Wide-Band Piezoelectric Polymer Acoustic Sources

PETER A. LEWIN, SENIOR MEMBER, IEEE, AND MARK E. SCHAFER, MEMBER, IEEE

Abstract—**The design of a wide-band acoustic source made of piezoelectric polymer (PVDF) material is described. The source was developed for the characterization and absolute calibration of ultrasonic hydrophone probes. Construction details are described and performance characteristics of the wide-band PVDF transmitter, including its transmitting voltage response and directivity patterns, are compared with theoretical predictions in the frequency range up to 40 MHz. The Krimholtz–Leedom–Mattaei (KLM) model was used to examine the influence of the PVDF polymer film thickness, the backing acoustic impedance, the cable length, and the electrical source resistance on overall transmit transfer characteristics. A comparison is made with traditional piezoelectric ceramic acoustic sources, and it is shown that piezopolymer transmitters exhibit some improved properties and are well suited for certain ultrasound dosimetry applications. In particular, the polymer sources have been found useful in measurements based on swept-frequency excitation. Those measurements allow characterization of transmitters and receivers to be performed as a virtually continuous function of frequency.**

I. Introduction

ONLY A DECADE ago, the frequencies used in medical ultrasound imaging rarely exceeded 2.25 MHz. Since then, ultrasound imaging systems have progressed towards the routine use of much higher frequencies. For example, general-purpose imaging and flow mapping (Doppler) devices employ frequencies as high as 7.5 MHz. Specialized scanners for "small parts" imaging use 10-MHz transducers, and some opthalmological systems generate 20-MHz ultrasound waves.

Another source of higher frequency ultrasound energy is associated with nonlinear wave propagation phenomena. Evidence has been accumulating over the past few years that these phenomena can no longer be ignored in medical ultrasound [1]. Ultrasound imaging systems use focused transducers which can produce instantaneous peak pressure amplitudes sufficiently high to induce nonlinear propagation phenomena. Those phenomena are associated with the generation of higher harmonics in the propagation medium [1]–[5]. The resulting distortion in the pressure–time waveforms indicates that the ultrasound waves may contain frequency components of tens of megahertz. In addition, lithotripters used to disintegrate kidney stones without the need for open surgery make use of acoustic shock waves which exhibit rise times as brief as 30 ns; this means that the frequency content of the wave is well beyond 30 MHz. Consequently, the ultrasound probe used to record the nonlinear pressure–time waveform should be sufficiently wide-band to detect all the higher harmonics present in the waveform. The contribution of the harmonics can be approximated using well-known theory [1], [3], [5]. The theory indicates that neglecting the sixth and higher harmonics introduces approximately a five-percent error in the measurement of peak intensity of the distorted wave. Thus the quantitative determination of the fifth harmonic should be sufficient to account for nonlinear propagation phenomena. Consequently, a correct measurement of the pressure–time waveform generated by an imaging transducer of nominal frequency of 5 MHz will require a minimum hydrophone bandwidth of 25–30 MHz if nonlinear distortion is present.

Commercially available hydrophone probes are usually not calibrated beyond 10 MHz. One reason for this, apart from all the practical problems associated with the actual calibration [6], [7], is a lack of broad-band acoustic sources. Imaging transducers, although often described as having broad-band characteristics, can hardly serve as acoustic sources over the full frequency range of interest. Their bandwidth is almost always limited due to the matching layers and the resonant character of piezoceramic material used (e.g., lead zirconate titanate (PZT)). The 3-dB bandwidth of a high-quality imaging transducer is on the order of 70 percent of the fundamental frequency, and thus a single transducer of this type is incapable of providing sufficient acoustic output over the 1–40 MHz range.

Although currently available NBS calibrated acoustic power sources [8] can be used up to 21 MHz, these sources cannot be used in calibration of ultrasonic probes as a continuous function of frequency [9]–[11]. This is because the sources are highly resonant devices and are calibrated at discrete frequencies (i.e., 1-MHz intervals). This represents a drawback, since the calibration of ultrasonic probes at such intervals will overlook any rapid variations in frequency response of the hydrophone, due, for instance, to spurious mechanical or radial mode resonances [12]. Providing that a wide-band acoustic source is available, this drawback can be overcome by using the calibration technique which combines the two-transducer reciprocity technique [7] and the time-delay spectrometry (TDS) approach [9]–[11]. This technique allows hydrophone probe free-field parameters, such as frequency response, directivity, and effective radius, to be determined as a continuous function of frequency in reverberant en-

Manuscript received February 18, 1987; revised June 23, 1987.
P. A. Lewin is with the Department of Electrical and Computer Engineering, Biomedical Engineering and Science Institute, Drexel University, Philadelphia, PA 19104.
M. E. Schafer is with Sonic Technologies, Inc., Horsham, PA 19044.
IEEE Log Number 8718127.

0885-3010/88/0300-0175$01.00 © 1988 IEEE

vironments. The theoretical background and fundamental limitations of the TDS technique are thoroughly discussed in a separate paper [11]. The typical overall uncertainties in the reciprocity calibration performed using the TDS measurement procedure have been estimated to range between ± 1 dB (at 1 MHz) and ± 1.3 dB (at 15 MHz). Measurements undertaken by an independent laboratory in the frequency range from 1 to 10 MHz [9] confirm these uncertainties. As mentioned above, although the combination of the TDS and reciprocity techniques allows absolute calibration at frequencies up to 15 MHz [12], the ultrasonic hydrophone probes currently available are in general calibrated as a continuous function of frequency from 1 to 10 MHz. The National Physical Laboratory (U.K.) can perform discrete frequency calibrations at 1-MHz intervals up to 15 MHz.

