Low-Cost Microwave Imaging Portable Device for Breast Cancer Diagnosis

Alexandre de Jesus Aragão (✉ alexandreja@ifsp.edu.br)

Escola Politecnica da USP: Universidade de Sao Paulo Escola Politecnica <https://orcid.org/0000-0003-0660-8327>

Dionisio de Carvalho

Escola Politecnica da USP: Universidade de Sao Paulo Escola Politecnica

Bruno Sanches

Escola Politecnica da USP: Universidade de Sao Paulo Escola Politecnica

Wilhelmus A. M. van Noije

Escola Politecnica da USP: Universidade de Sao Paulo Escola Politecnica

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Low-cost Microwave Imaging Portable Device for Breast Cancer Diagnosis

Alexandre de Jesus Aragão^{1,2} · Dionisio de Carvalho¹ · Bruno Sanches¹ ·
Wilhelmus A. M. Van Noije¹

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Abstract This paper presents a Microwave Imaging apparatus for breast cancer detection, aiming at early diagnosis, an effective way of reducing mortality rates. In 2020, breast cancer has surpassed lung cancer cases worldwide, with more impact in low-income countries, which motivates seeking a non-invasive and affordable screening equipment. Focusing on low-cost and portability, a US\$150.00 embedded hardware transceiver plus antennas platform has been picked out, together with the development of a dry-coupling silicone rubber bra hold by a handheld device. Initially, a losses study to analyze the hardware ability to detect malignant tumors depth in the breast was conducted. Next, simulations and tests employing the platform along with a breast phantom were carried out. This phantom mimics the dielectric breast tissue properties while a confocal algorithm was applied to backscattered signals to generate reconstructed images. Results established a correspondence between simulated and experimental data, SCR

and SMR above 7.0 dB and 23.0 dB, respectively, in the resulting images and tumor location precision below 0.3 cm. Based on that, the low-cost portable proposed system results showed its ability as an adjunct early breast cancer diagnosis tool.

Keywords Microwave imaging · Breast cancer detection · Ultra wideband radar · Breast phantom · Confocal algorithm · Dielectric constant

1 Introduction

Microwave Imaging (MI) for breast cancer diagnosis has evolved during the last two decades as a useful adjunct noninvasive technique over current well-established methods in medical screening diagnoses. Likewise, recent trials results are prospecting acceptance of this modality in future clinical practice [1–4]. Most of the MI arrangements employ a microwave transceiver connected to a set of antennas which collects backscattered signals. Reconstructed images are generated from permittivity contrast exploitation between breast internal tissues and malignant tumors: a tomographic approach provides a quantitative dielectric profile, whereas a radar-based choice lays out a qualitative breast intensity map. MI advantages rely on the facts that microwave signals are non-ionizing and suitable to dense breasts, which is not achievable with mammography [5].

The World Health Organization (WHO) has reported that in 2020 breast cancer surpassed lung cancer statistics, reaching over 2.2 million new cases [6]. Furthermore, according to WHO, breast cancer mortality rate distribution is worrisome: 12.1% per 100,000 inhabitants in high Human Development Index (HDI) countries against 20.1% in low HDI ones. Late diagnosis due to scarcity of early detection programs, along with the

Alexandre de Jesus Aragão
E-mail: alexandrejaragao@usp.br
<https://orcid.org/0000-0003-0660-8327>

Dionisio de Carvalho
<https://orcid.org/0000-0001-8511-3896>

Bruno Sanches
<https://orcid.org/0000-0002-5717-5800>

Wilhelmus A. M. Van Noije
<https://orcid.org/0000-0002-9010-3083>

¹ Department of Electronic Systems Engineering of the Escola Politécnica, USP, São Paulo, Brazil.

² Department of Electrical Engineering of Federal Institute of Education, Science and Technology, São Paulo, Brazil.

lack of adequate treatment, both leading to delayed actions represent the primary causes of these disparities. In fact, if the tumor is identified in its initial stage, the cure rate is augmented relevantly. One-third of mammography detected breast cancers had less than 1.0 cm tumor and no evidence of axillary lymph node metastases [7].

MI systems are potentially low-cost and portable. Taking advantages of these characteristics, their development could increase the diagnosis offer to the socially vulnerable population. In this work, the conduction of a breast cancer, dry-setup MI system realization using a low-cost platform is presented. The starting point was the front-end loss scenario simulation, targeting key parameters determination for hardware specifications. These resulting data allowed the choice of a Commercially available Off-The-Shelf (COTS) low-cost hardware platform. A silicone rubber anatomic bra was used as antenna breast dielectric coupling medium, also regarding mechanical flexibility, comfort and hygiene to the patient. The portable handheld system was simulated, and experimental tests were performed employing a breast phantom assembled with materials that mimic the internal dielectric characteristics of a real breast. Imaging results were obtained applying a radar-based DAS algorithm and quality metrics were employed to assess the device performance.

