Non-Planar Pad-Printed Thick-Film Focused High-Frequency Ultrasonic Transducers for Imaging and Therapeutic Applications

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***Abstract*—Pad-printed thick-film transducers have been shown to be an interesting alternative to lapped bulk piezoceramics, because the film is deposited with the required thickness, size, and geometry, thus avoiding any subsequent machining to achieve geometrical focusing. Their electromechanical properties are close to those of bulk ceramics with similar composition despite having a higher porosity. In this paper, pad-printed high-frequency transducers based on a low-loss piezoceramic composition are designed and fabricated. High-porosity ceramic cylinders with a spherical top surface are used as the backing substrate. The transducers are characterized in view of imaging applications and their imaging capabilities are evaluated with phantoms containing spherical inclusions and in different biological tissues. In addition, the transducers are evaluated for their capability to produce high-acoustic intensities at frequencies around 20 MHz. High-intensity measurements, obtained with a calibrated hydrophone, show that transducer performance is promising for applications that would require the same device to be used for imaging and for therapy. Nevertheless, the transducer design can be improved, and simulation studies are performed to find a better compromise between low-power and high-power performance. The size, geometry, and constitutive materials of optimized configurations are proposed and their feasibility is discussed.**

I. Introduction

Thick-film (TF) technologies have been shown to allow high-performance piezoelectric layers to be produced with thicknesses in the range of 10 to 50 μm, and several applications have been proposed [1], [2]. Using such films, ultrasonic transducers have been developed for high-resolution imaging [3]–[6]. Screen-printing [6] has allowed planar transducers to be obtained, whereas composite sol-gel [7]–[13] and, more recently, electrophoresis [14], [15], ink-jet printing [16], and pad-printing [17] technologies have been investigated to deposit films on non-planar substrates. One of the advantages of pad-printing is that it can easily be implemented at industrial scales because of its low cost and relatively simple process. In particular, this technology has been used to produce thick films on a curved, porous ceramic substrate that served as a transducer backing. Transducers fabricated in this fashion have been shown to be suitable for medical imaging [17]. The goal of this work is to determine if such transducers can be considered for therapeutic applications for which efficiency becomes an important issue. After a description of the fabrication process, transducer pulse-echo response is measured and images are produced. High-power operation is then investigated. Finally, alternative designs are proposed and simulation results are presented to evaluate the compromises between imaging and high-intensity performance.

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II. Methods

A. Transducer Fabrication

The transducer structure is shown in Fig. 1. Fig. 2 shows some steps of the fabrication process and Fig. 3 shows a cross section of the front face. The transducer backing consists of a porous ceramic cylinder, the front face of which was machined to have a spherical curvature with a radius of 12.5 mm. The active piezoelectric films on the curved face of the cylinder, as well as the top and bottom electrodes, were deposited using pad printing. Contact electrodes were added along the side of the cylinder by silk-screen printing. A thorough description of the manufacturing method can be found in [17]. The backing material (SU37, InSensor, Kvistgaard, Denmark) was based on Ferroperm composition Pz37 (Ferroperm Piezoceramics A/S, Kvistgaard, Denmark), a porous engineered structure version of lead zirconate titanate (PZT). The thick film InSensor TF2100 was a Navy Type-I powder modified for thick-film manufacturing and formulated into a pad-printable paste. The film thickness is defined by the number of prints that are stacked and is chosen to obtain a transducer frequency of around 20 MHz; 25 prints were required to obtain a film thickness of around 40 μm after sintering. The bottom electrode is gold, with a thickness of approximately 4 μm, and the top one is silver; their contact electrodes made of the same material. The silver electrodes are approximately 1 μm thick. The aperture, defined by the diameter of the top electrode, is 2.5 mm.

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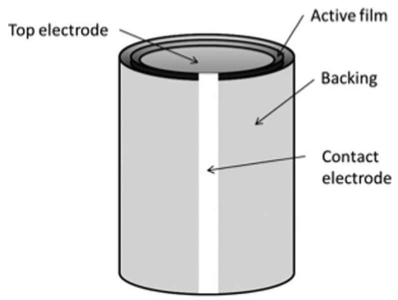

Fig. 1. Transducer structure. Cylindrical backing with spherically machined front face, piezoelectric film between back and front electrodes, and contact electrode connected to back electrode (cylinder diameter 3 mm, electrode diameter 2.5 mm, radius of curvature 12.5 mm, and height of cylinder 5 mm).