From the above, there is a need for acoustic sources that can produce predictable and uniform acoustic output in the widest possible bandwidth. Such sources will allow hydrophone calibration over a wider frequency range (to 40 MHz) using methods based on the swept-frequency systems and the TDS technique, without the necessity of changing transmitters [9]–[13]. Apart from applications in ultrasound dosimetry and exposimetry, wide-band sources may also be useful in tissue characterization procedures, as they facilitate measurements of medium propagation parameters such as attenuation as a continuous function of frequency.

This paper describes the design, construction, and performance of wide-band piezoelectric acoustic sources (transmitters) made using polyvinylidene difluoride (PVDF). These sources are intended to provide absolute calibration of ultrasound hydrophone probes (receivers). Although the theoretical predictions indicate that the polymer sources described in the present work can be used at frequencies beyond 40 MHz, the upper cutoff frequency of the spectrum analyzer used here was limited to 40 MHz. Since the wide-band PVDF sources were characterized using broad-band miniature polymer hydrophone probes, a brief summary of the basic properties of the PVDF hydrophone probes follows.

II. Broad-Band PVDF Polymer Ultrasound Receivers

In the time since the pioneering ideas presented in 1977 [14], several PVDF hydrophone designs have been developed. The designs employ either the original hoop membrane approach [15], [16] or needle-type construction [17]. Comparison of the performance properties and applications of the two designs can be found in [18]. Briefly, the PVDF hydrophone probes currently available exhibit uniform frequency response (to within ± 1.5 dB) over a frequency range which extends beyond 10 MHz [12], [19]. Their absolute calibration as a function of frequency is usually known in the frequency range from 1 to 10 MHz. Recently, the results of a calibration performed as a virtually continuous function of frequency in the 1–15 MHz frequency range were published [18]. The use of

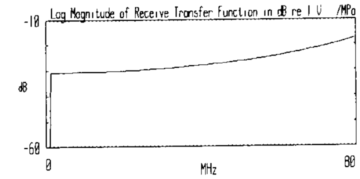

Fig. 1. Theoretically predicted receiving voltage sensitivity of 9-μm 0.5-mm-diameter PVDF membrane hydrophone terminated into 1 MΩ through 1 m of coaxial cable.

PVDF hydrophone probes provided the evidence that indeed the analysis of ultrasound pulses used in medical imaging systems has to account for nonlinear propagation phenomena.

The angular response of the polymer probes is, especially in the case of needlelike design, close to the theoretical predictions, based on a uniform piston receiver model [17], [20], [21]. Recently, a modification of the needle-type probe design was suggested [22]. In this design, the polymer sensor is mounted on a convex backing support, and hence the probe shape corresponds to a fraction of a sphere. This construction should, in principle, exhibit further improvement in directivity patterns, bringing them closer to the more desirable spherical shape.

The hydrophone used here was made of PVDF material and was similar to the miniature needle-type probe described in [17]. It employed a 9-μm-thick polymer film, and its effective diameter was determined to be approximately 0.5 mm. In the frequency range from 1 to 15 MHz this hydrophone was absolutely calibrated by the reciprocity method combined with the TDS technique [9]–[13]. Beyond 15 MHz a calibration by substitution was carried out using a prototype 9-μm PVDF membrane hydrophone as a reference and assuming that, in general, the membrane hydrophone exhibits uniform and flat sensitivity to about 70–80 MHz. The support for this assumption comes from Fig. 1, which shows a simulation of the receiving frequency response for a 9-μm membrane hydrophone. The results of this simulation are consistent with the data published in [21], [23]. In [23] the frequency response beyond 15 MHz was obtained for several ultrasonic hydrophone probes by comparison of the voltage measured at the terminals of the probes against the corresponding voltage produced by a reference membrane hydrophone. The frequency response of the reference hydrophone was theoretically predicted [19].

The miniature needle-type hydrophone (the Danish Institute of Biomedical Engineering, Copenhagen, DK-2605, Denmark) was chosen for the calibration of the transmitter because of its ease of positioning in the measurement arrangement used. Over the frequency range from 1 to 15 MHz the frequency response of the needle-type hydrophone was essentially flat (± 1.5 dB). Between 15 and 40 MHz the frequency response of the needle-type hydrophone was adjusted to account for a theoretically predicted increase in the sensitivity of the reference membrane hydrophone (see Figs. 1 and 2).

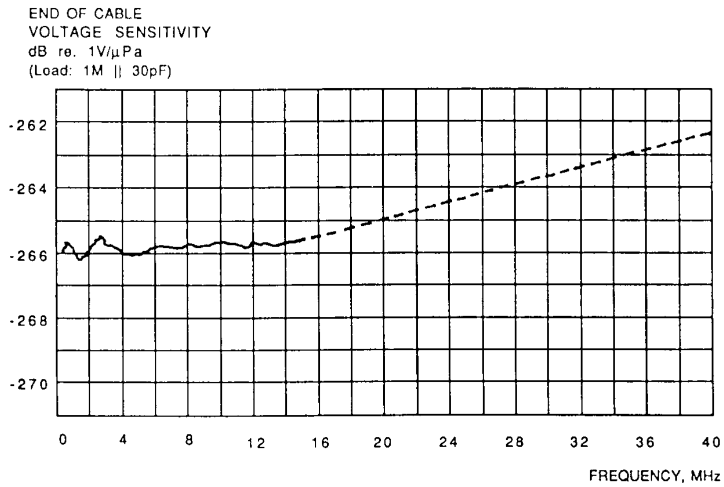

Fig. 2. Experimentally determined overall frequency response of needle-type miniature PVDF hydrophone used here as calibrated receiver. Frequency range of absolute calibration using time-delay spectrometry and reciprocity technique: 1-15 MHz. In frequency range 15-40 MHz, frequency response was derived; for details see Section II.

