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Challenges in the Design of Microwave Imaging Systems for Breast Cancer Detection

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***Abstract*—Among the various breast imaging modalities for breast cancer detection, microwave imaging is attractive due to the high contrast in dielectric properties between the cancerous and normal tissue. Due to this reason, this modality has received a significant interest and attention from the microwave community. This paper presents the survey of the ongoing research in the field of microwave imaging of biological tissues, with major focus on the breast tumor detection application. The existing microwave imaging systems are categorized on the basis of the employed measurement concepts. The advantages and disadvantages of the implemented imaging techniques are discussed. The fundamental tradeoffs between the various system requirements are indicated. Some strategies to overcome these limitations are outlined.**

***Index Terms*—biomedical imaging, medical diagnostic imaging, microwave imaging, microwave circuits, microwave antennas.**

I. Introduction

The last decade has seen a significant increase in research concerning microwave-based systems for detection of breast cancer [1]-[11]. Some of the modalities even have gained sufficient maturity to be tested in clinical environments [9]-[11]. The recent interest in microwave imaging for this purpose is mainly driven by the improved performance and wide availability of low cost microwave devices, the rapid increase in computational power for calculation of complex electromagnetic problems, the improvement of human body models, and the increased number of reported electromagnetic properties of human tissue [12]-[14].

Although microwave imaging has the potential to offer improved sensitivity and specificity, a number of challenges exist. The breast tissue is heterogeneous, resulting in a complex field distribution in the body. Efforts should be made towards suppression of clutter due to interferences from the breast skin, nipple, chest wall, and so on. Normal breast tissue is lossy at microwave frequencies, and tumors are very small in cases of early detection. The losses introduce a tradeoff between spatial resolution and penetration depth. Employing higher frequencies to obtain better resolution and to allow the use of small antenna elements results in lower electromagnetic field penetration inside the lossy biological tissue. In addition, the higher resolution enlarges the size of the corresponding electromagnetic problem, which leads to an increase in the computational time required to reconstruct the image. Individual anatomical features of the patient also add a certain degree of complexity to the imaging procedure.

In addition to the difficulties mentioned above, each imaging technique also faces specific challenges related to the system configuration and imaging scenario, since different methods are based on different physical effects.

Microwave imaging systems for breast cancer detection can be divided into two major categories: active and passive systems (refer to Fig. 1).

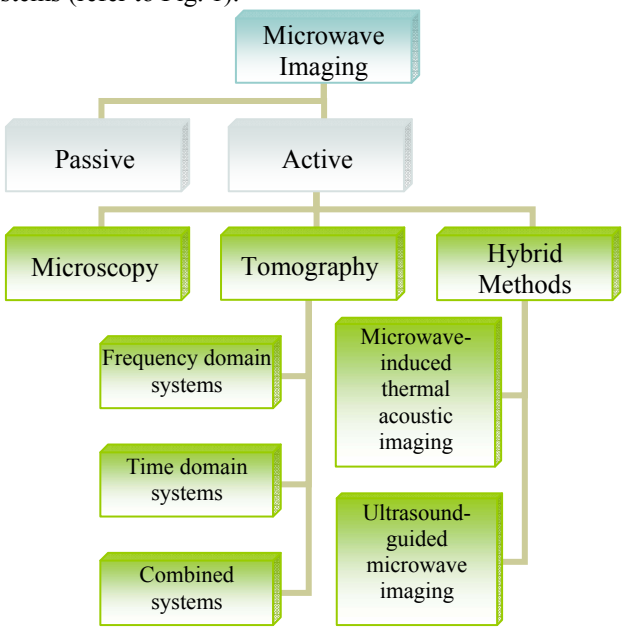

Figure 1. Microwave imaging systems for breast cancer diagnosis

In active systems, the sensing is accomplished by probing the biological object with self-generated energy. In passive systems, on the other hand, this energy is generated by the object. Active microwave imaging, in turn, can be divided into microwave tomography and ultrawideband (UWB) radar techniques, microwave microscopy, and hybrid modalities, which will be discussed next.

II. Microwave Radiometry

Passive microwave imaging is usually referred to as microwave radiometry and is based on the measurement of the electromagnetic field spontaneously emitted by warm bodies according to Planck's law:

$$I(f,T) = \mu\varepsilon \frac{2hf^3}{e^{\frac{hf}{kT}} - 1}, \tag{1}$$

where I is the spectral radiance of electromagnetic radiation for a medium with permeability μ and permittivity ε, f is the frequency, T is the temperature of the black body, $h = 6.63 \cdot 10^{-34}$ J·s, $k = 1.38 \cdot 10^{-23}$ J/K, and $e = 2.71$.