B. Materials and Transducer Characterization

The properties of materials used within the transducer structure were determined using the procedures described in [5] and [17]. To deduce the electromechanical properties (corresponding to the thickness mode), measurement of the electrical impedance as a function of frequency was performed on a representative structure. Then, the value of the thickness of each layer as well as the volume fraction of porosity in the substrate and the thick film were deduced to evaluate the wave velocity, density, and acoustic impedance.

The transducer was characterized for pulse-echo performance under optimized conditions. To this end, it was excited with an impulse from a pulser/receiver (Panametrics 5900, Olympus NDT, Waltham, MA). A planar quartz reflector was positioned normal to and in the focal plane of the transducer and the resulting echo was digitized. From this signal, the bandwidth and center frequency were determined.

C. Imaging Performance

The imaging performance of the transducer was first evaluated using a phantom with spherical, anechoic inclusions of uniform size. The phantom consisted of slabs of tissue-mimicking material with each slab containing a random distribution of uniform-diameter anechoic spheres. The phantom allowed for a quantification of the minimum feature size that can be resolved by a transducer. The feature size resolution is related to the 3-D beam properties of the focal zone. Second, images were acquired from an *ex vivo* rabbit eye, an externalized mouse embryo, and an *in utero* mouse embryo.

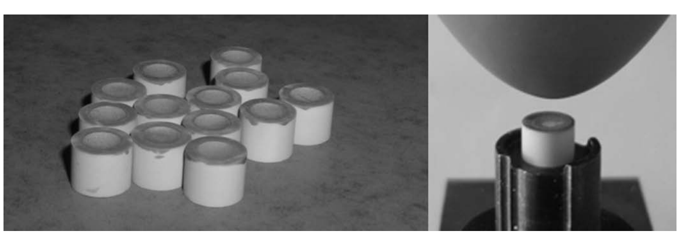

Fig. 2. (left) Porous ceramic cylindrical substrates used as backings, with spherically machined surface. (right) Structure during film deposition process by pad-printing. The silicone rubber pad (shown at top of image) is being lowered to deposit a layer of ceramic slurry on the curved surface of the substrate, upon which the rear electrode has previously been deposited.

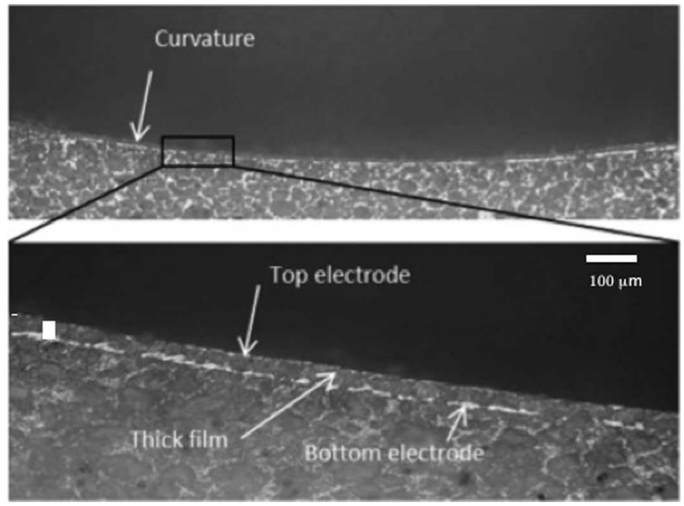

Fig. 3. Cross section of transducer's front face, observed by optical microscope.

D. High-Intensity Performance

To evaluate the high-intensity performance, the acoustic pressure at the geometric focus of the transducer was measured for different values of input voltage. A one-period sine wave was generated and amplified (150A100B, Amplifier Research Corp., Souderton, PA). A 200-μm aperture needle hydrophone (Precision Acoustics Ltd., Dorchester, Dorset, UK) calibrated up to at least 20 MHz was used. It was placed at the location of highest pressure, i.e., approximately 12 mm from the transducer surface, facing its center.