III. Wide-Band PVDF Polymer Transmitters

In the past decade there have been relatively few attempts made to employ PVDF polymer material in medical imaging transducers [24], [25], nor was PVDF seriously considered for broad-band design applications. This is so despite the fact that the PVDF material has several features that make it useful in medical imaging. First of all, the nature of the PVDF material facilitates construction of transducers that have short impulse responses, without the need for matching layers. PVDF's acoustic impedance is relatively close to that of tissue (approximately 4.5 MRayl), thus minimizing reflections at the interface between the imaging transducer and tissue. This is discussed further in Section III-A. Finally, PVDF is mechanically flexible, thus potentially improving new transducer designs. However, a relatively low electromechanical coupling factor, which reduces transmitting efficiency, has contributed to limited interest in using PVDF in transmitter applications. Since the primary goal in the design to be described was bandwidth maximization, this relatively poor transmitter performance (in comparison with the conventional piezoelectric ceramic transducers) was not considered to be a serious drawback. Acoustic power produced by the wide-band transmitter was of secondary interest because of the use of the inherently narrow-band calibration procedure based on time-delay spectrometry [10], [11], [13], [26], [27]. In practice, the TDS procedure allows a signal-to-noise ratio of at least 60–75 dB to be maintained in the 1–40 MHz frequency range.

A. Theoretical Modeling

The theoretical calculation scheme for design optimization employed the Krimholtz-Leedom-Mattaei (KLM) model proposed in [28] and applied recently in [29], [30]. This model allows for the acoustic source to be considered using the transmission line approach and is particularly suited for medical imaging transducer design. The KLM model has been derived from the well-known and widely used Mason model [31]. The design procedure for ultrasonic transducers is based on the one-dimensional approximation of several layered structures. In general, those layers consist of piezoelectric material, acoustic impedance matching layers, acoustic backing layers, and front port loading material (propagation medium). A brief comparison with ceramic transducers serves to emphasize the simplifications in the design obtained when using PVDF material. Most transducers using piezoelectric ceramics, such as lead zirconate titanate (PZT), will deliver maximum acoustic power to the load when the thickness of the piezoelectric material is equal to half the acoustic wavelength in the material. However, because of the large acoustic impedance mismatch between the ceramic (typically, 30 MRayl) and the propagation medium (approximately 1.5 MRayl), a significant part of the acoustic wave generated in the ceramic material is reflected. These reflections can introduce ringing in the transducer impulse response, which decreases the transducer bandwidth.

The overall performance of the transmitter is largely controlled by the thickness of the element used, the acoustic impedance of the backing material, and the electrical matching. These parameters were input variables to the KLM model which was used to determine transmit transfer function (transmitting voltage response) in dB re 1 Pa cm/V. This function defines acoustic output at a given axial distance from the transmitter for a unity voltage excitation signal. A more detailed definition of the transmitting voltage response is given in Section V-A. Unlike the design optimization procedure for medical imaging transducers, which requires neutralization of the clamped capacitance of the piezoelectric material by using a parallel inductance, no inductive components were used in the simulation described below to maximize the bandwidth.

A number of transmitter parameters were simulated, including effective diameter, PVDF film thickness, acoustic backing material, electrical source resistance, and cable length. Characteristic parameters of the PVDF material (Kynar® and Solef®) were taken from data supplied by the manufacturers.

The diameter of the active element was set to 10 mm. The size limitation comes from the consideration of the distance between the transmitter and the hydrophone probe and the location of the transmitter's near-to-far field transition region. With the upper frequency of interest of 40 MHz, it would be impractical to have effective transmitter diameter larger than 10 mm. The 10-mm diameter requires approximately 50-cm separation distance between the transmitter and a receiver to approximate free-field measurement conditions in a water tank.

The thicknesses of the PVDF material governs the bandwidth and, to a certain extent, the transmitting efficiency. The simulation results presented below are based on 9- and 28-μm thicknesses of the polymer film, since those are the most commonly available thicknesses. Acoustic impedance of the backing was simulated in the range from 1.5 to 40 MRayl. Note that while the 1.5-MRayl impedance corresponds to a half-wavelength resonance structure, the 40-MRayl backing impedance shifts the fundamental frequency to approximately the quarter-wave resonance value. Source resistance was varied from 50 to 1000 Ω, and the length of the 50-Ω coaxial cable terminating the sensor element was varied from 0.4 to 1.5 m. Selected results of the simulations are presented in Figs. 3–8.

The influence of the thickness of the polymer film can be readily seen from Figs. 3 and 4. In Fig. 3 the transmitting response for a 9-μm film with a 12-MRayl backing was calculated. The half-wavelength resonance frequency for a 9-μm film corresponds to approximately 120 MHz. With the backing used the calculated resonance frequency dropped to approximately 56 MHz, beyond the scale shown in Fig. 3. In Fig. 4 the thickness used was 28 μm, with the corresponding half-wavelength resonance at approximately 38 MHz. With a 12-MRayl backing the resonance frequency dropped to approximately 16 MHz. From the figures it is obvious that the thinner film leads to a decrease in the acoustic output at frequencies below 20 MHz; however, the 9-μm film response is much more uniform over the whole frequency range. Note that in both figures the vertical scales are in dB.

The influence of the acoustic backing material is apparent from a comparison of Figs. 4 and 5. The plots of Figs. 2 and 3 correspond to a 28-μm film thickness backed

®Kynar is a registered service mark of the Pennwatt Corporation, King of Prussia, PA. Solef is a registered service mark of the Solvay Corporation, Brussels, Belgium.

