I-MUSIC approach.

VII. CONCLUSIONS

A 2-D oil-in-gelatin breast phantom was made for the evaluation of the scanning capability of the I-MUSIC technique using three antennas with different radiation/coupling properties. The I-MUSIC was also compared to non-coherent migration and WB-MUSIC approaches. Measurements and equivalent numerical simulations were carried out by scanning the phantom in a multi-monostatic configuration across 36 uniformly distributed angles. The I-MUSIC method offers better focusing capabilities and larger dynamic range between clutter and tumor levels when compared to the other algorithms under consideration. In particular, the antipodal Vivaldi antenna outperforms the coplanar and the semi-folded monopoles as it enables more pronounced artifact clutter mitigation in reconstructions. Considering the limited 1.5:1 dielectric contrast between tumor and fibroconnective tissues, the 5-mm diameter of the tumor and the independence from \textit{a priori} antenna characterization, the I-MUSIC system presents promising features for early stage breast cancer diagnostics.

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Fig. 5 Reconstructions based on numerical results. Tumor marked with black circle.


**TABLE II**

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**TABLE III**

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<td>I-MUSIC</td>
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**APPENDIX**

In this section the relationship between the Non-coherent migration and the beamforming algorithm is briefly clarified.

Let us start by the recalling the beamforming equation

\[
\Phi_{BF}(r_k) = \left| \int_{T} \left( \sum_{m} s_{dm}(t - T + \tau_m(r_k)) \right) dt \right|^2 (A1)
\]

where \( s_{dm}(\cdot) \) is the signal collected at \( y_m \), \( T = \max(\tau_m) \) (over the trial positions and the sensors positions), \( \tau_m(r_k) = 2(y_m - y_k)/v \) (\( v \) being the assumed propagation speed) and \( W = [T, T + T_W] \) is the time integration window. In practical situations time \( T \) should also account for the time delay due to the propagation inside the antenna structure. By accepting that no information on the antenna is available then choice of \( W \) is not immediate. Here, time windowing is completely discarded so that (A1) becomes

\[
\Phi_{BF}(r_k) = \int_{T} \left( \sum_{m} s_{dm}(t - T + \tau_m(r_k)) \right) dt (A2)
\]

which by Parseval’s formula can be rewritten as

\[
\Phi_{BF}(r_k) = \int_{B} \left| \sum_{n} S_{dm}(f) \exp[j2\pi f \tau_m(r_k)] \right|^2 df (A3)
\]

where \( B \) is the signals frequency band and \( S_{dm}(f) \) denotes the Fourier transform of \( s_{dm}(t) \). By accounting for that, only a discrete set of frequencies is employed and by recalling the Green’s function asymptotic behavior, (A3) can be recast as

\[
\Phi_{BF}(r_k) = \sum_{n=1}^{N_f} \left\| \overline{A}_{nk} \right\|^2 (A4)
\]

where \( \overline{A}_{nk} \) is the Green’s function column as in (3) once the amplitude spreading terms has been compensated. Now, it is to realize that (A4) and (8) coincide apart from some normalizing weights defined in eq. (5).