E. Simulations

To explore the potential of thick-film structures for combined imaging and therapy applications, the intrinsic performance of four transducer structures has been simulated using a KLM-based model [18]. The transducer diameter was fixed at 2.5 mm and an operating frequency of around 17 MHz was targeted. The input parameters were determined as described in Section II-B.

The impulse responses in pulse-echo mode and the frequency domain curves were calculated as well as the efficiency, i.e., the ratio of acoustic power generated in water to the absorbed electrical power at the frequency of 17 MHz.

III. Results

A. Materials and Transducer Properties

The piezoelectric film's coupling factor in thickness mode was measured at 47%, its longitudinal wave velocity

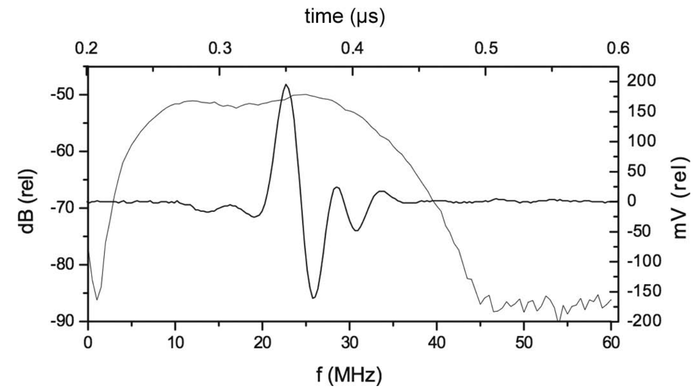

Fig. 4. Quartz plate pulse/echo response using optimized pulser-receiver settings revealed a center frequency of 19.5 MHz and −6-dB bandwidth of 135%.

and acoustic impedance were determined respectively at 3250 m/s and 17 MRa. The porous PZT substrate (used as transducer backing) showed an acoustic impedance of 16 MRa.

The pulse/echo response is shown in Fig. 4. When excited in optimized conditions, it revealed a center frequency of 19.5 MHz and a −6-dB bandwidth of 135%. This response displays a significantly higher axial resolution (and bandwidth) than that of KLM simulations in a standard 50-Ω electrical environment (see Section III-D), because of optimized electrical matching and pulser-receiver settings.

B. Imaging Performance

Before performing animal imaging, a high-frequency phantom [19] was used to estimate the 3-D resolution performance of the transducer. The transducer was mounted to a motorized stage and then RF data were acquired as the transducer was laterally translated across the phantom.

The results of imaging the tissue-mimicking phantom are shown in Fig. 5. The phantom images showed that the transducer was able to resolve the 825-µm [Fig. 5(a)] and 530-µm-diameter anechoic spheres [Fig. 5(b)] but not the 400-µm-diameter anechoic spheres [Fig. 5(c)]. The slab with no anechoic spheres [Fig. 5(d)] showed a speckle pattern similar to the slab with 400-µm spheres, providing further evidence that the 400-µm spheres were not resolved. The phantom results were consistent with the theoretical lateral beam width of approximately 0.5 mm.

After the phantom experiments, the transducer was used for animal imaging. Fig. 6(a) shows an image of an *ex vivo* rabbit eye with the key anatomical features indicated. Fig. 6(b) shows an *in vivo* left ventricle of an adult mouse. Fig. 6(c) shows an externalized, *in vivo* mouse embryo with the embryo, umbilical vessels, and placenta visible. Fig. 6(d) shows another image of a mouse embryo, this time with the embryo *in utero*. The acoustic attenuation within the dermis of the mother reduces signal strength but the embryo was still visible, as was the uterus.