According to this equation the radiation distribution depends on the frequency and also on the temperature. Therefore, radiometry is often called thermography.

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Temperature is the key parameter that is used here to predict the presence of malignancy [9], [15], [16]. There are several contributing factors to the temperature elevation related to the tumor presence: the malignant cells are more metabolically active and produce more heat, they have a reduced thermoregulatory capacity, and it is recognized that localized increases in blood volume can be associated with early tumor growth. The last factor is especially important, because it allows early breast cancer detection using microwave imaging. The thermal noise power measured by the radiometer is related to the local temperature distribution in the breast, allowing for its reconstruction from the data collected from various antenna positions (refer to Fig. 2).

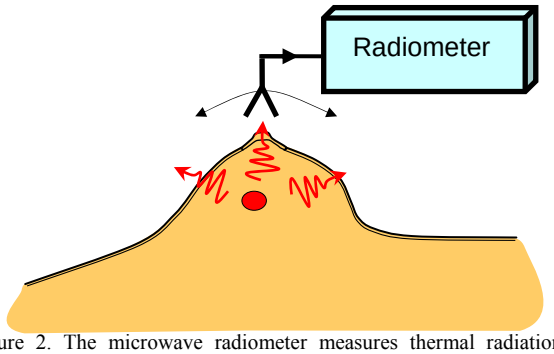

Figure 2. The microwave radiometer measures thermal radiation emitted by tumor.

According to (1), the peak of electromagnetic radiation is in the infrared range, but high tissue attenuation at these frequencies limits the application of the infrared thermography to measurements of skin temperature. In the microwave frequency band, the intensity of radiation is about ten million times less but tissue attenuation is also considerably lower. Furthermore, the power P radiated in this band is directly proportional to the brightness temperature T_B on the absolute scale:

$$P = k\, T_B\, \Delta f \quad , \tag{2}$$

where Δf is the system bandwidth.

Passive techniques have a number of benefits. Among these are that the patient and medical personnel are not exposed to electromagnetic radiation and that the method can be used for detecting breast cancer in men, which can not be done with mammography.

The main challenge to this method is to detect a very low level power radiated by tumors, which raises technical problems. A solution could be to use cooling systems to reduce the temperature of the microwave detector [15].

The other difficulty concerns the estimation of the spatial temperature distribution inside the body. The single frequency radiometry enables measurement of average temperature of a certain area. Therefore, it is hard to distinguish between a cool target close to the skin and a hot target located deep in the breast. The measured brightness temperature in these two cases may be the same although the targets are quite different. This problem can be solved by using microwave radiometry at different frequencies [17]. The method is based on dispersive properties of the tissue: (the intensity of thermal radiation increases at higher frequencies and, at the same time, the penetration depth into biological tissues decreases). Careful analysis of the measured radiometric data at several frequencies allows estimation of the depth and the size of the heat source.

III. Microwave Tomography

The word tomography comes from the Greek words "to slice" (tomos) and "to write" (graphein). This term comes to microwave imaging from the image processing algorithms and is also known from magnetic resonance tomography (MRT) and X-ray computed tomography (CT), where the image of the internal structures of a body is represented slice by slice. Today, the application of active microwave imaging methods to visualize the interior of the body is also often referred to as microwave tomography in spite of their ability to directly acquire three-dimensional (3D) images.

A. Frequency Domain Systems

Frequency domain systems are based on inverse scattering techniques, in which a microwave transmitter illuminates an object and scattered fields at numerous locations are obtained from the measurements by subtracting the incident field (refer to Fig. 3). Using this information the electromagnetic properties of the body are reconstructed. This electromagnetic problem is governed by the wave equation

$$\nabla \times \nabla \times \vec{E}(\vec{r}) - \omega^2 \mu_0 \varepsilon(\vec{r}) \vec{E}(\vec{r}) = -j\omega\mu_0 \vec{J}(\vec{r}), \tag{3}$$

where $\vec{E}(\vec{r})$ is the electric field, $\vec{r}$ is the spatial coordinate, $\varepsilon(\vec{r})$ is the unknown distribution of complex permittivity, $\vec{J}$ is the source current density, $\omega = 2\pi f$, $\mu_0 = 4\pi \cdot 10^{-7}$ H/m, and j is the imaginary unit, assuming the time factor $e^{-j\omega t}$.

