Data Acquisition System for Harmonic Motion Microwave Doppler Imaging

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***Abstract*— Harmonic Motion Microwave Doppler Imaging (HMMDI) is a hybrid method proposed for breast tumor detection, which images the coupled dielectric and elastic properties of the tissue. In this paper, the performance of a data acquisition system for HMMDI method is evaluated on breast phantom materials. A breast fat phantom including fibro-glandular and tumor phantom regions is produced. The phantom is excited using a focused ultrasound probe and a microwave transmitter. The received microwave signal level is measured on three different points inside the phantom (fat, fibro-glandular, and tumor regions). The experimental results using the designed homodyne receiver proved the effectiveness of the proposed setup. In tumor phantom region, the signal level decreased about 3 dB compared to the signal level obtained from the fibro-glandular phantom area, whereas this signal was about 4 dB higher than the received signal from the fat phantom.**

I. INTRODUCTION

Breast cancer is one of the most common types of cancer among women [1]. Due to this fact, providing methods to assess the early-stage diagnosis and treatment of the breast cancer is of a great interest for researchers in the last decades. Mammography is being utilized as the primary imaging technique for breast cancer detection. However, this technique has been reported to suffer from missed detection, false alarms, employment of ionizing radiation, and patient discomfort [2]. Microwave imaging, has been proposed as a safe and efficient tool for breast tumor detection [3]-[6]. However, several studies show the limitation of microwave imaging in distinguishing the fibro-glandular and the malign tissues due to their similar dielectric properties [7]. Some other techniques such as contrast agents or other imaging modalities can support this imaging method [8], [9]. Recently, the Harmonic Motion Microwave Doppler Imaging (HMMDI) method was proposed as a hybrid method to detect breast tumor [10]-[12], [16]. In this method, microwave signal is transmitted to the tissue meanwhile the tissue is vibrated locally using focused ultrasound. The signal level at the Doppler (vibration) frequency on the received microwave signal due to the harmonic motion of the excited area is a feature to distinguish the tumor. This feature depends on both elastic and electrical properties of the tissue. The performance of the HMMDI method is investigated with simulations in [10]-[12]. Also, phantom experiments are conducted using a spectrum analyzer as a receiver [12], [16]. In this work, we investigate the received HMMDI signal in time-domain using a homodyne receiver, on breast phantom materials.

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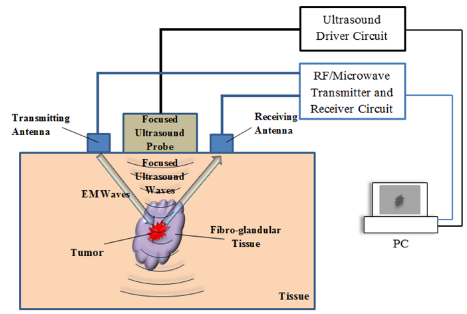

Fig. 1. The HMMDI Method setup. Ultrasound transducer generates local harmonic motion inside the tissue. A microwave transceiver system is used for detecting the Doppler signal component of the ultrasonically vibrated region [10].

II. METHOD

The HMMDI method employs a focused ultrasound transducer driven with an amplitude modulated signal to generate local harmonic motion inside the tissue (Fig. 1).

The short term time average volumetric force applied to the tissue due to amplitude modulated acoustic radiation force can be expressed as [10]:

$$F = \frac{\alpha P_0^2}{\rho c_s^2} cos^2(\Delta\omega t) \tag{1}$$

where α (1/m) is the absorption constant of the tissue, c_s(m/s) is the speed of ultrasound in tissue, P_0 (Pa) is the amplitude of the pressure wave, $\Delta\omega$ (rad/s) is the modulation frequency, and ρ (kg/m^3) is the density of tissue, and F is force per unit volume (kg/s^{2}cm^2).

The displacement of the local tissue depends on the applied force and the elastic properties of the local tissue. A continuous wave microwave signal is transmitted to the tissue during mechanical excitation. Backscattered signal is phase modulated due to the vibration of the focal region inside the tissue. The first harmonic of this modulation is sensed by the receiver. The amplitude of this component depends on both the elastic and electrical properties of the focal region [10]:

$$S_{Rx\_Doppler}(t) = \pm B\frac{K}{2}\cos((\omega_m + \Delta\omega)t + \frac{4\pi R}{\lambda} + \phi) \qquad (2)$$

where R is the distance from antennas to locally vibrating tumor, ω_m is the operating frequency of the radar, $\Delta\omega$ is the vibration frequency of the tissue, λ is the wavelength of the microwave signal, and ϕ is a constant phase depending on the total path length. K is the change in phase (in radians) of the signal, which depends on the elastic properties of the focal region. B is the magnitude of the received signal for the case without vibration, which depends on the electrical properties of the focal region and the background tissue.

