Ultrasound Imaging System: A Review

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Abstract - Ultrasound imaging is an efficient, noninvasive, method for medical diagnosis. It is now a mature technique to the extent that it has a well established place in clinical practice. This doesn’t mean, however that the innovation in techniques and instrumentation has slowed down. Indeed opposite is true. In order to provide the patient oriented medical services, in and out of hospitals, an engineering community is doing research to transform a conventional standalone system into a low power, low cost, real time portable version. This review takes a look at some of the attempts which have been published as well as some of the issues which are still yet to be resolved.

I. INTRODUCTION

Ultrasound is defined as acoustic wave with frequencies above those which can be detected by the human ear, from about 20 KHz to several hundred MHz (Human hearing is in the range 20-20KHz). Ultrasound for medical imaging applications typically uses only the portion of the ultrasound spectrum from 1 MHz to 15 MHz due to the combined needs of good resolution (small wave length) and good penetrating ability (not too high a frequency)

They are generated by converting a radio frequency (RF) electrical signal into mechanical vibration via a transducer. The ultrasound waves propagate into the tissues of the body where a portion is reflected, which is used to generate the ultrasound image. Employed ultrasound waves allow obtaining information about the structure and nature of tissues and organs of the body. It is also used to visualize the heart, and measure the blood flows in arteries and veins [1].

Ultrasound imaging is valuable for several reasons. For one, it’s inexpensive—at least compared with CT (computed tomography) and PET (positron-emission tomography) scanning, or with MRI (magnetic resonance imaging). Also, the low-amplitude ultrasound waves used for imaging do not involve ionizing radiation and are thus harmless to the patient, so repeated scans can be made without worry. And with this technique it is not difficult to get real-time images [2].

Science its beginning in the 1950’s medical ultrasound has become to grow and mature. As standalone machines become increasingly complex, a new segment of the engineering community is doing just the opposite, trying to make medical ultrasound machines smaller, more power - efficient and less costly than they have ever been [3] [4].

These portable or handheld systems can become ultrasonic stethoscopes that allow physicians to perform ultrasound examination almost anywhere and at any time. For instance, a hand-held system can provide point-of-care diagnosis in remote locations, battlefields, emergency rooms, and private clinics. It can also be used in trauma or minimally invasive ultrasound-guided procedures such as central catheter insertion. With wide-ranging applications in clinics, developing countries, and the military, the demand for portable ultrasound systems has increased rapidly in the last decade. The P10 from Siemens (Berlin, Germany) and the VSCAN from GE Healthcare (Waukesha, WI) are examples of recently introduced pocket-sized ultrasound systems [5].

However, the most important supporting technologies required to design ultrasound imaging systems include transducers, transmit/receive circuitry design and beamforming. This paper reviews some of the attempts which have been published to address all the parameters described above. And figure 1 will illustrate the improvement in the quality and clinical usefulness of ultrasound image which has occurred over the last 25 years [6].
pads for each element. A gold plated polyester sheet covered all 1024 transducers to complete the connection. Due to the PCB traces that crossed over each other, crosstalk was a large portion of the overall signal. However this design was sufficient to generate a proof of concept [3][7].

Eames et al. continued the work by Girard et al. at the University of Virginia with the creation of a 60 x 60 (3600 element) transducer array. Eames et al. looked to improve upon the problems Girard et al. faced with crosstalk in their device by creating 3600 straight through holes. Eames et al. design resulted in a slightly lower resonance frequency of the piezoceramics than was anticipated, probably due to the element thickness. Further problems such as grating lobes were introduced with the low aspect ratio which was caused by the dimensions of the transducer [3][9].

Transducer design has continued to progress and the type of transducer greatly depends upon the application of the ultrasound system. The frequency of ultrasound probes vary roughly between 1 to 10 MHz depending on the application. Most of the systems have the ability to use transducers that range from 1 to 10 MHz[3].

Nowadays the transducers which are in clinical use almost exclusively use a piezoelectric materials, of which the artificial ferroelectric ceramic, lead zirconate titanate (PZT), is the most common[2]. However, as progress in MEMS – fabrication continues capacitive ultrasonic transducers are beginning to compete with piezoelectrics. These capacitive micromachined ultrasonic transducers (CMUTs) hold the promise of dramatically reducing the cost associated with ultrasonic transducers along with providing revolutionary advances in current technology [3].