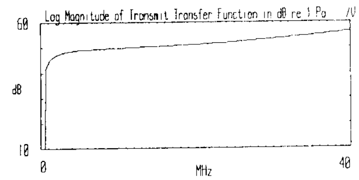

Fig. 3. Simulated transmitting voltage response of 10-mm-diameter 9-μm-thick PVDF polymer acoustic source in water at axial distance of 50 cm. Backing impedance is 12 MRayl.

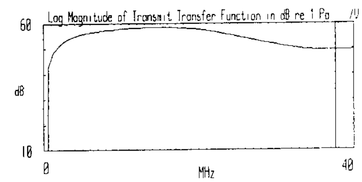

Fig. 4. Simulated transmitting voltage response of 10-mm-diameter 28-μm-thick PVDF polymer acoustic source in water at axial distance of 50 cm. Backing impedance is 12 MRayl.

by an acoustic impedance of 12 and 40 MRayl, respectively, and a source resistance of 150 Ω. The lower backing impedance produces a more uniform and wide-band frequency response at the slight expense of the acoustic output at lower frequencies.

The influence of the electrical matching can be evaluated from Figs. 6 and 7. In Fig. 6 the frequency response plot was obtained using a source resistance value of 50 Ω. The plot in Fig. 7 was calculated with the resistance value increased tenfold to 500 Ω. As could be expected, the acoustic output of the transmitter decreases when the source resistance increases. On the other hand, a higher source resistance results in a more uniform transmitting frequency response at the lower megahertz range of frequencies.

The influence of the cable length between the PVDF sensor element and the voltage generator was examined by calculating the impulse response for different cable lengths. The results of the simulations are shown in Figs. 8(a) and 8(b); the former shows the transmitter impulse response with a 0.5-m cable, while the latter shows the transmitter impulse response with a 2-m cable. The results clearly indicated that the shortest possible cable length is desirable to obtain the shortest impulse response. However, for practical reasons the cable length was set to 1 m.

Based on the results of the simulations described, a 9-μm-thick PVDF polymer film was selected for the wide-band acoustic source. Although this thickness does not yield the highest acoustic output level it was considered appropriate for TDS calibrations because it provided the widest uniform frequency response. The value of acoustic impedance of the backing was selected close to 12 MRayls (see Figs. 4 and 5). The value of the source resistance required to optimize the bandwidth was experimentally determined to be approximately 150 Ω. The construction of the wide-band transmitter is described in the next section.

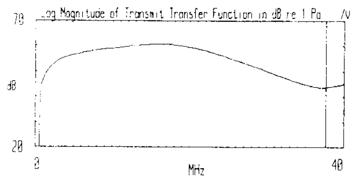

Fig. 5. Simulated transmitting voltage response of 10-mm-diameter 28-μm-thick PVDF polymer acoustic source in water at axial distance of 50 cm. Backing impedance is 40 MRayl.

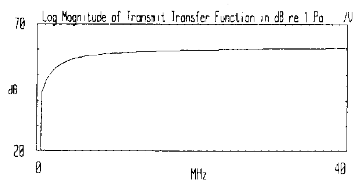

Fig. 6. Simulated transmitting voltage response for 10-mm-diameter 9-μm PVDF acoustic source with electrical source resistance of 50 Ω.

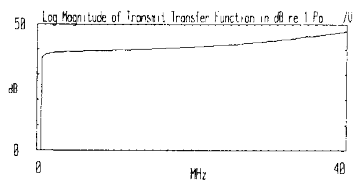

Fig. 7. Same as Fig. 4, except that electrical source resistance is set to 500 Ω.

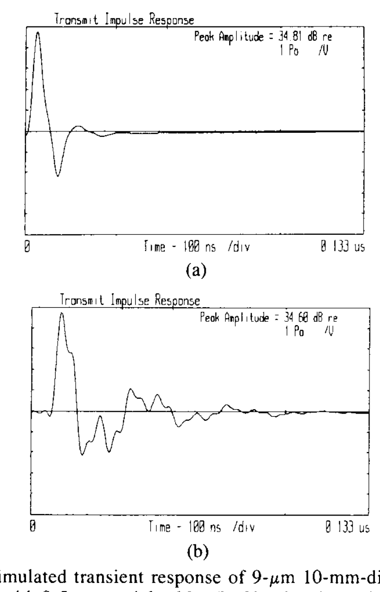

Fig. 8. (a) Simulated transient response of 9-μm 10-mm-diameter PVDF transmitter with 0.5-m coaxial cable. (b) Simulated transient response of 9-μm 10-mm-diameter PVDF transmitter with 2.0-m coaxial cable.

IV. Construction of the Wide-Band PVDF Transmitter

The construction of the PVDF polymer transmitter is shown schematically in Fig. 9. The piezoelectric element, 10 mm in diameter and made from 9-μm PVDF film (Kynar and Solef), was glued directly to the carefully polished backing material using a low-viscosity epoxy. This thickness of PVDF film should have a resonance frequency higher than 40 MHz (see Fig. 3), which was the upper cutoff frequency of the spectrum analyzer used here. The polymer material supplied was already poled and electroded. The electrode material was copper of 1000-A thickness. Electrical contacts were made to the rear of the element via a 0.2-mm diameter copper wire attached with conductive adhesive (Tra-duct®) and held in a thin insulating tube in the backing material. The other end of this wire was soldered to the inner lead of a standard BNC connector. Contact to the front face of the element was established via a similar copper wire attached to both the element and to the brass tube insert. This outer electrode was connected to the outer (ground) connection of the BNC connector. The stainless-steel tube was also grounded, thus providing effective electrical shielding. The hollow space behind the backing material was provided for ease of construction. To avoid quarter-wavelength filtering effects the front surface electrode was not covered by any protective coating.