The imaging examples of Fig. 6 exhibit a reasonable contrast and the different anatomical structures are clearly distinguished, showing that the transducer used in these studies delivered sufficiently high signal-to-noise ratio and axial resolution to create usable images. However, the transducer's high f-number (focal length/aperture) of 6 made it less than ideal for pure imaging applications

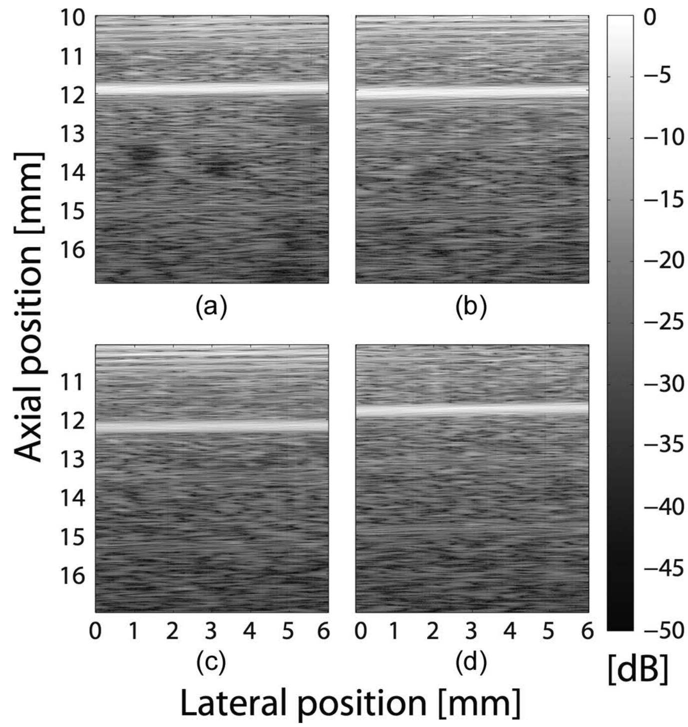

Fig. 5. Set of images showing visibility of the anechoic spheres within the tissue-mimicking material as a function of sphere diameter. (a) The 825-µm spheres were resolved, as were (b) the 530-µm spheres. (c) The 400-µm spheres were not resolved and (d) the background material with no spheres showed a speckle pattern similar to the 400-µm sphere case. (The bright echo around 12 mm is a secondary reflection of the front surface of the phantom).

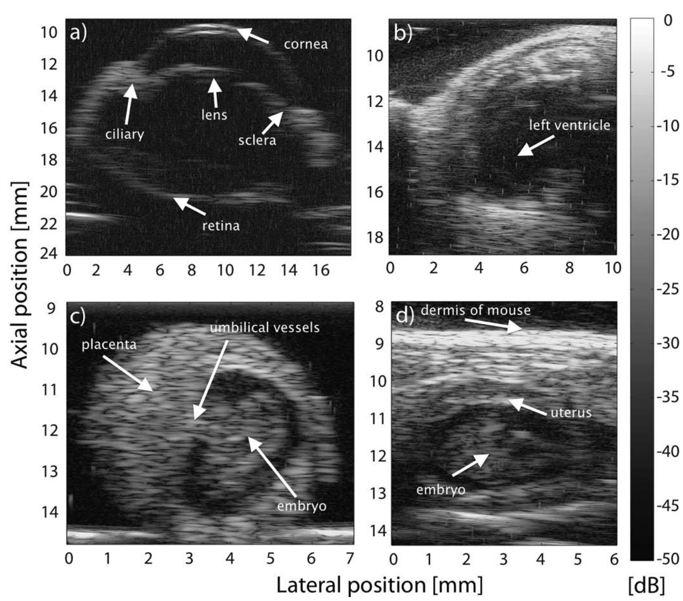

Fig. 6. (a) *Ex vivo* rabbit eye, (b) *in vivo* left ventricle of an adult mouse heart, (c) externalized *in vivo* mouse embryo, and (d) *in utero* mouse embryo imaged at 20 MHz.