A. Phantom Development

Oil in gelatin emulsion are suitable for elastography and microwave phantoms [13], [14]. The main ingredients of them are gelatin, safflower oil, deionized water, p-toluic acid, n-propanol, formaldehyde, surfactant, and kerosene. Electrical and elastic properties of phantoms can be controlled by the amount of oil. In addition, it was reported that the elastic constant of the phantom material can be decreased by baking it at 50 °C [15].

Three different types of phantoms (normal breast fat, normal fibro-glandular, tumor) are produced in this study. The amounts of materials for preparing about 1 liter phantom volume are given in TABLE I. The fibro-glandular tissue phantom is obtained by baking tumor phantom at 50°C for 4 days. The phantom is prepared inside a glass bowl with 11 cm diameter and 5 cm height. The phantom development consists of two stages. Firstly, the fat phantom is prepared, and poured into the container up to a height of 2.5 cm from the bottom of the mold. The phantom is leaved to solidify for one day. The fibro-glandular tissue phantom is prepared inside a falcon tube of 25 mm diameter. After the baking period, 16 mm height portion of it is cut and placed on this fat phantom. The tumor phantom of 13 mm diameter and 7 mm height is inserted on the fat phantom 30 mm away from the middle of the container. Another fat phantom is prepared and poured in the same mold until the surface level reached a 3 cm in height. The phantom is left for 7 days at room temperature for solidification. Different stages of phantom development in addition to its MRI image are shown in Fig. 2. The dielectric, elastic and ultrasonic properties of the developed phantoms were measured to verify their characteristics as tissue mimicking phantoms to be used in HMMDI method assessment [16], [17].

TABLE 1

Amount of materials for preparing 1 liter realistic phantom

Tissue Type Ingredient	Fat	Tumor/Glandular
Kerosene (ml)	333	49.5
Oil (ml)	333	49.5
P-Toluic Acid (mgr)	167	800
N-Propanol (ml)	8.3	39.6
Water (ml)	158	753.5
Gelatin (gr)	28.6	135
Formaldehyde (ml)	1.67	7.9
Surfactant (gr)	140	10.1

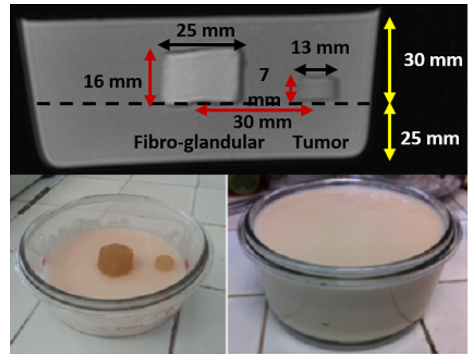

Fig. 2. Homogeneous breast phantom including fibro-glandular and tumor phantoms. Top: MRI image of Phantom, bottom-left: Fibro-glandular and Tumor phantoms placed on the fat phantom at the first stage. Bottom-right: Final condition of the breast phantom after the second stage

III. Results

The performance of the proposed data acquisition system in HMMDI method is evaluated experimentally using the developed phantom materials. The block diagram of the system setup is shown in Fig. 3.

In this study, a single element focused ultrasound (FUS) transducer H-102 (Sonic Concepts, WA, USA) is used for generating vibrations inside the tissue. It has 64 mm diameter and 63.2 mm geometric focus. It is used in the third harmonic frequency (3.32 MHz). The axial and lateral full width at half maximum intensity beam widths are measured as 5 mm and 0.6 mm, respectively. A 2-cycle 10 Hz sinusoidal burst signal (with 1 Hz pulse repetition frequency) is generated from waveform generator 1 and used to amplitude modulate the 3.32 MHz signal generated by the waveform generator 2. The AM signal is amplified with a high power RF amplifier (150A100B, Amplifier Research, WA, USA) of 52 dB gain. Spatial peak pulse average intensity (I_{sppa}) of the ultrasound beam is 271.4 W/cm^2. The reason of high intensity requirement (above safety limits) is that the Youngs Modulus of the fibro-glandular phantom was about 5 times higher than the reference value and the dielectric constant of the fat phantom was about 2 times of its reference value [17]. Simulation studies show that for real tissues lower intensity values inside safety limits would be sufficient [12].