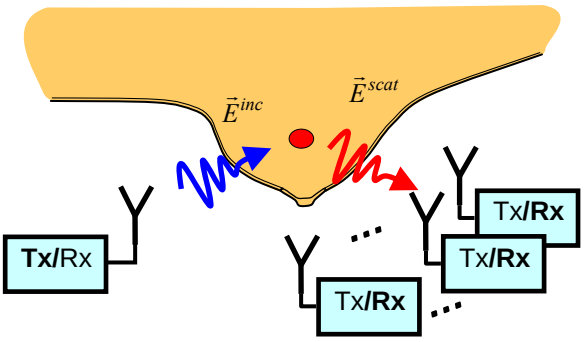

Figure 3. A number of microwave transceivers is used to measure the incident and scattered fields.

The inverse problem is to determine the position and permittivity of the scatterer (e.g, the distribution of complex permittivity $\varepsilon(\vec{r})$) from the measured scattered fields.

The solution to this problem is usually carried out by means of an optimization procedure, reducing the difference between the measured and calculated data using a forward solution of (3).

The overall radiated power for such systems is much lower than that from a typical cell-phone transmitter aiming to reduce exposure to electromagnetic radiation as much as possible. Of course, the spatial resolution of microwave tomography cannot compete with the spatial resolution achieved with CT, simply because of the large difference in wavelength. Nevertheless, the high dielectric contrast at microwave frequencies makes these instruments very sensitive to the presence of malignant tissue.

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Attempts to solve the fullwave 3D inverse scattering problem in an exact manner result in a high computational cost, which is presently the main problem for such systems. However, this problem can be solved already in the next decade taking into account the current rate of development in computer technology.

In practice, field measurements are carried out using two (transmitting and receiving) antennas in bistatic configuration and mechanical scanning [3], [18], [19] or an array of antennas, scanned electronically [1], [2], [20], [21]. In the first case, having obtained the scattered signal data for a given antenna location, the results are stored in the data processor and the antenna is moved to a new position. The measurement procedure is then repeated for a number of antenna locations. The mechanical scanning systems, in particular in bistatic configuration, usually suffer from a long data acquisition time (up to several hours), which requires high accuracy of antenna positioning and stability of the system. The optimal acquisition time for an imafing system is, however, equal or less than the patient's respiration cycle to ensure the stability of the object to image. This can be achieved with electronically scanned antenna arrays, as it is shown in Fig. 3. In this configuration each antenna operates in either transmit or receive mode in order to maximize the amount of measurement data that can be recorded. Such a configuration provides reduced measurement time in comparison to mechanical scanning, and involves minimal discomfort, so the procedure is acceptable to the patient.

A highly sensitive receiving system is required to provide a careful measurement of the scattered field component. High sensitivity can be achieved using superheterodyne receiver architecture with careful filtering of the detected signal. The dynamic range of the reported imaging systems is more than 120 dB [1].

In many microwave tomographs a coupling medium between the antennas and the body is introduced in order to avoid strong reflections from the air/skin interface. Therefore the patient orientation for these systems (Fig. 3) is different from one in microwave radiometry (Fig. 2) with the patient lying in a prone position with her breast suspended down into the tomograph. The electrical properties of the medium are chosen to be close to the properties of the body to enhance the coupling of electromagnetic energy into the breast. In practice however, the electrical properties of the medium depends on temperature, and any temperature drifts and unpredictable local temperature gradients affect the measurement accuracy of the system.

A photograph of an active microwave 3D imaging system prototype that uses frequency-domain measurements in conjunction with multi-channel receiver architecture and electronic scanning is shown in Fig. 4 [8].

The microwave imaging system contains 32 measurement channels. Each channel can operate in transmit and receive mode, feeding one transmitting antenna at a time and measuring the scattered field at all other antennas operating as receivers, respectively. The antennas are submerged in a tank with a glycerin-water coupling liquid which mimics the electrical parameters of the breast. To irradiate an imaging domain, 32 antennas positioned in a cylindrical setup with a radius of 8 cm are used as shown in Fig. 4. The antennas are oriented horizontally and positioned in 4 rows with 8 antennas in each row. The antenna system is designed assuming that, during the examination, the patient lies prone atop a measuring tank with one breast pendant in the imaging antenna array.