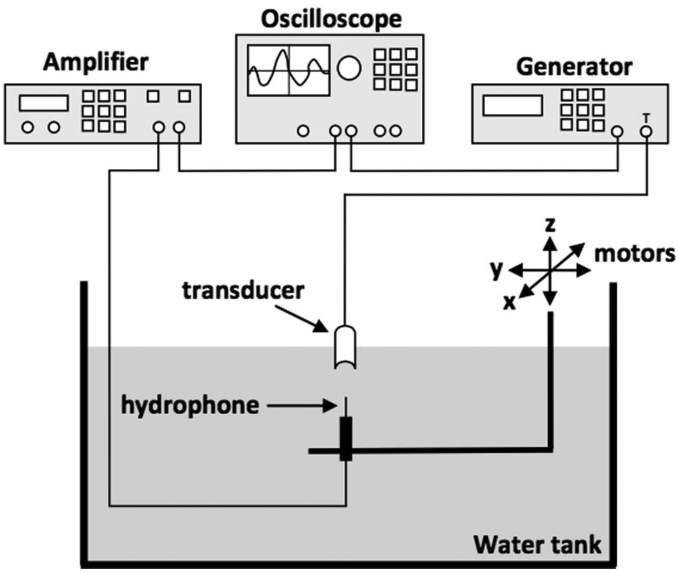

Fig. 7. Experimental setup for pressure measurement at the transducer's geometric focus: a high-voltage burst is sent to the transducer and a calibrated hydrophone is placed at the point of highest pressure, i.e., at the focal point, in water.

because the lateral resolution is on the order of 0.5 mm. Typical high-frequency imaging transducers at these frequencies have lateral resolutions on the order of 100 to 200 μm [20]. Nevertheless, these initial imaging studies show that thick-film transducers were able to generate acoustic signals sufficient for imaging applications.

C. High-Power Operation

The experimental setup used to evaluate the high-intensity performance of the pad-printed transducers is shown in Fig. 7. The results of acoustic pressure at the geometric focus measured for different values of input voltage are presented in Fig. 8. The transducer delivered pressure values of greater than 3 MPa for an input voltage of 140 V peak-to-peak. A fairly linear pressure versus voltage curve with no saturation was observed

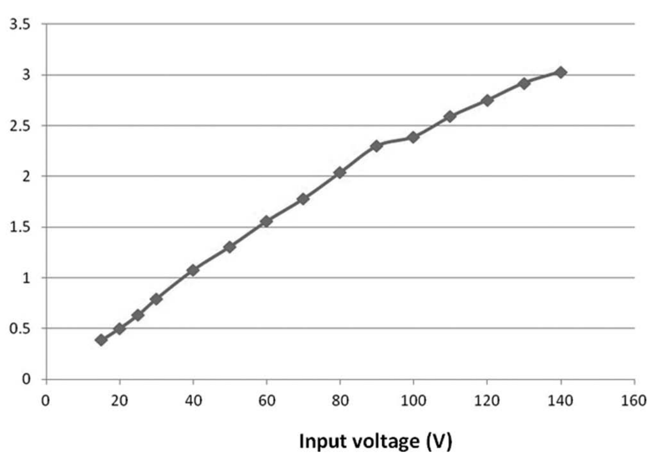

Fig. 8. Acoustic pressure (in megapascals) at focal point versus input voltage (in volts).

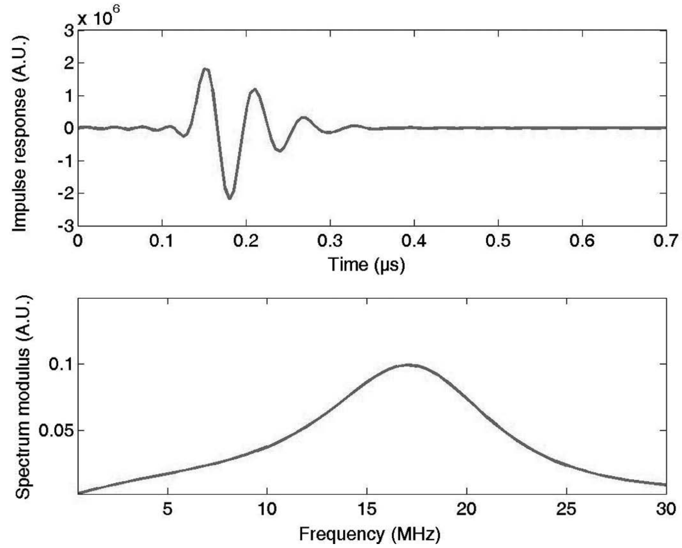

Fig. 9. Pulse-echo and frequency response of transducer with standard backing (16 MRa) and thin bottom electrode (4 μm).

