Ultrasonic transducers for medical diagnostic imaging

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Abstract Over the past decades, ultrasound imaging technology has made tremendous progress in obtaining important diagnostic information from patients in a rapid, noninvasive manner. Although the technology has benefited from sophisticated signal processing technology and imaging system integration, much of this progress has been derived from the development of ultrasonic transducers that are in direct contact with patients. An overview of medical ultrasonic imaging transducers is presented in this review that describes their structure, types, and application fields. The structural components of a typical transducer are presented in detail including an active layer, acoustic matching layers, a backing block, an acoustic lens, and kerfs. The types of transducers are classified according to the dimensions of ultrasound images: one-dimensional array, mechanical wobbling, and two-dimensional array transducers. Advantages of each transducer over the other and the technical issues for further performance enhancement are described. Application of the transducers to various clinical imaging fields is also reviewed.

Keywords Ultrasonic transducer · Ultrasound image · 1D array · Mechanical wobbling transducer · 2D array

1 Introduction

Medical ultrasound imaging for diagnosis has advantages, such as reasonable cost, real-time imaging, portability, and its harmless effect, over computerized tomography (CT) and magnetic resonance imaging (MRI) [1]. However, the resolution of the ultrasound imaging system is usually lower than that of CT and MRI systems. The ultrasonic imaging system consists of ultrasonic transducers and an imaging system. The imaging system controls the ultrasonic transducer in order to transmit and receive the ultrasound, and creates an ultrasound image with a set of data from the transducer. Depending on the type of the transducer and the imaging system, the images may be either two-dimensional (2D) or three-dimensional (3D). Ultrasound imaging technology has benefited from increasingly sophisticated computer technology, and system integration has ensured better image quality, data acquisition, analysis, and display. However, much of this progress has been derived from the development of transducers that are in direct contact with patients, which has expanded the possibilities for maximizing patient diagnostic information. This paper reviews the structure, type, and role of the transducers in realizing high-quality ultrasonic images.

There are different types of transducers used in various fields such as cardiology, obstetrics, gynecology, urology, orthopedics, and ophthalmology, as illustrated in Fig. 1. The position, size, and properties of objects being observed determine the shape, size, type, and frequency of the transducer required to achieve the field of view appropriate for a specific application [2–6]. The transducers are broadly classified into a one-dimensional (1D) array transducer, mechanical wobbling transducer, and 2D array transducer. The 1D array transducer, comprising several tens or

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Fig. 1 Photograph of ultrasonic transducers [5]

hundreds of active elements in a linear mode, generates a 2D planar image when all the elements are operated simultaneously or in sequence. The mechanical wobbling transducer is composed of a 1D array and a mechanism that can control the precise position of the 1D array to form a 3D image by combining several 2D images created with the 1D array. The 2D array transducer produces a pyramidal beam pattern to acquire a volumetric image instantly. This paper reviews detailed operation principle, structure, and application of these transducers.

2 Structure of the transducer

The parameters of the transducer performance, which influence the quality of ultrasound images, are the axial and lateral resolution and sensitivity [7]. The axial resolution is determined mostly by the frequency of the ultrasound wave. As the frequency increases, the wavelength decreases, which is advantageous because it provides a better distinction between a target and other objects. The lateral resolution along the direction orthogonal to the axial direction is determined by the beam profile of the transducer. A narrower beam leads to better resolution along the lateral direction. The sensitivity of the transducer determines the contrast ratio of the ultrasonic images. A transducer with higher sensitivity can generate a brighter image of the target. The transducer is designed to acquire high-quality images by enhancing these performance parameters.

A typical 1D array transducer is composed of an active layer, acoustic matching layers, a backing block, an acoustic lens, kerfs, a ground sheet (GRS), and a signal flexible printed circuit board (FPCB), as illustrated in Fig. 2. The active layer is usually made of a piezoelectric material—mostly piezoceramic. The active layer generates an ultrasound wave in response to an electric driving signal, receives the wave reflected at the boundary of an

organ, and converts the received ultrasound wave to an electric signal by means of the piezoelectric effect. However, the big difference in the acoustic impedance between piezoceramic elements and a human body prevents the efficient transfer of ultrasonic energy between the two media. The acoustic matching layers are used to facilitate the transfer of ultrasound energy [8]. Each matching layer has a thickness of one-quarter wavelength at the center frequency of the transducer. The backing block is used to absorb the ultrasound wave propagating backward from the piezoelectric element. If the backward wave is reflected at the bottom of the backing block and returned to the piezoelectric element, it can cause noise in the ultrasound image. Thus, the backing block should have a high attenuation. In addition to this material damping, several structural variations have been implemented to increase the scattering effects inside the backing block, e.g., inserting grooves or rods in the block [9–11]. The backing block commonly has an acoustic impedance between 3 and 5 Mrayl [12]. If the backing block has an acoustic impedance that is too high, the acoustic energy generated by the piezoelectric element will be wasted by the backing block and few ultrasound waves will be transmitted to the human body. The acoustic lens protects the ultrasonic transducer from exterior damage, and focuses the ultrasound beam onto a specified point based on Snell's law [13]. Materials with low attenuation constants are preferred to reduce the loss of ultrasound energy inside the lens [14, 15]. Typical acoustic lenses are made of rubber materials for comfortable contact between the transducer and patients. The kerf is a gap between arrayed piezoelectric elements that isolates each element from its neighboring elements to reduce the crosstalk between them. The crosstalk seriously degrades the transducer performance. Therefore, various shapes and materials of the kerf have been developed to decrease the crosstalk [16, 17].