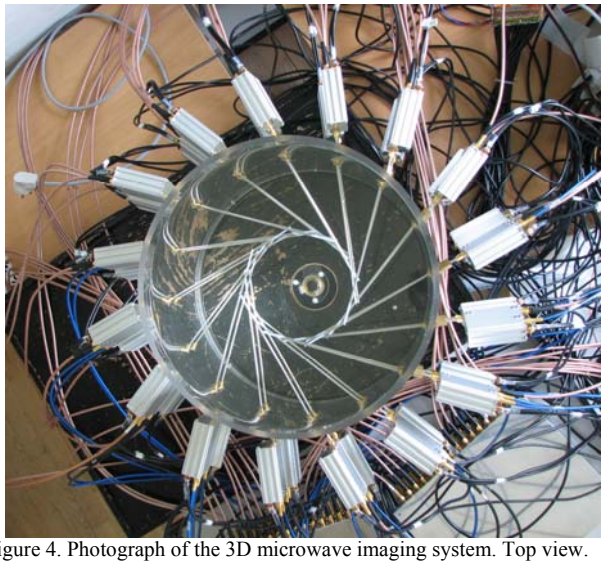

Figure 4. Photograph of the 3D microwave imaging system. Top view.

The alternative way to measure the scattered field from the breast is a modulated scattering technique [22]-[24].The system consists of two large horn antennas, as shown in Fig. 5.

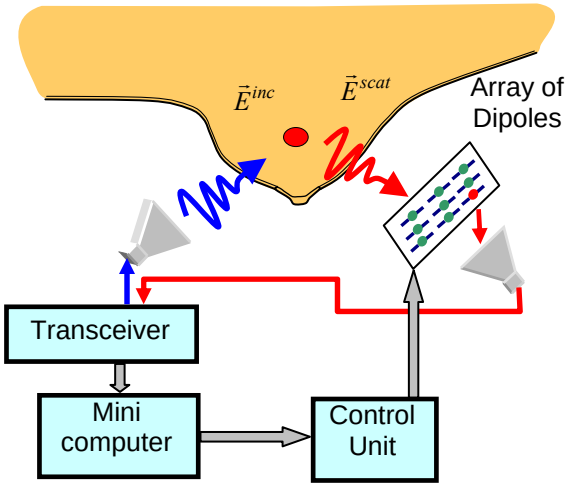

Figure 5. In a modulated scattering technique, the detected by transceiver signal is proportional to the electric field at the position of the selected dipole.

The transmitting antenna illuminates the investigation area. The measured scattered field is provided by the probe array of the imaging system, which is placed in front of the collector aperture. The probe array is a dipole array, with a step size of half the wavelength in the coupling medium (which is usually based on water solutions) [22]. Each array element is loaded with a diode. Modulation of the diodes results in a signal at the output of the collector aperture. The measured signal is proportional to the field at the position of the selected dipole. The array scanning is rapidly performed in a sequential way. This way of measuring the electromagnetic field distribution provides a high data

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acquisition rate.

B. Time Domain Systems

Time domain techniques are often referred to as UWB radar techniques. The approach originates from military and ground-penetrating radar (GPR) applications and was proposed for breast cancer detection in the late nineties. They employ a well developed radar principle, where radiated low power short pulses are received at various locations with a probe antenna or alternatively by an array of antennas. A time delay between radiated and received pulses as well as the shape of the received signal contain information about the scatterer. The processed signals for various locations of a probe antenna or from array elements are combined to form a 2D or 3D image showing the location of a highly reflective object representing malignant tissue.

Due to the mentioned contrast in dielectric properties between normal and malignant tissue the tumor microwave scattering cross-section is larger than that of an equivalent size normal breast tissue [4].

Similarly to frequency domain systems there are three major system configurations: monostatic, bistatic, and multistatic. In the monostatic configuration, the transmitter is also used as a receiver and is mechanically moved across the breast to form a synthetic aperture. In the bistatic configuration, one transmitting and one receiving antenna are used as a pair and moved across the breast to form a synthetic aperture. In the multistatic setup, an array of antennas is used for data collection.

UWB radar techniques have the advantage of constituting a simple approach to locate strong scatterers in the breast, avoiding full-wave electromagnetic analysis.

The dispersion presents a formidable challenge in this technique [25] and therefore, for simplicity, the frequency-dependence of all breast tissues or the frequency-dependent radiation patterns of the antennas are usually ignored [7], [26].

Strong scattering from the breast skin is compensated by aligning signals with respect to the skin reflection, and subsequently subtracting the averaged calibration signal from each measured signal. In order to enhance the detection process, various signal detection enhancement techniques such as confocal microwave imaging, space-time beamforming or time-reverse wave focusing can be employed.