V. The Calibration of the PVDF Transmitters

The calibration of the wide-band PVDF transmitters included determination of the transmitting voltage response at a given distance, measurement of the directivity patterns at different frequencies and calculation of the effective transmitting aperture.

A. Transmitting Voltage Response

As already mentioned, the transmitting voltage response defines acoustic output for a unity-voltage sinusoidal excitation signal at a given axial distance from the transmitter. This response was obtained by measuring the transmitted pressure with a hydrophone probe (see Section II). The experimental setup employed was similar to the one described in [26], based on the time delay spectrometry (TDS) technique. Briefly, the broad-band PVDF transducer was driven by the sine swept signal from the tracking generator of a spectrum analyzer (HP 3585A). The swept signal was amplified by a wide-band power amplifier (ENI). The signal detected by the hydrophone was fed into the spectrum analyzer input and displayed.

Since the transmitter's excitation (or driving voltage amplitude), the frequency response of the hydrophone probe, and the distance between the transmitter and the hydrophone were all known, the transmitting voltage response was determined as [32]

$$S_v = E_h d / M_t E_d \tag{1}$$

where E_h is the voltage measured at the terminals of the PVDF hydrophone probe terminated with the input impedance of the oscilloscope or spectrum analyzer (typically 1 MΩ in parallel with 30 pF), d is the axial distance between the hydrophone probe and the transmitter, M_t is

®Registered service mark of Tra-con, Medford, MA.

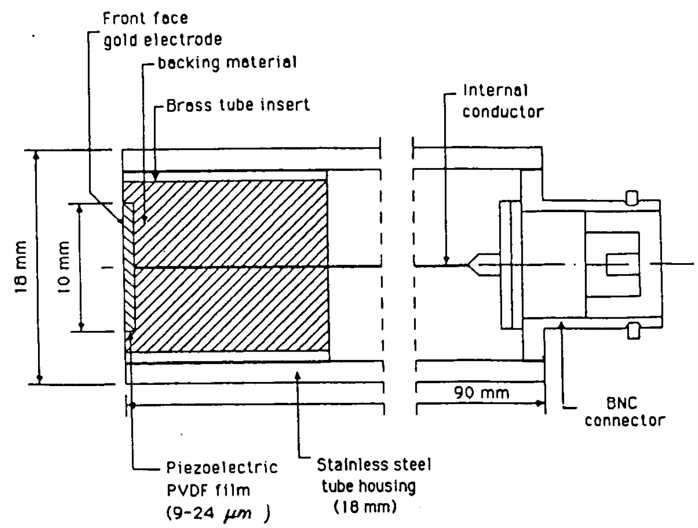

Fig. 9. Schematic construction of wide-band PVDF polymer acoustic source

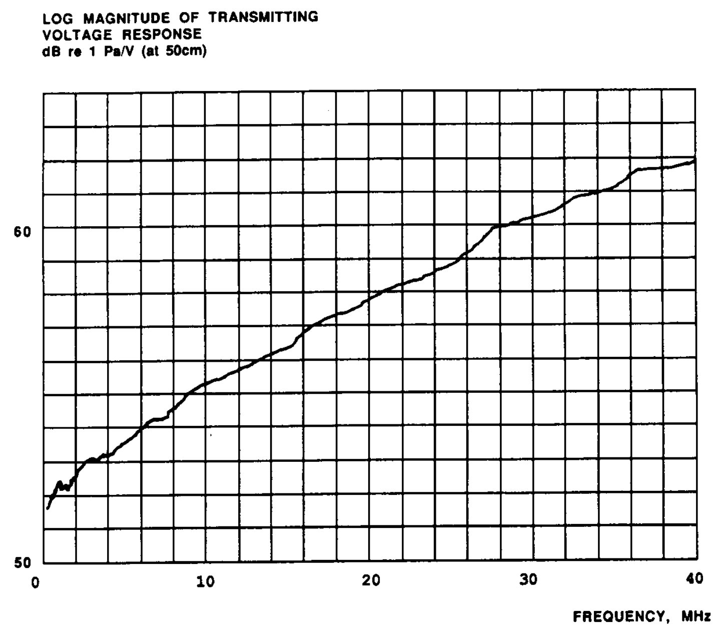

Fig. 10. Typical transmitting voltage response of 10-mm-diameter 9-μm PVDF polymer source measured in water at axial distance of 50 cm. Excitation voltage was supplied via 1-m coaxial cable. Source resistance: 150 Ω.

the end-of-cable voltage sensitivity of the hydrophone probe, and E_d is the amplitude of the transmitter driving voltage. A typical transmitting voltage response of the PVDF transducer measured in water at the axial distance of 50 cm is shown in Fig. 10 (note the expanded vertical scale in comparison with Fig. 3; also see Section VI for a more detailed discussion).

As an example of the wide-band behavior of the PVDF source, Fig. 11 shows the pressure–time waveform generated by the polymer source in water. The transmitter was driven with a single cycle burst at 7 MHz. Note the lack of ringing and very good fidelity to the driving signal.

