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Development of 15 MHz 2-2 piezo-composite ultrasound linear array transducers for ophthalmic imaging

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ABSTRACT

High frequency (>15 MHz) ultrasound imaging is capable of producing high spatial resolution B-mode and color Doppler images for the efficient diagnosis of ocular diseases. However, commercial high frequency ultrasound imaging systems cannot currently provide its full capability for ophthalmic applications because either single element or annular array transducers are used in conjunction with mechanical scanning. To overcome the current limitations, 15 MHz 128-element linear array transducers were developed, which are suitable to cover the whole eye from the eyelid in anterior segment to orbital fundus in posterior segment. To achieve the highest electromechanical coupling coefficient, 2-2 piezo-composite was designed and fabricated as an active layer. For the design, the simple model of the thickness-mode oscillation was derived and used in the 1-D model simulator such as PiezoCAD. Also, the usability of the simple model was verified by comparing its simulation results with those by finite element model-based PZFlex. In addition, easy fabrication procedure for the high frequency 2-2 piezo-composite arrays was proposed. In performance evaluation, the pulse-echo response of a representative composite element showed the center frequency of 14.4 MHz and the −6 dB fractional bandwidth of 69.4%. Measured insertion loss was achieved higher than −33.9 dB and crosstalk level between elements less than −30 dB. From the evaluation results, it is expected that the linear array transducer developed using the proposed design and fabrication methods can provide a potential alternative for high frequency ophthalmic imaging with the desired capability.

1. Introduction

Ultrasound biomicroscopy (UBM), a commercialized high frequency ultrasound (>15 MHz) imaging system, has been occupying its unique status in ophthalmic imaging among the other imaging modalities, i.e., optical coherence tomography (OCT), fundus photography, and fluorescence angiography, which are widely used in clinical practice [1,2]; unlike other imaging modalities, the UBM is used for real-time diagnosis of anterior segmental diseases with a relatively deep imaging depth and a large field of view regardless whether the suspicious lesion is in optically transparent or opaque media. The UBM commonly employs single element transducers with a center frequency ranging between 30 and 50 MHz [2]. Note that the spatial resolution of ultrasound images is improved as the center frequency increases. The conventional A-scan, B-scan, and Doppler imaging allow the UBM to well describe not only the complex anatomy but also the pathological functions of the anterior segment of the eye [2]. A large range of anterior segmental diseases from glaucoma to cataract can be diagnosed with this instrument, providing the spatial resolution on the order of 20–40 μm [3]. However, the high spatial resolution is available only in the penetration depth to the front 4–5 mm of the eye due to the frequency-dependent attenuation of ultrasound [4]. Therefore, the center frequency range is not suitable for diagnosis of ocular diseases occurring in the posterior segment of the eye.

The posterior segment has a relatively simple anatomical structure compared with the anterior region; it includes retina, fovea, optic nerve, and retinal blood vessels. Especially, the retina is a layered tissue responsible for delivering vision information to the brain. The systemic diseases regarding the circulation (i.e. diabetic retinopathy from diabetes, hypertensive retinopathy from

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cardiovascular diseases) as well as typical ocular diseases (i.e. macular degeneration and retinal detachment) occur in this area [1]. In order to diagnose the posterior segmental diseases, the ultrasonic transducers with the frequency range of 7–20 MHz are appropriate because its image penetration depth should be at least 30 mm with a few hundred micron resolution. Note that this imaging depth covers the whole eye from the upper eyelid to the orbital apex [5].

Although single element transducers have been generally used in the UBM for ophthalmic imaging, the devices suffer from a limited depth of focus (DOF) due to a single geometrical focus and thus poor spatial resolution outside DOF. To overcome the problem, high frequency ultrasound annular array transducers were fabricated and their capability of dynamic focusing was demonstrated in [6,7]. Like the single element transducers, however, the annular arrays should be mechanically moved to form an image. This causes not only slow frame rate but also incapability of color Doppler imaging (CDI). Without the CDI, the physiological status of the ocular vasculature cannot be efficiently monitored, thus restricting the provision of clinically useful information [8–10].

Also, high frequency ultrasound linear arrays have been developed for small animal imaging and ophthalmic applications [11–13]. Since the electronic scanning is employed for linear array transducers, it is possible to secure the dynamic focusing, high frame rate, and the CDI capability. This leads to improving spatial and temporal resolutions. However, those high frequency arrays are dedicated only for imaging the anterior area since their center frequencies are as high as above 30 MHz. Due to the frequency dependent attenuation, it is difficult to image the posterior segment of the eye with such a high frequency ultrasound. For B-mode imaging, therefore, either single or dual element transducers with the center frequency in the range of 10–20 MHz are used [5,14]. In contrast, the CDI for the retinal vasculatures relies on the 7.5 MHz linear array transducers in spite of weak sensitivity and poor reproducibility caused by a poor spatial resolution in the proposed frequency range [10]. For the purpose of imaging the posterior retinal region, therefore, 20 MHz 192-element convex arrays have been previously proposed to cover the large imaging area around the retina [15–17]; the convex arrays have the same advantages as the aforementioned linear array transducers and also overcome the limitations of the linear arrays such as a limited field of view and a poor spatial resolution.