To develop high-performance ultrasonic transducers, many researches have been carried out to improve their structure and components. The most significant effort is the use of a good active layer. The most common piezoelectric materials used in commercial transducers are piezoceramic materials that are cheap, easily available, and well-characterized. However, since the efficiency of piezoceramics for transmitting ultrasound waves to a human body is low due to their high impedance, piezoelectric composite materials have been developed to decrease the impedance [18, 19]. The piezoelectric composite material consists of a piezoceramic arrayed in a certain fashion and a low impedance polymeric material filled in between the arrayed piezoceramic. This method also increases the electromechanical coupling coefficient, which is a measure of the conversion efficiency between acoustic and electrical energies [20]. Piezoelectric single crystals are another

Fig. 2 Schematic structure of a 1D array transducer

alternative for the active layer, which has a superior electromechanical coupling coefficient but a limited usable temperature range [21]. Additionally, a multi-layered piezoelectric structure has been developed for a better electrical impedance match with an imaging system [22]. It is fabricated by laminating piezoelectric sheets along their thickness. Another structure is related to a quarter wavelength resonant mode of a piezoelectric layer [23, 24]. The use of a rigid material for the backing block results in a node of deformation at the boundary between the piezoelectric layer and the backing block. Therefore, the piezoelectric layer is deformed toward the acoustic matching layer, and more acoustic energy can be efficiently transmitted to the body.

For the active layers, apart from the piezoelectric materials, a capacitive micromachined ultrasound transducer (CMUT) and a piezoelectric micromachined ultrasound transducer (PMUT) have been developed [25, 26]. A CMUT has a thin metalized membrane that is suspended by insulating posts over a conductive silicon substrate. When an alternating voltage is applied between the membrane and substrate, the membrane is moved by Coulomb forces against the surface tension of the membrane, which generates ultrasound waves. Conversely, detection currents are generated by the change in the capacitance when the biased membrane is moved by the reflected waves. The CMUT has a higher electromechanical coupling coefficient than the piezoelectric material, and it can be fabricated to a small size by using photolithography processes. The PMUT has a structure similar to that of the CMUT except that the PMUT has a piezoelectric layer deposited on top of the silicon membrane.

To acquire a bright ultrasound image, the acoustic energy propagating in the transducer and human body has to be increased by operating the ultrasonic transducers with a high voltage. However, the increased acoustic energy is converted to thermal energy due to various attenuation mechanisms, which induces a temperature rise in the

transducer. The high temperature of the transducer may cause patient's skin to burn and degrade the transducer performance. Therefore, thermally dispersive structures have been developed to mitigate the temperature rise [27, 28].

3 Types of transducers

3.1 Transducers for cross-sectional 2D images

As described in the Introduction, the 1D array transducer is used to obtain cross-sectional 2D images, as illustrated in Fig. 3. The 1D array is composed of piezoelectric elements arrayed one-dimensionally along the azimuthal direction. The 1D array transducer is classified into a linear array, a convex array, and a phased array in accordance with the image shapes [7]. Basically, the linear array drives a few of the piezoelectric elements to generate an ultrasound beam to scan a line as illustrated in Fig. 4a. The beam profile

Fig. 3 2D ultrasound image of a uterus using the 1D array transducer [5]

Fig. 4 Schematic of the 1D array transducer: **a** linear array, **b** convex array, and **c** phased array

along the azimuthal direction can be changed by controlling the number of operated elements. Thus, the ultrasound image of the linear array has a rectangular shape. Since the linear array is normally used for precise imaging, its operating frequency is high. In contrast, the convex array is used to acquire a wide and deep ultrasound image at the cost of the resolution. For this reason, the piezoelectric elements of the convex array are arranged in a curved fashion along the azimuthal direction as illustrated in Fig. 4b. The method of acquiring an image using a convex array is the same as that when using a linear array but the ultrasound image of the convex array has a fan shape. However, in the case of a target object behind obstacles, such as a heart behind ribs, it is difficult to obtain an ultrasound image using the linear array or convex array. For this case, a phased array can be used for imaging by steering the ultrasound beam, as illustrated in Fig. 4c. When all of the piezoelectric elements are controlled to operate sequentially, the phased array can steer the ultrasound beam. The ultrasound image of the phased array has a circular cone shape.