2 Materials and Methods

2.1 System Outline

Fig. 1 depicts the system hardware block diagram along with the front-end model proposal.

Fig. 1: Breast cancer examination device hardware block diagram and front-end model.

The apparatus was idealized to be standalone and contained in a handheld support that fits in a custom developed bra (Fig. 2a). A housing in the bra allows

the handheld system to be manually rotated around the breast (Fig. 2b). The hardware platform is controlled and configured by custom firmware, communication protocol and software. The transceiver is triggered by the software, transmitting Impulse Radio Ultra Wideband (IR-UWB) monopulses. The backscattered signals are collected and then applied to an Imaging Unit (UI) that runs the improved confocal algorithm [8]. This procedure should be replicated in different angles, surrounding the whole breast. An energy map image is generated when all predefined positions are performed.

Fig. 2: Front-end and phantom mounting details.

2.2 Front-End

The front-end system (Fig. 1) consists of a heterogeneous breast model that both in the simulation and experimental environments intends to represent a patient examination condition. The model was shaped as a parabola sloper, which corresponds to a pattern of real bra moldings [9]. The chosen size was 34B (imperial system), which fits most women breasts [10]. The base diameter is 15.0 cm, and the internal structure thicknesses considered in the base dimensions were: 0.2 cm of skin, 8.6 cm of adipose material and 6.0 cm of glandular tissue, where the lactiferous lobes are concentrated. This model was designed based on the ordinary mammographic anatomy of a real breast [11]. The glandular tissue was considered as a solid block, thinking of the worst-case scenario of pregnancy and lactation.

All the simulations were performed in CST Studio Suite®, as shown in Fig. 3, which illustrates the 3D cross section scenario. The applied breast internal structures dielectric characteristics were from the simulation software bio tissue library. The bra and hardware support are silicone rubber made, and its dielectric profile was extracted from [12]. The tumor dielectric characteristics were extracted from [13]. A relative permeability of one was adopted to all materials, according to [14].

Fig. 3: Simulated front-end 3D cross section.

Experimental tests were performed using a breast phantom which employs materials with dielectric constants similar to real breast tissues [15]: Skin was simulated with a sorbitol layer (Fig. 2c), adipose tissue was mimicked with sand and liquid glycerin took the glandular area place. The tumor simulation was made using polyacrylamide (PAM) spheres as shown if Fig. 2d.

2.3 Channel Losses

To evaluate the spreading and material losses, a front-end performance study was carried out based on the plane wave equations for lossy dielectric media. Also considered were waves transmission and reflection at normal incidence, both from [16,17].

Eqs. (1) and (2) calculate the losses due to spreading and materials attenuation (Fig. 4a):

$$\frac{P_{ra}}{P_{ta}} = \frac{G_{tx}G_{rx}\lambda^2}{(4\pi)^2D^2} \quad (1)$$

where P_{ra} is the power at the receiving antenna, P_{ta} is the power from the transmitting antenna, G_{tx} and G_{rx} are respectively the transmitting and receiving antenna gains, λ is the medium wavelength and D is the distance between antennas;

$$\alpha = \omega \sqrt{\frac{\mu\epsilon}{2} \left[\sqrt{1 + \left(\frac{\sigma}{\omega\epsilon}\right)^2} - 1 \right]} \quad (2)$$

where α is the medium attenuation factor, ω is the angular wave velocity, μ , ϵ and σ are respectively the material permeability, permittivity and conductivity.

Eqs. (3) and (4) estimates the loss due to signals traveling over the different breast tissues (Fig. 4b):

$$\Gamma_{(a,b)} = \frac{\eta_b - \eta_a}{\eta_b + \eta_a} \quad (3)$$

$$T_{(a,b)} = \frac{2\eta_b}{\eta_b + \eta_a} \quad (4)$$

where Γ and T are respectively the reflection and transmitting wave coefficients, η_a and η_b are respectively the calculated a and b materials impedances (Eq. 5):

$$\eta_n = \sqrt{\mu_n/\epsilon_n} \quad (5)$$

(a) Simulated setup.

(b) Front-end loss setup [15].

Fig. 4: Losses study scenario: 2.0 mm bra coupling media, 2.0 mm skin layer and variable thickness fatty tissue to analyze losses versus breast depth.