This paper presents design, fabrication, and performance evaluation of a 15 MHz 128-element linear array transducer for ophthalmic imaging. The design goal was that the linear array transducer can cover up to the posterior fundus from the anterior region of the eye, offering permissible image width for the effective diagnosis of ophthalmic diseases. For this, the penetration depth should be deeper than 30 mm and the spatial resolution should be less than 200 μm. Therefore, the center frequency of 15 MHz was chosen to make a reasonable compromise between resolution and penetration. For the transducer, we designed and fabricated 2-2 piezo-composite as an active layer to achieve high coupling coefficients and large ceramic volume fractions for maximum device capacitance [12]. The 2-2 piezo-composite design was conducted based on a simple model of the thickness-mode oscillations in thin plates of 2-2 piezo-composite; the theoretical background of the simple model is provided in this paper. The validation of the derived model was performed by comparing the simulation results obtained by the simple model with that by a finite element model (FEM): electrical impedance of the composite element and pulse-echo response of the array element. In addition, easy fabrication procedure for high frequency 2-2 piezo-composite arrays is proposed. Finally, the performance and characteristics of the fabricated transducer were ascertained by measuring its center frequency, bandwidth, electrical impedance, crosstalk, and insertion loss.

2. Array design and modeling

2.1. 2-2 piezo-composite design

Both 1-3 and 2-2 piezo-composites have advantages over the conventional piezoceramic materials as follows: high coupling coefficient that enhances the efficiency of electromechanical energy conversion and low acoustic impedance that allows for convenient acoustic matching [11]. While the 1-3 composite has a complex structure with the ceramic rods incorporating into the polymer bed, the 2-2 composite is composed of alternate beam-shape pillars of ceramic and polymer. The latter has been used more frequently in the high frequency transducers because the fabrication is relatively easy compared to the former. Furthermore, it has been recently reported that the performance of the 2-2 composites is better than that of the 1-3 composites in the high frequency range [13]. Therefore, the 2-2 composite was chosen for the 15 MHz linear array transducers in this study.

The most popular method for predicting the performance of the ultrasonic transducers is a 1-D KLM equivalent circuit model, introduced by Leedom, Krimholtz, and Matthaei [18]. A PiezoCAD software package (Sonic Concepts, Woodinville, WA) based on the KLM modeling has proven to be a very useful tool for designing single element transducers in the thickness mode oscillation. By using the basic material properties and layer thickness, we can estimate the performance of the one element of the ultrasonic transducer. However, there is a limitation in using the 1-D equivalent model for designing array transducers. The major drawback is that the crosstalk between the adjacent elements and the effect of the kerf filler cannot be estimated. Also, both 1-3 and 2-2 piezo-composites cannot be simulated with the 1-D equivalent model, which are often used as an active layer in the medical ultrasound imaging transducers. In the case of designing the array transducers, the FEM is known to be more accurate in order to assist the selection of the optimal materials and predict the real array performance. However, accurate prediction is restricted by the accuracy of the material properties used in the simulation [19]. In addition, it is time consuming to achieve an optimal design if starting from scratch. Therefore, we chose to perform initial design through the 1-D model simulation and subsequently FEM-based design. For this, we derived a simple model of the thickness-mode oscillations in thin plates of the 2-2 piezo-composite and used it in the 1-D model simulation.

The piezo-composite can be recognized as a homogeneous material that properly operates in a thickness mode [20,21] if the sizes of ceramic and kerf polymer (i.e., lateral periodicity) are less than a half wavelength. This enables us to calculate the material parameters of the 2-2 piezo-composite by using the equations as follows:

$$\bar{C}_{33}^{E} = \gamma_C \left(C_{33}^{E} - \frac{\gamma_P \left(C_{13}^{E} - C_{12} \right)^2}{\gamma_C C_{11} + \gamma_P C_{11}^{E}} \right) + \gamma_P C_{11}, \tag{1}$$

$$\bar{e}_{33} = \gamma_C \left(e_{33} - \frac{\gamma_P e_{31} \left(C_{13}^{E} - C_{12} \right)}{\gamma_C C_{11} + \gamma_P C_{11}^{E}} \right), \tag{2}$$

$$\bar{\varepsilon}_{33}^{S} = \gamma_C \left(\varepsilon_{33}^{S} - \frac{\gamma_P e_{31}^2}{\gamma_C C_{11} + \gamma_P C_{11}^{E}} \right) + \gamma_P \varepsilon_P, \tag{3}$$

$$\bar{C}_{33}^{D} = \bar{C}_{33}^{E} + \frac{(\bar{e}_{33})^2}{\bar{\varepsilon}_{33}^{S}}. \tag{4}$$

Note that the material parameters used in this paper are defined in the glossary of symbols (see Table 1) and the detailed derivation is presented in Appendix. The new material parameters are associated with the thickness mode oscillation of the