Open-ended waveguide antennas are used for transmitting and receiving microwave signals. The antennas are placed in H-plane configuration and 15 mm away from each other. They are sticked to the bottom of the phantom container with water-proof silicon adhesive glue in order not to block the ultrasound transducer at the upper side. Glass is used as a mold for the breast phantoms. The dielectric material used in the antenna is vegetable oil ($\varepsilon_r = 2.5$, $\sigma = 0.05$ S/m [12]). Oil is prepared and poured into the antenna and the opening of the waveguide antenna is closed using a plastic material. The plastic is fixed to the aluminum using a waterproof adhesive.

The focus of the FUS transducer is adjusted manually to point at 30 mm depth from the top surface of the phantom mold. The transmitting microwave antenna is fed by the Agilent E8257C Signal Generator with an output power of

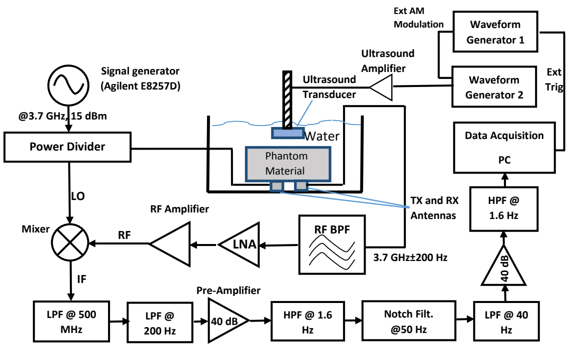

Fig. 3. Block diagram of data acquisition setup

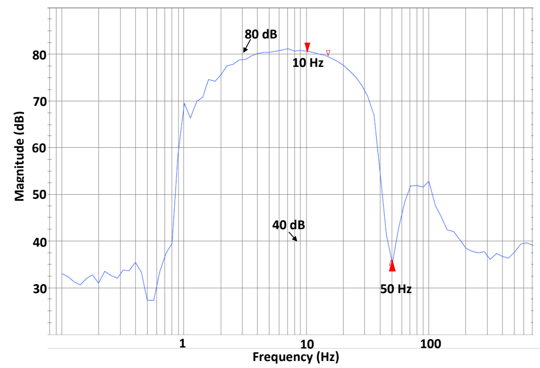

Fig. 4. The designed baseband-amplifier frequency response

+15 dBm at 3.7 GHz frequency. For data acquisition in time domain, the signal is down-converted to baseband. For this purpose, the receiver antenna is connected to a band pass filter (K&L Microwave 5C50-3700/U100-O/O), low noise amplifier (Mini Circuit ZX60-3800LN-S+), and wide-band amplifier (Mini Circuit ZX60-V83-S+), respectively (Fig. 3). The filtered and amplified signal is fed to the mixer (MITEQ DM0204 LA1). The signal from signal generator is split using power divider (Mini Circuit ZFRSC-42-S+) and fed as the local oscillator to the mixer other input. The output from the mixer is filtered using a low-pass filter (Mini Circuit NLP-50). In addition to the proposed homodyne receiver system, further amplification and filtering sections are necessary due to the received low level noisy signal. Since we are interested in the signal at the Doppler shift frequency around 10 Hz, the signal is filtered by the low-pass filter to remove the unwanted high frequencies. The output of this filter is connected to the commercial instrumentation amplifier (LT1167). The amplified signal is passed through a high-pass filter to eliminate DC offset. Furthermore, the notch filter is considered to eliminate the 50 Hz line noise. The signal is filtered adding another low-pass filter with 40 Hz cut-off frequency. Additionally, the output is amplified by a gain of 40 dB. The frequency response of the presented circuitry is given in Fig. 4. The received data from the RF receiver section which is amplified by total 80 dB gain, is collected using data acquisition card. In this experiment, CM series 2004 model data acquisition card of ACQUITEK is used, which supports up to 1 MS/s maximum sampling rate, and 16 bit A/D resolution. The signal at each point is collected for 4 second with 15 kHz sampling frequency. In this duration of receiving signal, there are at least three trigger pulses which contain our desired signal due to the vibration caused by the ultrasound waves. The part of the signal which occurs during the existence of the ultrasound pulse contains information about coupled elastic and dielectric properties of the focal region. The acquired data is further processed by band-pass IIR Butterworth filter ($f_{c1} = 1$ Hz, $f_{c2} = 25$ Hz) to obtain the signal in the interested frequency range. The received data after digital filtering is illustrated in Fig. 5 for cases of with and without FUS transducer excitation in fat, fibro-glandular, and tumor regions. The trigger signal from the waveform generator is shown in red dashed line. There is an unwanted noise even when there is no ultrasound excitation. However, the received signal due to the excitation is clearly observable in all cases. As it is very close to the glass mold sidewall, the signal from the tumor part is distorted by the reflected signal from the glass. Therefore, just the first peak of the received signal from tumor is observed to be in the trigger pulse duration. A bigger phantom mold should be used for future studies to prevent this problem. The first peak values (ie. the received signal amplitude for the first excitation cycle) of the three received pulses are averaged to get the HMMDI data. The maximum signal is achieved when the focus is on the fibro-glandular phantom. The signal amplitude decreases about 3 dB in the tumor phantom. In the fat phantom, the signal level further decreases by an amount of 4 dB. These results are in a good agreement with the results obtained by directly connecting the receiver antenna to a spectrum analyzer [16], [17]. Even though the results are from specific points of the phantom, they prove the efficiency of the designed data acquisition system. It is shown that a low level displacement signal at Doppler frequency can be acquired for further processing steps. To detect the tumor inside the fibro-glandular in the fat phantom and to distinguish them from each other the phantom must be fully scanned.