Since measurement accuracy and a high dynamic range are key requirements for achieving a good image quality, all UWB radar imaging methods reported so far rely on frequency domain measurements using continuous wave systems, such as the one presented in Fig. 3. The time-domain representation is then achieved using an inverse Fourier transform [27]-[34]. The noise floor of the measurement system directly depends on its bandwidth, as followed from (2), and the maximum radiated power is also limited due to electromagnetic radiation risks. For that reason, the practical application of the pulsed time domain measurement systems for breast cancer detection is limited.

C. Combined Systems

The combination of the two previously described frequency and time domain techniques potentially allows to reconstruct high resolution images displaying the location, size, and complex permittivity of tumors inside the body at a relatively low computational cost.

This method has emerged with the purpose of overcoming the specific disadvantages of both included techniques. The UWB radar techniques are implemented in order to determine the scatterers spatial location and dimensions. Then, using this *a priori* information, frequency domain techniques are used to reconstruct the dielectric properties of the target. This can significantly reduce a computational time required for solution of the inverse scattering problem.

The technique was proposed recently without reference to the measurement procedure [35]. Taking into account the previous considerations one can conclude, that this combined technique will also be based on frequency domain measurements.

IV. Microwave Microscopy

The principle of the microwave microscope is the change in the resonant frequency of an open-ended microwave cavity resonator, which results from the interaction of the electromagnetic field of the resonator and objects positioned under the skin, such as breast tumors (Fig. 6).

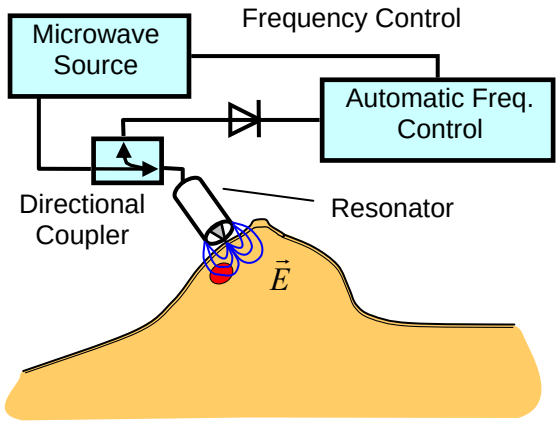

Figure 6. Microwave microscopy. The resonant frequency of the resonator changes due to tumor presence.

This can be illustrated using the relation for the fractional change in resonant frequency of a cavity perturbed by a change in the permittivity $\Delta\varepsilon$:

$$\frac{\omega-\omega_0}{\omega_0}=\frac{-\int_{V_0}\Delta\varepsilon\left|\vec{E}_0\right|^2 d\upsilon}{\int_{V_0}\left(\varepsilon\left|\vec{E}_0\right|^2+\mu\left|\vec{H}_0\right|^2\right)d\upsilon}, \tag{4}$$

where ω_0 and ω are the resonant frequencies of the original and perturbed cavity, respectively; $\vec{E}_0, \vec{H}_0$ are the fields of the original cavity with volume V_0 [36].

Equation (4) shows that any increase in ε at any point in the cavity will decrease the resonant frequency.

This technique provides a high spatial resolution since it is based on the near-field wave-tissue interaction, which is not limited by the diffraction limit. Near-field microwave microscopy has been successfully used for surface characterization of biological tissues with reported spatial resolution in the range of λ/50 to λ/1000. Recently, this

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technique was proposed for the breast cancer diagnosis [25], [37]. The maximum reported effective detection depth is few centimeters. For deeply situated tumors the system unfortunately looses its subwavelength focusing abilities.

Microwave microscopy for breast cancer detection offers a number of important advantages: it avoids special processing of skin tissues and implementation of complex inverse scattering algorithms; it operates in a very narrow frequency range, thus there is no need of complex dispersive dielectric models of the breast tissues; *a priori* knowledge of the dielectric properties of the tumor is not critical to the success of the detection; the method can be used for detecting breast cancer in men.

V. Hybrid Methods

A. Microwave-Induced Thermal Acoustic Imaging

Microwave-induced thermal acoustic imaging combines the advantages of high contrast in the conductivity of malignant tissues at microwave frequencies and the high spatial resolution of ultrasound imaging.