To compare the PVDF material performance with that of conventional piezoelectric ceramics, a PZT transmitter was constructed. The transmitter was made of a 10-mm-diameter 22-MHz plane piezoelectric ceramic disk. The ceramic material was of PZT 24 type (trademark of Ferroperm A/S, Denmark) and was polarized in the thickness direction. The relatively flat frequency response was obtained by using a backing acoustic impedance of approximately 18 MRayl. Fig. 12 shows the transmit transfer

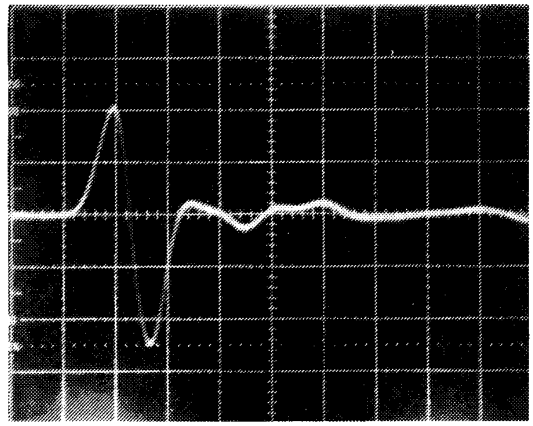

Fig. 11. Pressure-time waveform generated by polymer source of Fig. 10 in water and recorded by needle-type PVDF hydrophone probe. Source was excited by single-cycle 7-MHz burst. Horizontal: 50 ns/div. Vertical: 1 mV/div.

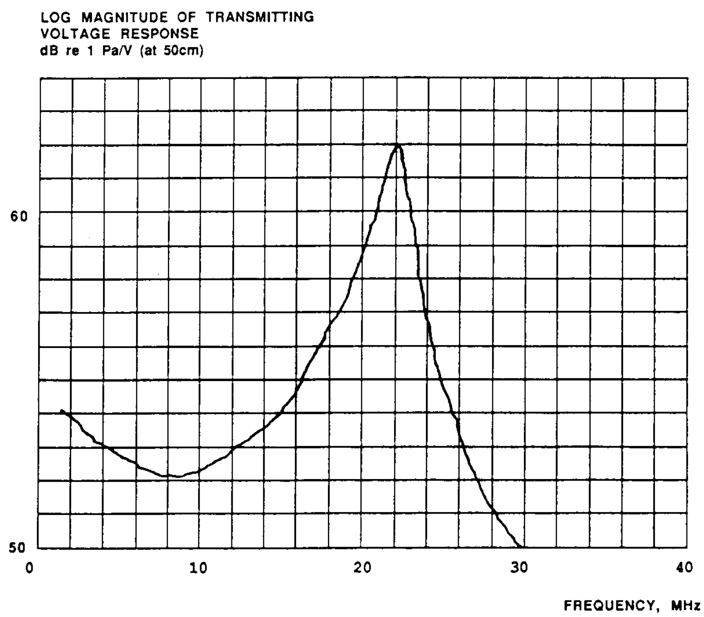

Fig. 12. Typical transmitting voltge of 10-mm-diameter 22-MHz plane piezoelectric ceramic source measured in water at axial distance of 50 cm. Excitation voltage was supplied via 1-m coaxial cable. Source resistance: 150 Ω.

function (transmitting voltage response) determined according to the calibration procedure described earlier.

B. Directivity Patterns

The directivity patterns of the PVDF polymer transmitters were obtained in the following way. At a particular frequency the distance of the last axial maximum was calculated, and then the hydrophone probe was placed at this point. The acoustical axis of the transmitter was then determined using the procedure described in [33]. This was performed by assessing the symmetry of the amplitude distribution in two perpendicular planes, each of which included the acoustic axis of the transmitter. Subsequently, the transmitter was rotated about an axis through its face while the voltage received at the terminals of the hydrophone probe was recorded. Some typical results taken in approximately 0.1° steps are shown in Fig. 13.

C. Effective Aperture Size

Generally, the effective aperture size (or radius for circular transducers) is a function of frequency and is almost never identical with the geometrical radius [34]. To determine the effective diameter, frequency spectra were measured for various angles of incidence. From these spectra the angle corresponding to the first zero of the

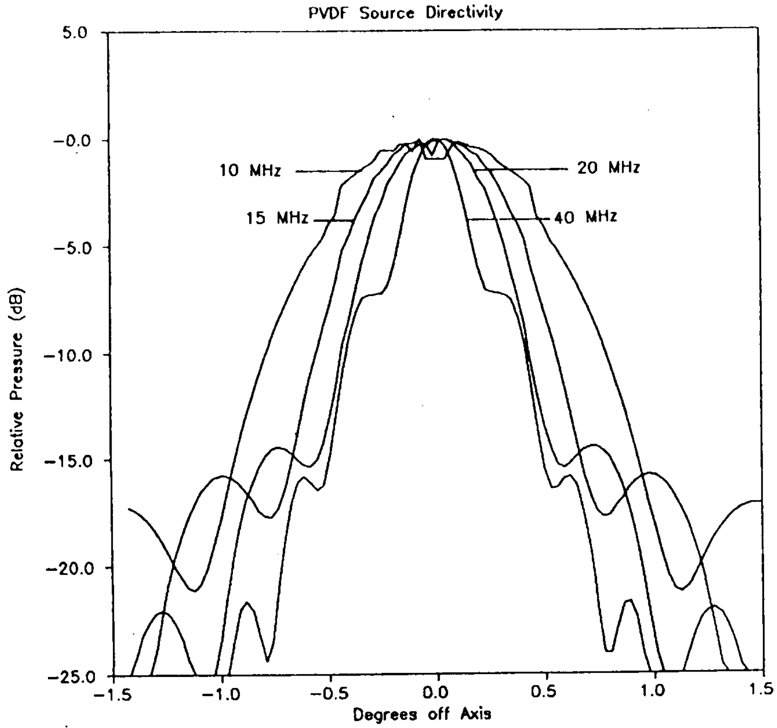

Fig. 13. Typical directivity patterns of wide-band PVDF acoustic source in water.

major lobe was found as a function of frequency [13]. This angle can be related to the effective diameter assuming a circular piston transducer [35].