The Radar Cross Section (RCS) was calculated applying Spherical Bessel functions using series approximation and recursion. Although exact methods of RCS prediction are very complex even for simple shape objects, approximate methods become the viable alternative to analyze RCS behavior [17].

2.4 Hardware Platform

Radar-based MI exams can be implemented either in frequency or time domain. Although the latter is more cost effective, with reduced scan time, the former confers higher Signal to Noise Ratio (SNR). The time domain SNR drawback is caused by the high frequencies attenuation through the breast tissue, which is overcome by means of increasing measurement averaged repetitions and using a lower frequency range [18].

Another concern is the system antenna set, which can be fixed (stationary array), rotating (synthetic array) or entire device rotation (hardware array). Seeking for low-cost, the hardware array set with the antennas embedded in the Printed Circuit Board (PCB) facilitates the signal behaviors analysis, because the components and circuits are the same for all sweeping, thus there are no inherent errors due to mismatching in antennas or circuits. Moreover, this set operating in time domain eliminates the mutual coupling between antennas, which would be a more complex inverse problem.

Based on the aforementioned, the chosen hardware was a US\$150.00 UWB platform from Novelda[®] [19] which complies with the United States Federal Communications Commission (FCC) part 15 [20]. It comprises an IR-UWB transceiver controlled by a 32-bits microcontroller, a Power Management Unit (PMU), embedded Rx and Tx antennas, RAM memory and USB communication. Table 1 summarizes its main characteristics. Firmware, communication protocol and a software that interacts with the end user and runs the imaging algorithm were developed for the system achievement.

Table 1: Hardware platform characteristics [15].

Transmitting Signal	Gaussian monopulse
Derivative Order	11th
Central Frequency	6.40 GHz
Bandwidth	2.17 GHz
Receiver Sensitivity	-87.32 dB
Dynamic Range (DR)	64.00 dB
ADC ENOB	10.34 bits

2.5 Dry-Based Breast Coupling

Fig. 2 depicts a dry-based breast coupling proposal. This setup reduces the signal attenuation caused by the dielectric constant difference between the air and skin. Instead of immersing the breast into a liquid saline or glycerin based solution, a silicone rubber bra is used as the antenna-breast coupling medium. The dielectric characteristics of this material resembles the breast internal fatty tissue [12]. A 5.65 dB of loss has been reported using Matlab[®] in a scenario employing a 0.2 cm silicone rubber piece between the antenna and the breast. It is less than a half compared to the air dielectric coupling (12.20 dB calculated loss).

The following advantages also arise from a dry setup design with silicone rubber:

- It is a low-cost and high availability raw material.
- Easy to format and malleable, allows bra construction according to best fit to the patient, involving the entire breast, introducing comfortable conditions to the exam.
- It is a synthetic polymer that brings more hygiene to the exam procedure.

2.6 Improved Confocal Algorithm

Among the time domain radar-based image reconstruction algorithms, Delay and Sum (DAS) [21] based solutions have proved to be robust with low computational cost. Moreover, several proposals have achieved efficient results even with the heterogeneous dense breast consideration [8, 22].

This work applies the DAS Improved Confocal Algorithm proposed in [8] to the image reconstruction. The improvements include the negative wave cycle rectification and cardioid factor compensation, which provides higher tumor contrast in heterogeneous breasts. Besides, filtering steps have been added due to the changes in the applied signal derivative order: a time window filtering that considers the distance between the antenna and the breast radius and a filter that removes the undesired side lobes from the derivative Gaussian monopulse, highlighting the backscattered main lobe.

2.7 Data Acquisition and Analysis

All the experiments were performed with the handheld model developed: hardware platform and breast phantom (Fig. 2). The operational central frequency was 6.4 GHz with 2.17 GHz bandwidth, and the signals were applied in 64 equidistant angles. For each point, the signal collected is the mean of 1,000 consecutive measurements in order to increase the response SNR. The 1.0 cm diameter mimicked tumor PAM sphere was buried centrally in front of the first antenna position, 3.0 cm distant from the hardware platform central point. In a second scenario the tumor was buried 2.0 cm distant from the hardware in the lower part of the breast, near the muscle area. The sweeping is real-time, and the necessary mean time for changing the sweeping position is estimated in about 30 s.

The Signal to Mean (SMR) and Signal to Clutter Ratio (SCR) quality metrics were employed to quantitatively analyze the images produced by both simulation and experimental scenarios:

$$SMR = 20 \log \left(\frac{I_{max}}{I_{mean}} \right) \quad (6)$$

where I_{max} and I_{mean} are respectively the maximum and mean image pixel intensity values, and:

$$SCR = 20 \log \left(\frac{I_{max}}{I_{clutter}} \right) \quad (7)$$

where $I_{clutter}$ is the maximum image pixel intensity value in a non tumor region.