However, none of the 1D array transducers mentioned above can be used to control an elevation beam profile because the length of the piezoelectric elements is fixed. The image is likely to be blurry in the area other than the focal zone of the transducer because the ultrasound beam is scattered outside the focal zone. For this reason, 1.25D, 1.5D, and 1.75D array transducers have been developed to modify the ultrasound beam on the elevation and depth plane [29, 30]. These array transducers have a structure in which the piezoelectric elements are arrayed along the elevation direction in addition to the azimuthal direction to drive the piezoelectric elements in the elevation direction as well. Although these transducers have better ultrasound beam controllability along the elevation of the transducer, they are just upgraded versions of the 1D array. Therefore, they retain the limitations of 1D array transducers.

3.2 Transducers for 3D images

In order to acquire the volumetric image shown in Fig. 5, two different types of 3D ultrasound imaging transducers are used: (1) a mechanical wobbling transducer that generates a 3D image by combining multiple 2D images from the 1D array, and (2) a 2D array transducer that comprises several thousand piezoelectric elements arrayed on a plane to transmit an ultrasound beam in a pyramid shape [31].

First, the mechanical wobbling transducer acquires the 3D image data using a mechanical sequential scanning method, which means that the 3D information of the measured object is presented in multiple 2D images. Thus, the 1D array and the mechanism to control the movement of the 1D array compose the mechanical wobbling

Fig. 5 3D ultrasound image of a fetus using the 3D imaging transducer [5]

transducer as shown in Fig. 6. The mechanism controls the rotational behavior of the 1D array in a prescribed position by applying a dynamic force generated from a servo or stepping motor to the 1D array. A conical volume data set can be acquired as the 1D array rotates in a semicircle around the central axis while a pyramidal data set can be obtained as the 1D array moves in a fan-like arc according to a prescribed angle [32, 33]. The technical issue is the optimization of the scanning rate, scanning angle, long-term reliability, and compliance for better performance of the transducer.

The 2D array transducer can generate real-time 3D ultrasound images through volumetric steering of the ultrasound beam. Since the 2D array transducer consists of

Fig. 6 Schematic structure of a mechanical transducer

thousands of piezoelectric elements arrayed along both the azimuthal and elevation directions as illustrated in Fig. 7, the volumetric data set can be acquired instantly through electronic control of the piezoelectric elements both horizontally and vertically. The embodiment of the 2D array transducer is a technically challenging issue that is related to electrical wiring of all the piezoelectric elements and reducing the crosstalk among the dense elements. All the piezoelectric elements fabricated in the small footprint area are connected to a controlling electronic circuit with a multi-layered FPCB or conductive backing block [34, 35]. Since it is not easy to handle thousands of cable bundles to connect the 2D array with an imaging system, it is necessary to implement specific integrated circuit (ASIC) chips inside the transducer for pre-processing the image data [36, 37].

4 Application

Ultrasonic transducers of these basic structures can be modified in terms of their shape, size, type, and operating frequency for various imaging applications. In the field of cardiology, for instance, an ultrasound image of the heart behind ribs can be obtained with a transesophageal echocardiogram (TEE) transducer that includes a small 1D phased array or 2D array transducer [38]. The TEE transducer should be small so that it can be inserted in the patient's mouth and esophagus. The convex array and

Fig. 7 Schematic structure of a 2D array transducer

mechanical wobbling transducers are used to obtain wide and deep images of a fetus, uterus, and ovary through the abdomen in the field of obstetrics and gynecology [39]. Breast is usually imaged with 1D linear array transducers from skin surfaces. In the field of urology and endocrine system, the linear array transducer is used to obtain an ultrasound image of a prostate, bladder, testis, and thyroid from the skin surface. Additionally, an endo-vaginal or endo-anal transducer having a thin rod shape is also used to obtain the images of uterus and prostate through the vagina or anus [40]. In the vascular system, an artery image can be acquired with an intravascular ultrasound (IVUS) transducer using a miniaturized ultrasonic transducer built-in catheter [41]. Additionally, a 1D linear array transducer operating at high frequencies is used to acquire high-resolution images of a tendon, muscle, ligament, cornea, and eyeball in the field of orthopedics or ophthalmology [42]. Continuing development of transducer technology is playing a key role in enhancing the 3-D imaging performance to replace current 2-D sonography by providing real-time capability and interactivity.

5 Conclusions

In this paper, medical ultrasonic imaging transducers were reviewed and their structure, type, and application fields were described. The active and passive components of the transducer were described in detail. The technical issues related to the development of each component were also presented. Continuous development in signal processing and precision machining technology offers new opportunities for enhancing the ultrasound transducer's performance. In the future, more compact and integrated ultrasonic transducers will be studied for generating high-resolution real-time images. It is expected that 3-D ultrasound imaging will be a routine part of patient diagnosis and management in the future. New applications of the transducers are also expected through fusion with other imaging modalities.

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