A microwave pulse generator is used here to irradiate the breast (refer to Fig. 7). The microwave power P_V absorbed per unit volume of tissue is proportional to its electric conductivity σ:

$$P_V = \frac{1}{2}\sigma\left|\vec{E}\right|^2 \text{ W/m}^3, \tag{5}$$

where $\vec{E}$ is the amplitude of the electric field intensity inside the tissue [38]. This absorption stimulates thermoelastic expansion of tissues and induces thermoacoustic waves, which can be detected by an acoustic sensor array positioned outside the breast. The measured excess pressure is a function of the microwave pulse width and the fractional energy absorption per unit volume of tissue at certain position. The acoustic waves generated in this manner carry the information about the microwave energy absorption properties of the tissues under irradiation. According to (5), the microwave energy absorbed by tumor and normal breast tissues will be significantly different and a stronger acoustic wave will be produced by the tumor.

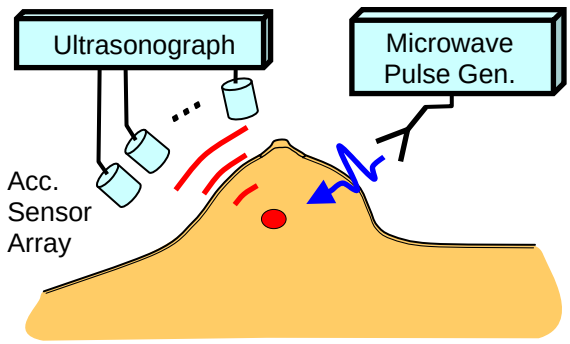

Figure 7. In microwave-induced thermal acoustic imaging, the acoustic waves are emitted by tumor heated with electromagnetic pulse

The main concern about the clinical implementation of this technique is the increased level of electromagnetic radiation. The microwave power of the pulse required for the modality is from one to several tens of kilo-watts [38]-[40], which even in average is much higher than used by other microwave imaging systems discussed above.

The majority of challenges facing microwave-induced thermal acoustic imaging system are referred to inhomogeneity of the breast tissue. This leads to a nonuniform microwave energy distribution, strong interference from the skin and chest wall, and consequently, to complicated image reconstruction algorithms. The biological tissues should be heated by the microwave source in a uniform manner, otherwise thermal acoustic signals will be induced by a nonuniform microwave energy distribution, resulting in images difficult to interpret. The excitation of undesirable high-order electromagnetic field modes in the breast tissue also contributes to a non-uniform microwave energy distribution. Because the breast skin, breast tissues, chest wall, and tumor absorb the microwave energy and convert the energy to heat according to (5), all of them produce thermal acoustic signals. The measured thermal acoustic waveforms include responses from the tumor, as well as from other healthy breast tissues. The thermal acoustic signals generated by the skin are much stronger than those by a small tumor because of the high conductivity of the skin and the acoustic sensors being very close to the skin. Because of non-uniform sound speed in biological tissues, the arrival time of the acoustic pulse generated at a location cannot be determined accurately. All these factors make it difficult to approximate the back propagation properties of thermal acoustic signals inside the breast.

The skin response is usually compensated by averaging in a similar way to the radar technique. The clutter can also be reduced using dispersive properties of the tissue and multifrequency operation, like is the case in radiometry, for example. The information collected from the multifrequency stimulation can help to mitigate the challenges mentioned above.

B. Ultrasound-Guided Microwave Imaging

An attractive potential offers multimodality imaging combining information collected from different systems to provide a more complete diagnostic tool that covers the full range of physiological and pathological conditions of tissues. Ultrasound-guided microwave imaging is such a combination of two modalities, where microwave image reconstruction is guided by ultrasonography. The ultrasound imaging is used to collect *a priory* information about the breast structure and geometry of embedded objects. This helps to generate an optimal mesh with well refined target region for effective numerical analysis of the electromagnetic problem [41]. Consequently, the spatial resolution of microwave imaging can be enhanced resulting in more accurate imaging of tumors.

VI. Conclusion

Microwave imaging has a considerable potential for breast cancer detection due to the high contrast in tissue dielectric properties at microwave frequencies. A number of promising strategies for microwave imaging have appeared recently trying to exploit this potential, among them thermography, microwave tomography, microwave microscopy, microwave-induced thermal acoustic imaging, and ultrasound-guided microwave imaging. Each method has weaknesses and strengths, but the amount of research

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and development exploring every possible approach continues to grow.

Summarizing a wide variety of proposed techniques, it should be noted that certain systems, such as passive systems, frequency-domain 2D tomographic systems, and recently, ultrawideband network-analyzer based systems, have already reached clinical trials.

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