IV. Conclusion

In this paper, the data acquisition setup was proposed for HMMDI method and it was assessed using the developed phantom materials. The phantom materials were vibrated locally using a FUS probe. Microwave signals were transmitted to the phantom during ultrasound excitation. The amplitude of the received signal at the Doppler (vibration) frequency was down-converted to the base-band using the designed base-band circuit. Furthermore, the signal was amplified and filtered to receive the desired signal level for further processing. The results show that the data from different tissue phantoms are detectable using the proposed receiver setup and they have a potential for detecting tumor inside fat and fibro-glandular tissues. In future studies, the method should be evaluated for detecting the tumor inside the fibro-glandular phantom which have both high dielectric constants.

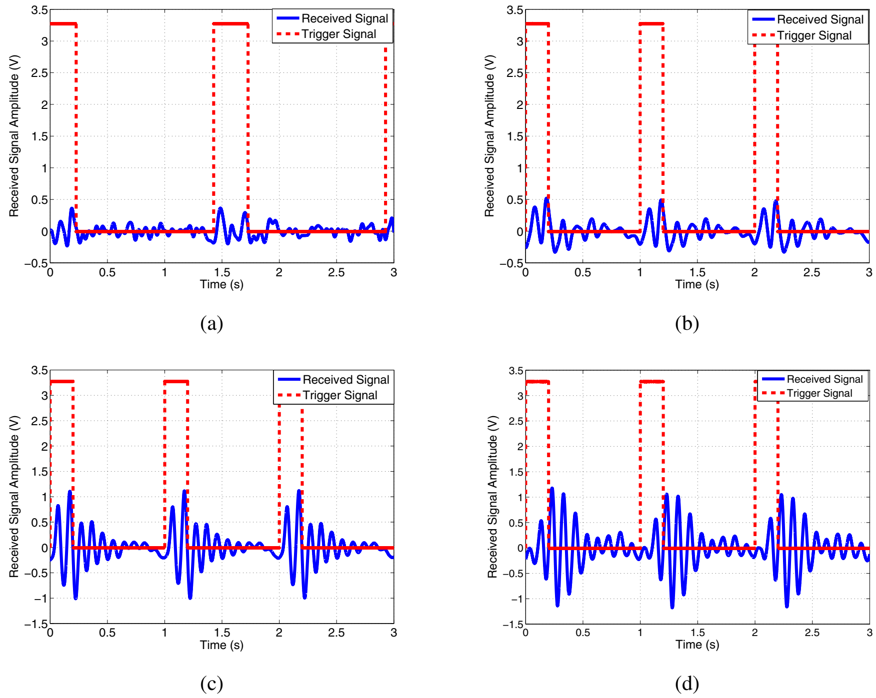

Fig. 5. The received signal (a) without FUS transducer excitation, (b) with FUS transducer excitation from the fat region (c) from the fibro-glandular region, and (d) from the tumor area. The trigger signal from the waveform generator is shown in red dashed line.

In addition, better signal processing methods should be implemented for de-noising and removing the artifacts in the detected signals. Two dimensional scan of phantom could give better data to image the dielectric-elastic properties of tissue phantoms.

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