Capacitive micromachined ultrasonic transducers (CMUTs) invented in the mid-1990s [10] have come long way in the last two decades and recently reached the market for medical ultrasound imaging (Hitachi-Mappie). Considering the production of conventional ultrasonic transducer probes alone, which amounts to a global market of about $1 billion annually, one can guess that CMUTs can be the next big MEMS product in the medical field [11].

Oralkan et al. was the first to present a pulse-echo phased array B - scan sector image using 128 CMUT elements in a 1D transducer array. They also showed the evidence that CMUTs can compete with piezoelectrics in terms of efficiency and bandwidth [11][12].

An overview of the CMUT technology is shown in figure 2. A direct current voltage is applied between the membrane and the substrate which pulls the two
together by electrostatic forces. The pulsing of which will generate an ultrasonic signal.

![Diagram of a CMUT](image)

**Fig. 2**: Diagram of a CMUT (from Oralkan et al.) [12].

However, this new technology is not without some current drawbacks. Large electric field is required to drive these transducers and the process can present some problem. Due to these large electric fields insulating layers can break down and dc bias fields must be constantly modified to compensate. Also since these transducers respond in a nonlinear fashion, it becomes difficult to measure nonlinearity of living tissue [3] [2].

For low-cost applications CMUT continue to offer promise for the field. Even though low cost design have been implemented with piezoelectric transducer, they generally provide much of the cost associated with ultrasound systems, still cost >$1,000. Therefore in order to create a truly low-cost solution, new technology such as CMUTs has to be developed to the point of commercialization [3].

**III. HARDWARE IMPLEMENTATION**

Many of the ultrasound imaging systems are pulse-echo systems. Where a piezoelectric or micromachined transducer is used to generate the ultrasound pulse, as well as to receive the reflected echo. In conventional ultrasound systems, the transducer array elements are located in handheld probe, which is connected to the main processing unit via a cable bundle. Transmit pulsers and receive amplifiers are located in the main processing unit [13].

If all the electronics needed for pulse generation and reception could be integrated in the scanner or probe head, all caballing could be omitted. One important step to reach this target is to miniaturize driver and receiver electronics in order to comply with portable requirements. Furthermore, multiplexing and beamforming circuitry (to be discussed later) can also be integrated with the transducer array to minimize the number of active electronic channels and number of physical connections between the probe and backend system [11]. Thus, the flexibility in medical environment would be greatly enhanced.

**A. Transmit Circuitry Design**

Another major difficulty in the development of portable ultrasound imaging system is the design of power system which effectively use low voltage sources. Traditional ultrasound systems rely on high voltage and current to drive the piezoelectric transducers.

Owen et al [3] [14] has developed a 12lb plug in class D switch mode amplifier to drive single element high intensity focused ultrasound transducers. The system provided 140W of acoustic energy to a 70% efficient PZT transducer. Owen et al concluded their device was comparable to available commercial applications.

According to Lewis et al [3] [15] the majority of ultrasound drivers and RF amplifiers are generally built with an output impedance of 50ohms. In order to obtain the maximum power transfer matching circuitry must be used to transfer power to the transducer. However, in matching impedances which are generally complex, systems incur additional costs and complexity. Lewis et al worked to develop driving circuitry with an output impedance of 0.3 ohms which transferred power with 95% efficiency to the transducer.

**B. Receive Circuitry Design**

The receive circuitry designed for low-cost, portable ultrasound systems must be sensitive to receive the acoustic waveform and use low voltage source. Traditional ultrasounds rely on high voltages and current to drive sensitive circuitry to receive the acoustic waveform [3].

The sonic window, a low-cost portable ultrasound system developed by Fuller et al. [3][4], has been looking to accomplish this. The project takes many of the suggestions from the transducer design, electronics, and beamforming (to be discussed later) into account in the design of this low-cost, portable system.

![Block Diagram of the Sonic Window](image)

**Fig. 3.** Block Diagram of the Sonic Window by Fuller et al [4]
The system was implemented with a 2D array (32 x 32) very similar to the array introduced by Girard et al. Each receive channel consists of on-chip transmit protection shunting device, a variable gain preamplifier, a band pass filter, a sample and hold circuit, and an analog-to-digital circuit with memory. By placing the transmit protection devices on the chips eliminates the need for bulky expensive power consuming switching elements.