The effective radius of the constructed PVDF transmitters was experimentally determined for several frequencies in the frequency range 1-20 MHz using the methodology described in [6] and [21]. The results of the measurements are summarized in Table I. All values listed in Table I were obtained by calculating average values of repetitive (minimum 3) measurements performed at the frequencies considered. Deviation from the nominal diameter was defined in the following way:

$$\text{percent deviation} = \{(\text{effective diameter-nominal diameter})/\text{nominal diameter}\} \cdot 100 \text{ percent}.$$

From the results listed in Table I, the effective diameter of the constructed transmitter matched its physical diameter to within 7 percent over the full frequency range considered.

TABLE I
EVALUATION OF THE EFFECTIVE DIAMETER OF THE WIDE-BAND PVDF TRANSMITTER WITH NOMINAL DIAMETER OF 9.8 mm[a]

Frequency	Effective Diameter (mm)	Deviation (Percent)
1	9.58	-2.24
3	9.53	-2.76
5	9.64	-1.63
8	9.69	-1.12
10	9.61	-1.94
12	9.48	-3.26
15	9.32	-4.90
20	9.19	-6.22
40	9.12	-6.96

[a]Deviation in percent is determined as (percent deviation) = {(effective diameter − nominal diameter)/nominal diameter} · 100 percent.

VI. DISCUSSION

The PVDF transmitters described here exhibit wide well-behaved frequency response. Generally good agreement exists between the slopes of the predicted (Fig. 3) and the measured (Fig. 10) transmitting voltage responses of the PVDF acoustic sources (0.21 and 0.23 dB/MHz, respectively). However, the absolute levels differ by approximately 6 dB at 1 MHz and 8 dB at 40 MHz, with the experimental values being higher. The discrepancies observed between the theoretical predictions and the measured response indicate that the electromechanical parameters of the constructed transducers were not identical with those used as input parameters to the KLM model. Data used for the modeling presented here were taken from the manufacturers' data sheets. However, the electrical, elastic, and piezoelectric properties depend on manufacturing and poling processes; the coupling coefficient and d and g constants may differ significantly between manufacturers and batches. In addition, since polymers are relatively soft materials, the thickness of the electrodes deposited on the surface may affect their electromechanical properties. Also, it is important to recognize that the electromechanical properties vary with frequency. In view of the above, the discrepancies found were not considered unreasonable. Since the primary goal of the simulation work was the corrected prediction of the overall frequency bandwidth and slope, the observed discrepancy between the predicted and measured absolute acoustic outputs was not examined further.

In comparison with conventional heavily damped piezoelectric ceramic transducers, PVDF transmitters exhibit lower transmitting efficiency in the lower megahertz frequency range (see Figs. 10 and 12). However, this lower transmitting efficiency is fully adequate when an inherently narrow-band measurement system, such as time-delay spectrometry is used [26]. The PVDF transmitters described here are routinely used for absolute calibration of hydrophone probes in the 1–15 MHz frequency range using the TDS principle combined with the reciprocity technique. Although the present measurements were only performed in the frequency range from 1 to 40 MHz, the theoretical predictions indicate that the PVDF transmitters are usable at frequencies beyond 40 MHz. This makes the transmitters particularly well suited for wide-band calibration procedures and broad-band acoustical measurements of materials based on the TDS principle.

The measured directivity patterns of the PVDF transmitters are in excellent agreement with those predicted using a uniform piston model (see Fig. 13) [20], [21], [35]. To minimize measurement errors due to the sharpness of the transmitted beam at higher frequencies, a precise positioning system was used during the calibration and measurement procedures. For the frequency range up to 40 MHz, the positioning system used here, which had a step interval of 10 μm, has proven to be satisfactory. The effective radius of the PVDF transmitters was found to be essentially constant (to within ± 7 percent) over the entire frequency range. This further facilitates calibration procedures since the knowledge of the effective radius is required to account for diffraction corrections [7]. Recent advances in the piezomaterials design and the advent of more sensitive copolymers materials, such as polyvinylidene fluoride trifluoroethylene copolymer P(VDF-TrFR) [25], hold promise that more efficient wide-band transmitters can soon be constructed.

Acknowledgment

The authors would like to express their gratitude to Dr. George K. Keilman for his valuable discussions on the KLM model.