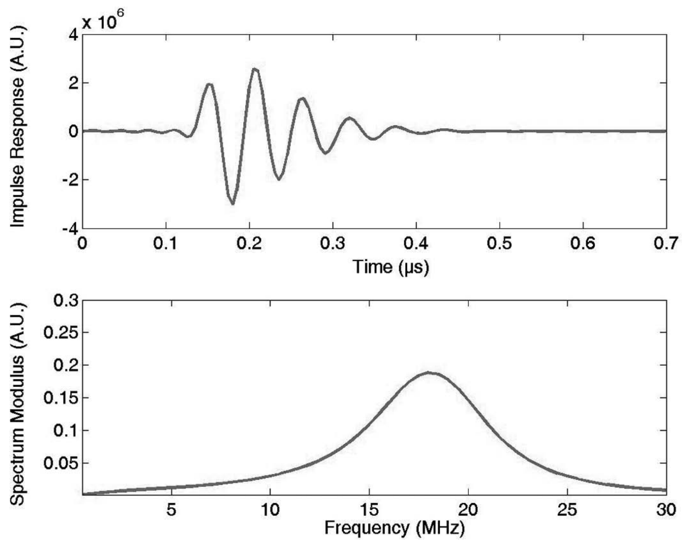

Fig. 10. Pulse-echo and frequency response of transducer with high-porosity backing (10 MRa) and thin bottom electrode (4 μm).

D. Optimization Simulations

To explore different trade-offs between imaging and high-intensity performance, alternate designs were evaluated through a simulation study. Simulation results are shown in Figs. 9, 10, and 11; the −6-dB bandwidth in pulse-echo mode and efficiency in transmit mode at 17 MHz are summarized in Table I. Configurations were defined considering two backings: one with the current material (16 MRa) and the other with a higher porosity that has a lower acoustic impedance (10 MRa). The thickness of the gold bottom electrode was either the typical value of 4 μm or a lager value of 15 μm. The currently used configuration produced a short impulse response that corresponded to a −6-dB bandwidth of 60% (Fig. 9) and the efficiency was 11%. The high-porosity-backing/thick-

TABLE I. Simulated Transducer Performance.

Transducer	−6-dB bandwidth (%)	Efficiency at 17 MHz (%)
Standard backing/thin bottom electrode	62	11
High-porosity backing/thin bottom electrode	37	15
Standard backing/thick bottom electrode	31	23
High-porosity backing/thick bottom electrode	25	31

bottom-electrode structure exhibited a longer impulse response, a −6-dB bandwidth around 25%, and an efficiency of more than 30% (Fig. 11). This would improve performance for therapeutic applications, when compared with the conventional configuration, and would generate almost three times more acoustic power for a given electrical input while reducing internal heating of the transducer by 25%. However, the imaging performance would deteriorate because of the increase in pulse length and resulting degradation in the axial resolution.

Fig. 10 shows the performance of a configuration that exhibits intermediate performance, and could be the basis for the development of a high-frequency transducer that could be used both for imaging and therapy: high-porosity backing and standard thickness (4 μm) back electrode.

IV. Conclusions

The initial pad-printed transducer design, with a 16-MRa backing and a 4-μm-thick gold back electrode, is well adapted to imaging applications, albeit with a suboptimal f-number that can easily be corrected by modifying the curvature and diameter of the backing material. However, in the context of intensity, even though relatively high pressures were generated over short time intervals, excessive heating would occur when high power was delivered over longer durations. Indeed, the efficiency of this transducer is expected to be quite low (11% as predicted by simulation).