3 Results

3.1 Losses Analysis

Fig. 5 depicts the loss analysis due the operational central frequency versus breast depth, as discussed in Sec. 2.3 by Eqs. (1), (2), (3), and (4). It was noticed that loss by penetration increases in a mean factor of 4.7 dB/GHz. Frequencies below 5.0 GHz can achieve more than 6.5 cm reach with losses of less than 50.0 dB.

Dist.(mm) Freq.(GHz)	10	15	20	25	30	35	40	45	50	55	60	65
2.0	14	17	19	22	27	28	30	32	35	36	38	38
2.5	16	18	21	24	28	30	32	33	35	35	38	39
3.0	18	23	26	30	30	32	34	34	35	35	38	40
3.5	20	23	27	30	33	35	36	36	36	37	40	43
4.0	21	22	26	29	35	39	40	40	39	39	42	46
4.5	23	23	26	30	33	39	45	43	40	41	44	49
5.0	21	25	26	29	31	37	48	47	43	44	46	52
5.5	20	25	28	29	32	37	45	55	47	47	50	57
6.0	22	29	32	32	34	38	45	67	53	52	54	61
6.5	27	35	38	38	40	46	46	59	60	58	59	63
7.0	32	57	55	54	46	46	50	56	63	64	65	64
7.5	34	40	42	50	58	54	54	56	60	66	70	66
8.0	32	36	38	44	47	49	54	57	59	65	73	67
8.5	29	34	38	43	44	47	53	58	59	65	76	71
9.0	30	35	39	44	46	48	54	60	62	66	72	76
9.5	33	39	43	49	50	53	58	62	66	66	67	76
10.0	41	47	52	59	61	70	66	63	73	67	64	72

Fig. 5: Simulated intensity map: loss(dB) vs. frequency(GHz) vs. distance(mm). Green = 20dB, yellow = 50dB, red = 80dB.

Fig. 6 shows the calculated loss due to the RCS effect for several spherical tumor diameters. Data indicate that the loss decreases in a mean factor of 3.6 dB/GHz. A 7.0 GHz operational frequency would reduce loss in 20.0 dB for a 0.2 cm diameter tumor, for example, compared to 3.0 GHz.

Fig. 6: Calculated RCS loss vs. frequency vs. tumor diameters.

3.2 Measurements Results

Figs. 7a and 7b illustrates respectively the simulated and experimental reconstructed images for the tumor buried 3.0 cm depth in the breast. The calculated SCR and SMR for the simulated resulting image are respectively 8.7 dB and 30.3 dB. Likewise, 7.6 dB and 23.7 dB are the experimental calculated SCR and SMR.

The resulting image for the second experimental scenario is shown in Fig. 7c, where the calculated SCR was 8.0 dB and the SMR was 28.7 dB.

These results show firstly a relevant correspondence between simulated and experimental data: tumor locations are just about superposed (less than 0.3 cm offset). Secondly the tumor presence and position both in the experimental and simulated scenarios were detected, endorsing the device ability. Thirdly, the result-

ing images present SMR and SCR compatible with the chosen algorithm and other systems [15, 21, 22].

Fig. 7: Simulated and experimental measurement results. Dimensions in cm.

4 Discussion

The encouraging clinical evidences of breast cancer detection success using MI have brought several systems proposals with distinct approaches. Most of them toward non-portable equipment, whereby microwave signals are generated and acquired using Vector Network Analyzers (VNA) [5]. Consequently, the flexibility in frequency and power adjustments provides high precision with the onus of non-portability and equipment cost reaching tens or even hundreds of thousands US Dollars. This work proposed a low-cost and portable time-domain hardware platform choice, and the constraints arising from these options have been analyzed.

The entire system has been developed focusing clinical usage. The employment of silicone rubber as the antenna-breast coupling media brings robust mechanical characteristics to the project, and more comfort and hygienic conditions to patients during the exam, compared to the other existing methods. The hardware selection takes advantage of a standalone approach, with antennas, transceiver and radio frequency circuits embedded in a unique PCB, bringing portability to the device and increasing its robustness.

We analyzed the hardware ability to cope with the microwave signals losses crossing several breast tissues. RCS losses increase significantly as the tumor diameter reduces, as shown in Fig. 6; the specified 64 dB hardware DR is not met in tumor diameters below 0.4 cm. Nonetheless, the employment of higher frequencies circumvents this limitation, although the spreading and material losses results thwart this approach, as depicted in Fig. 5. In fact, increasing frequency takes to better tumor resolution reducing the signal penetration depth. For example, a minimum 5.3 GHz central operational frequency would detect a 0.8 cm diameter tumor in a 12.0 cm diameter breast. Indeed, frequencies higher than 7.0 GHz presented attenuation near the proposed hardware DR for breast depth above 4.0 cm. Considering these results, the hardware platform 6.4 GHz central frequency presented a reasonable tradeoff between penetration and resolution.