However, the system was not without problems that included routing between the individual elements and their respective receive channels, which resulted in a 6.74% channel loss. This could have occurred due to the long PCB traces that induced parasitic inductances, capacitances, and resistances [3].

The result from this prototype support the design of low cost, pocket seized, c scan ultrasound device [16].

Jonny Johansson et al [13] has designed a complete autonomous transmit and receive circuitry for a piezoelectric transducer in one single ASIC. The chip intended to be operated from a single lithium battery with supply voltage of 3.6v. Such a low supply voltage limits the amount of ultrasound energy that can be generated. To overcome this, the piezoelectric transducer is charged with an on-chip boost converter, which uses one external inductor to generate up to 40V on the crystal. The design has three main building blocks: a charge/discharge unit, an amplifier and a state machine to control the functionality.

For integrating CMUT arrays with electronic circuits several methods developed for a variety of MEMS application have been adapted [11]. The complete front end has been reported by Ira O Wygant et al [17]. A 2D, 16 x 16 CMUT array is flip-chip bonded to a custom-designed integrated circuit (IC) that comprises the frontend circuitry for the transducer elements. The array and IC are connected to an FPGA-based data acquisition system that can acquire volumetric imaging data in real time.

Fig. 4 : Block schematic of the complete chip outlining the main blocks and external connections. [13]

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Fig. 5 : Catheter-based ultrasound imaging probe with integrated electronics [17].

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Fig. 6 : Photograph of a CMUT array flip-chip bonded to the frontend IC. A cross section of the device is shown below the picture [17].

IV. BEAMFORMING ALGORITHM

Beamforming algorithms are used to steer and focus acoustic signals. During both transmit and receive operation, appropriate delay will be supplied to accomplish the focusing and steering needed [3] [18].

Currently, a significant percentage of the size and power of an ultrasound system is devoted to the beamformer, which is responsible for focusing the ultrasound beam. The standard beamformer, which consists of 64 to 128 transmit/receive channels, is straightforward to implement if design constraints such as size and power are relaxed. As ultrasound systems become more portable, however, the new beamformer architectures with fewer channels and lower power consumption than standard cart-based systems will be needed [3] [5].

Initially in order to form the 2D B-mode image, mechanical transducers were used to mechanically rotate the transducer. Analog delay lines and fast analog processing units allowed the advent of phased array systems in the 1980’s. However, after advances in silicon technology which lead to more sophisticated analog-digital circuits, beamforming algorithms were
programmed digitally. These digital signals soon replaced their mechanical counterparts [3] [19].

Traditionally three different methods were used to implement a time delay. 1) RF modulation onto an intermediate frequency. 2) up sampling the incoming signal using interpolation filter . 3) Nonuniform sampling of RF signal according to the needed time delay.[3] [20][21].

The oversampling method of implementing time delays soon became the popular method to implement time delays because of their relative simplicity and ease of integration. General Electrical patented a delta - sigma oversampling A/D, which suffers from a major flaw which reduces image quality significantly [3].

Freeman et al.[3][19] corrected this problem with the creation of the Delta - Sigma Oversampled ultrasound beamformer. This method, now serves as one of the best low-cost beamforming options available.

Ranganathan et al.[3][22] determined the simplest beamforming algorithm which yielded image quality, the developing the direct sampled I/Q (DSIQ) algorithm.

Man Minh et al [5] proposed a modified electronic Fresnel-based beamforming method for low-cost portable ultrasound systems. This method uses a unique combination of analog and digital beamforming methods. The advantage of this method is that a system with 4 to 8 transmit channels and 2 receive channels with a network of switches can be used to focus an array with 64 to 128 elements.

Fig. 7: A 4-transmit Fresnel (phase and sum) beamforming schematic for a 2-element array.

V. CONCLUSION

Looking back over the entire history of medical ultrasound imaging it is obvious that, medical ultrasound imaging technology is continually evolving and advancing all with the goal of improving patient care. Recent advancement in electronics, MEMS fabrication technology, and digital signal processing techniques have made medical ultrasound imaging to stand out from other imaging methods in providing real-time diagnostic capability at an affordable price while being physically portable.

REFERENCES


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