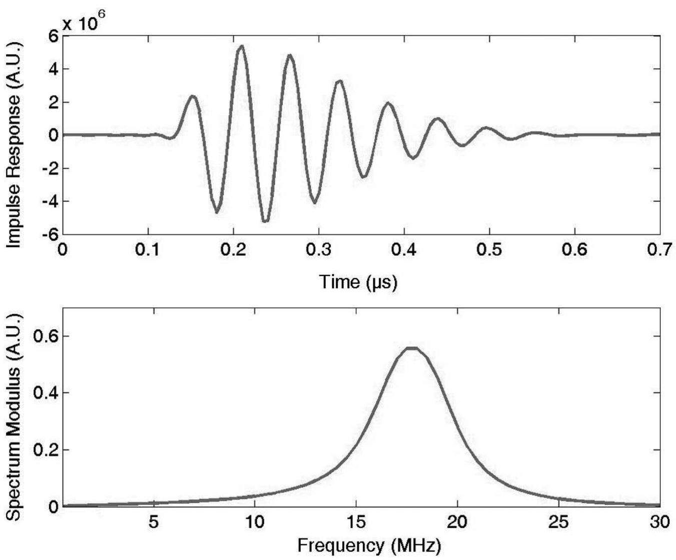

Fig. 11. Pulse-echo and frequency response of transducer with high-porosity backing (10 MRa) and thick bottom electrode (15 μm).

It has been shown that such piezoelectric thick-film structures on porous ceramic substrates can be optimized for therapeutic applications through changes in substrate porosity and bottom electrode thickness. Simulation showed that modifications to the transducer design would result in satisfactory performance for both applications, although at a moderate sacrifice in axial resolution. Such a transducer will be fabricated in the near future, based on the same piezoelectric thick film but using a high-porosity substrate as backing, with acoustic impedance around 10 to 12 MRa, i.e., the lowest feasible value that allows machinability. Its diameter and focal distance will also be modified to improve lateral resolution for imaging applications and to increase spatial peak intensity for therapeutic applications.

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Rasmus Lou-Møller received his M.Sc. degree in applied physics from the Technical University of Denmark in 2003. Since 2004, he has been working on the development of piezoceramic thick films and applications utilizing this technology at Meggitt A/S. He is currently head of the department InSensor of Meggitt A/S.

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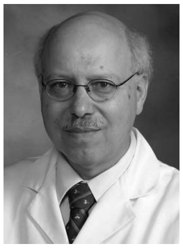

Ronald H. Silverman graduated from Brooklyn College (Brooklyn, NY) with a B.S. degree in chemistry in 1971. He received an M.S. degree in bioengineering from the Polytechnic Institute of New York (Brooklyn, NY) in 1979 and a Ph.D. degree in computer science from the same institution in 1990. Dr. Silverman worked in the Department of Ophthalmology of the Weill Cornell Medical College (New York, NY) from 1983 to 2010, where his research centered on ophthalmic applications of ultrasound. He has continued his work at the Department of Ophthalmology at Columbia University (New York, NY) as Professor of Ophthalmic Science since 2010. Dr. Silverman has also had a long-term relationship with the Frederic L. Lizzi Center for Biomedical Engineering of Riverside Research (New York, NY). Dr. Silverman's research interests primarily involve high-resolution ultrasound imaging, tissue characterization, bioeffects, ultrasound-enhanced drug delivery, and acoustic-optical interactions. His interdisciplinary work ranges from basic engineering to animal and clinical studies. Dr. Silverman is a Fellow of the American Institute of Ultrasound in Medicine and of the American Institute for Medical and Biological Engineering. He has served as a member of the Program Committees of the Association for Research in Vision and Ophthalmology and the International Society for Imaging in the Eye. He is a member of the IEEE, SPIE, the American Academy of Ophthalmology, the Acoustical Society of America, and the American Association for the Advancement of Science.

Wanda W. Wolny graduated from the University of Technology in Warsaw. She has been with Ferroperm since then, first as R&D Engineer, then as Department Manager, and finally as Technical Director. When Ferroperm Piezoceramics was created in 2001, she was a Managing Director until 2006, then Managing Director for R&D until 2011. During the last year and a half, she has been Director for Materials Science for Meggitt Plc and Vice President of Technology. Mrs. Wolny has coordinated a number of EU projects including Polecer and MIND networks. She was convenor for TF WG standard preparation resulting in 3 IEC standards on piezoelectric materials, was a Member of the EU Business Chamber from 2001 until 2010, and has been a Strategic Advisory Board member of the CIP since 2006. Mrs. Wolny is President of The Piezo Institute.