Finally, according to [7,23], 80% of breast cancers are located in the lactiferous ducts area, and tumors reaching up to 1.0 cm in most cases do not present metastasis. Based on that, the low-cost portable proposed system results showed its ability as a complementary early breast cancer diagnosis.

The research next steps include more different scenarios tests with several tumors sizes and locations, the inclusion of rotation sensors in the handheld, so that the examination position can be informed along with the backscattered signal. Also, a Bluetooth Low Energy protocol implementation allowing signals to be transmitted directly to a mobile device application and then a cloud computing IU for imaging reconstruction. The handheld along with Internet of Things facilities takes the project towards a product concept.

5 Conclusion

This work presented the implementation of a low-cost portable device for breast cancer screening using IR-UWB microwave signals. It comprises a US\$150.00 embedded antennas-transceiver hardware platform along with a silicone rubber bra that serves as breast mechanical and dielectric coupling, all integrated into a handheld system. The time-domain DAS Improved confocal algorithm generates an image by processing the backscattered signals. A front-end loss study together with system simulations and experimental tests employing a developed breast phantom have shown the hardware ability of overcoming losses inserted by the several breast tissues. Although frequencies higher than 3.0 GHz reduce the RCS losses, spreading and material propagation losses indicate that frequencies over 7.0 GHz do not present reasonable results for breast

depth above 4.0 cm. The resulting images presented SCR and SMR above 7.0 dB and 23.0 dB, respectively, and tumor location precision below 0.3 cm. The results revealed the system as a potential early diagnosis breast cancer auxiliary tool. The urgency in the research and development of new, less aggressive imaging techniques to the patients is relevant nowadays, mainly to reach low-income countries. This effort could enhance breast cancer diagnosis offer in medical facilities and, consequently, promote an increase in the number of women monitored.

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Conflict of Interest

The authors declare that they have no conflicts of interest.

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Figures

Figure 1

Breast cancer examination device hardware block diagram and front-end model.

(a) Complete apparatus.

(b) Bra and handheld.

(c) Bra and Sorbitol layer.

(d) Sand, glycerin and PAM.

Figure 2

Front-end and phantom mounting details.

Figure 3

Simulated front-end 3D cross section.

Figure 4

Losses study scenario: 2.0 mm bra coupling media, 2.0 mm skin layer and variable thickness fatty tissue to analyze losses versus breast depth.

Dist.(mm) Freq.(GHz)	10	15	20	25	30	35	40	45	50	55	60	65
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2.5	16	18	21	24	28	30	32	33	35	35	38	39
3.0	18	23	26	30	30	32	34	34	35	35	38	40
3.5	20	23	27	30	33	35	36	36	36	37	40	43
4.0	21	22	26	29	35	39	40	40	39	39	42	46
4.5	23	23	26	30	33	39	45	43	40	41	44	49
5.0	21	25	26	29	31	37	48	47	43	44	46	52
5.5	20	25	28	29	32	37	45	55	47	47	50	57
6.0	22	29	32	32	34	38	45	67	53	52	54	61
6.5	27	35	38	38	38	40	46	59	60	58	59	63
7.0	32	57	55	54	46	46	50	56	63	64	65	64
7.5	34	40	42	50	58	54	54	56	60	66	70	66
8.0	32	36	38	44	47	49	54	57	59	65	73	67
8.5	29	34	38	43	44	47	53	58	59	65	76	71
9.0	30	35	39	44	46	48	54	60	62	66	72	76
9.5	33	39	43	49	50	53	58	62	66	66	67	76
10.0	41	47	52	59	61	70	66	63	73	67	64	72

Figure 5

Simulated intensity map: loss(dB) vs. frequency(GHz) vs. distance(mm). Green = 20dB, yellow = 50dB, red = 80dB.

Figure 6

Calculated RCS loss vs. frequency vs. tumor diameters.

(a) Scenario 1 simulated.

SCR = 8.7 dB, SMR = 30.3 dB.

(b) Scenario 1 experimental.

SCR = 7.6 dB, SMR = 23.7 dB.

(c) Scenario 2 experimental.

SCR = 8.0 dB, SMR = 28.7 dB.

Figure 7

Simulated and experimental measurement results. Dimensions in cm.