References

[1] J. F. Greenleaf, Ed., *Ultrasound Tissue Characterization*. CRC Press, 1986.
[2] D. Shombert and G. R. Harris, "Use of miniature hydrophones to determine peak intensities typical of medical ultrasound devices," *IEEE Trans. Ultrason. Ferroelec. Freq. Contr.*, vol. UFFC-33, pp. 287–294, 1986.
[3] D. R. Bacon, "Finite amplitude distortion of the pulsed fields used in diagnostic ultrasound," *Ultrasound Med. Biol.*, vol. 10, pp. 189–195, 1984.
[4] H. C. Starrit, M. A. Perkins, F. A. Duck, and V. F. Humphrey, "Evidence for ultrasonic finite-amplitude distortion in muscle using medical equipment," *J. Acoust. Soc. Amer.*, vol. 77, pp. 302–306, 1985.
[5] T. G. Muir and E. L. Carstensen, "Prediction of nonlinear effects at biomedical frequencies and intensities," *Ultrasound Med. Biol.*, vol. 6, pp. 345–357, 1980.
[6] *The Characteristics and Calibration of Hydrophones for Operation in the Frequency Range 0.5 MHz to 15 MHz*, Int. Electrotech. Commission, Draft to TC29D-WG4.
[7] K. Brendel and G. Ludwig, "Calibration of ultrasonic standard probe transducers," *Acustica*, vol. 36, pp. 203–208, 1976.
[8] S. E. Fick, F. R. Breckenridge, C. E. Tscheigg, and D. E. Eitzen, "An ultrasonic absolute power transfer standard," *J. Res. NBS*, vol. 89, no. 2, pp. 209–212, 1984.
[9] W. B. Gloersen, G. R. Harris, H. F. Stewart, and P. A. Lewin, "A comparison of two calibration methods for ultrasonic hydrophone," *Ultrasound Med. Biol.*, vol. 8, pp. 545–548, 1982.
[10] P. C. Pedersen, T. J. Green, and P. A. Lewin, "Electroacoustic transducer characterization based on swept frequency measurements," in *Ultrasonic Int. '85 Proc.*, Guildford, UK: Butterworth Press, 1986, pp. 151–157.
[11] P. C. Pedersen *et al.*, "Application of Time-Delay Spectrometry for Calibration of Ultrasonic Transducers," this issue, pp. 185–205.
[12] P. A. Lewin, "Polymer hydrophones in biomedical ultrasonics," *Ferroelectrics*, vol. 60, pp. 127–139, 1984.
[13] P. A. Lewin, P. C. Pedersen, and M. E. Schafer, "A spectral analysis technique in ultrasound transducer characterization," in *IEEE Ultrasonics Symp. Proc.*, 1984, pp. 731–734.
[14] A. S. DeReggi, S. Edelman, S. C. Roth, H. Warner, and J. Wynn, "Piezoelectric polymer receiving arrays for ultrasonic applications," *J. Acoust. Soc. Amer.*, vol. 61, p. S1, 1977.
[15] G. R. Harris, "Sensitivity considerations for spot poled membrane," *IEEE Trans. Sonics Ultrason.*, vol. SU-29, pp. 370–377, 1982.
[16] R. C. Preston, D. R. Bacon, A. J. Livett, and K. Rajendran, "PVDF membrane hydrophone performance properties and their relevance to the measurement of the acoustic output of medical ultrasonic equipment," *J. Phys. E*, vol. 16, pp. 786–796, 1983.
[17] P. A. Lewin, "Miniature piezoelectric polymer ultrasonic hydrophone probes," *Ultrasonics*, vol. 19, pp. 213–216, 1981.
[18] P. A. Lewin and M. E. Schafer, "Ultrasonic probes in measurement practice," *Medical Devices Diagnost. Ind.*, pp. 40–45, May 1986.
[19] D. R. Bacon, "Characteristics of a PVDF membrane hydrophone for use in the range 1–100 MHz," *IEEE Trans. Sonics Ultrason.*, vol. SU-29, no. 1, pp. 18–25, 1982.
[20] D. G. Shombert, S. W. Smith, and G. R. Harris, "Angular response of miniature ultrasonic hydrophones," *Med. Phys.*, vol. 9, pp. 484–492, 1982.
[21] G. R. Harris, "Hydrophone measurements in diagnostic ultrasound fields," this issue, pp. 87–101.
[22] M. Platte, "A polyvinylidene fluoride needle hydrophone for ultrasonic applications," *Ultrasonics*, vol. 23, pp. 113–118, 1985.
[23] R. Smith, "Importance of frequency response of a hydrophone when characterized for medical ultrasound fields," in *Ultrasound in Medicine Conf. Proc.*, Inst. of Acoustics, UK, Apr. 1986, pp. 119–128.
[24] V. P. Jackson, E. Kelly-Fry, P. A. Rothchild, R. W. Holden, and S. A. Clark, "Automated breast sonography using a 7.5 MHz PVDF transducer: Preliminary clinical evaluation," *Radiology*, vol. 159, pp. 679–684, 1986.
[25] K. Kemura, N. Hashimoto, and H. Ohigashi, "Performance of a linear array transducer of Vinvylidene Fluoride Trifluoroethylene copolymer," *IEEE Trans. Sonics Ultrason.*, vol. SU-32, pp. 566–573, 1985.
[26] P. A. Lewin, "Calibration and performance evaluation of miniature ultrasonic hydrophones using time delay spectrometry," in *Proc. 1981 IEEE Ultrasonics Symp.*, 1982, pp. 660–664.
[27] R. C. Chivers, "Time-delay spectrometry for ultrasonic transducer characterization," *J. Phys. E.*, vol. 19, pp. 834–843, 1986.
[28] R. Krimholtz, D. A. Leedom, and G. L. Mattaei, "New equivalent circuits for elementary piezoelectric transducers," *Electron. Lett.*, vol. 6, pp. 398–399, 1970.
[29] G. W. Keilman, "Active broadband electrical impedance matching of piezoelectric transducers," master's thesis, Univ. of Washington, Seattle, 1981.
[30] A. R. Selfridge and S. Gelbach, "KLM transducer model implemen-

tation using transfer matrices," in *Proc. IEEE Ultrasonics Symp.*, 1985, pp. 875–877.
[31] W. P. Mason, *Electromechanical Transducers and Wave Filters*, 2nd Ed. Princeton, NJ: Van Nostrand, 1948, pp. 399–404.
[32] R. J. Bobber, *Electroacoustic Underwater Measurements.* Washington, DC: GPO, 1970, p. 25.
[33] P. A. Lewin and R. C. Chivers, "Two miniature ceramic ultrasonic probes," *J. Phys. E*, vol. 14, pp. 1420–1424, 1981.
[34] R. C. Chivers, L. Bosselar, and P. A. Filmore, "Effective area to be used in diffraction corrections," *J. Acoust. Soc. Amer.*, vol. 68, pp. 80–84, 1980.
[35] L. E. Kinsler, A. R. Fry, A. B. Coppens, and J. V. Sanders, *Fundamentals of Acoustics*, 3rd Ed. New York: Wiley, 1982, p. 107.

Peter A. Lewin (SM'85), for a photograph and biography please see page 86 of this issue.

Mark E. Schafer (S'83–M'83–S'85–M'86), for a photograph and biography please see page 109 of this issue.