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● *Original Contribution*

NEW PIEZOELECTRIC TRANSDUCERS FOR THERAPEUTIC ULTRASOUND

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Abstract—Therapeutic ultrasound (US) has been of increasing interest during the past few years. However, the development of this technique depends on the availability of high-performance transducers. These transducers have to be optimised for focusing and steering high-power ultrasonic energy within the target volume. Recently developed high-power 1-3 piezocomposite materials bring to therapeutic US the exceptional electroacoustical properties of piezocomposite technology: these are high efficiency, large bandwidth, predictable beam pattern, more flexibility in terms of shaping and definition of sampling in annular arrays, linear arrays or matrix arrays. The construction and evaluation of several prototypes illustrates the benefit of this new approach that opens the way to further progress in therapeutic US. **© 2000 World Federation for Ultrasound in Medicine & Biology.**

Key Words: **Therapeutic ultrasound, Transducers, Piezocomposite, High-intensity ultrasound, Cavitation, Shock waves.**

INTRODUCTION

Ultrasound (US) therapy has been the subject of research for many years (Fry et al. 1954; Sanghvi et al. 1996; ter Haar 1995), but it is only recently that this technique has found effective and widespread medical applications (Madersbacher et al. 1994; Gelet et al. 1996). The potential of this technique is extremely promising, but there remains progress to be made, notably in the area of the generation of ultrasonic waves. The performance of the transducer plays an important role in the US generation and it is, therefore, essential to have at one's disposal reliable transducers that are specifically adapted to each application. Each application of US therapy is expressed by a specific objective of action on the biological environment, generally in the form of modification or destruction of certain tissues or matter: burning, coagulation, fragmentation, cellular destruction, production of free radicals, medicinal release and activation, etc. The needs of each ultrasonic therapy application can, therefore, be expressed by a set of functions and constraints that the transducer must satisfy:

- The electroacoustical conversion function of the transducer must be adaptable to the modes of excitation used for the application: transmission of waves maintained for long periods of time, excitation by brief wave sequences, modulation of the frequency or modulation of the amplitude during the wave sequence, excitation by impulse, etc. The choice of an excitation mode is influenced by the effect desired for the environment and the tissues. At the level of the transducer, the required properties are expressed most notably by the need of good conversion efficiency, large bandwidth in the frequency domain and short impulse response in the time domain. The technological limitations of the transducer are limitation of the acoustic power per radiating surface area, limitation of the electrical voltage applied to the transducer, limitation of the acoustic pressure radiated at the surface of the transducer, etc.
- The focusing function plays a particularly important role in ultrasonic therapy (Rivens et al. 1996). One of

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the major advantages of focusing is to offer a means of treatment that is deep-reaching and localised, with few secondary effects for the healthy tissues. This objective requires the need for highly focused transducers having a focal distance-active diameter relation (f-number) on the order of 1 or smaller than 1. It is beneficial if the focusing effects of the beam can be controlled and the production of artefacts is avoided, such as lobes of radiation produced by parasitic vibrations of the transducer. The mastery of focusing therefore requires an understanding of the vibration conditions of the elements of the transducer. The objective of focusing must also take into account the constraints on acoustic access to the target area, which sometimes requires a geometric limit on the radiating surface. This geometrical constraint can take a fixed form (*e.g.*, the form of a truncation applied to a spherical dome surface) or take a variable form (*e.g.*, by electronic commutation of the elements of an array), to bypass the screening effect of an environment opaque to US. The focusing function is, therefore, expressed by numerous requirements on the geometry of radiating surfaces. Transducers for external use generally need a radiating surface of a large dimension and it is, therefore, necessary to limit weight and to enhance resistance to mechanical and thermal shocks. The possibilities offered by electronic focusing, thanks to the use of annular, linear or matrix array transducers, respond to many different needs of focusing adaptation.
- Treating a certain volume with an appropriate precision and reliability requires the use of beam scanning; this can be done mechanically or electronically, using an array transducer driven by an appropriate electronic system.

Beyond the basic ultrasonic functions, the need to take into account a certain number of other constraints plays an essential role in the functional design of scanners. These include electrical safety for the patient and for the user of the product, the reliability of the control of electroacoustical and thermal parameters of the transducer, the general longevity of the transducer and, notably, its resistance to repeated use at high acoustic power levels, its resistance to corrosion from contact with water, the biocompatibility of the materials and the electromagnetic compatibility with the other parts of the system.

1-3 PIEZOCOMPOSITE TECHNOLOGY AND ITS CONTRIBUTION TO US THERAPY

Piezocomposite materials with 1-3 connectivity are made up of fine small columns of piezoelectric ceramic embedded in a polymer material (Smith and Auld 1991). This structure enhances the electromechanical performance in the direction of the thickness of the plate. The physical properties of piezocomposites, notably their dielectric, elastic and piezoelectric characteristics, depend not only on the properties of the component materials, but also the relative proportions of these materials and on the geometric parameters of the microstructure. The principal characteristics of piezocomposite materials may be modelled (Hossack and Hayward 1991; Cao et al. 1992). The variation of dielectric permitivity, acoustic impedance, longitudinal propagation velocity and coupling coefficient, as a function of the volume fraction of the ceramic material within the composite, have been studied by Smith and Auld (1991).

Let us, nonetheless, take note that, for a 1-3 piezocomposite, the coupling coefficient obtained is superior to that of the ceramic material; hence, resulting in a better conversion of energy.

It is known that the mechanical losses of composite 1-3 materials are larger than those of the bulk ceramics generally used in this type of application. Thus, the electroacoustical conversion efficiency of a transducer produced by piezocomposite technology is *a priori* and, notably in the simplest cases, lower than that of a transducer produced by ceramic technology. However, in many applications, the efficiency of the transducer is not the only objective to be satisfied and piezocomposite technology permits the realisation of these other objectives. The particular properties of 1-3 piezocomposite materials are:
- The lower acoustic impedance of piezocomposites, on the order of 8 to 12 MRayl, facilitates the transfer of energy to water with a wide bandwidth;
- the thickness mode coupling coefficient k_T, on the order of 0.55 to 0.65, facilitates the energy conversion and the widening of the bandwidth (Smith and Auld 1991);
- the addition of a front layer is possible if this layer is conceived in such a way that its thermal dilatation coefficient is close to that of the piezocomposite material. This layer can fulfil different functions, notably acoustic adaptation and matching, electric insulation and mechanical and chemical resistance of the radiating surface of the transducer.
- a judicious selection of the thermomechanical properties of the polymers allows for conditioning of the active surface and facilitates the production of transducers of complex form;
- the very strong anisotropy of piezocomposite materials allows for reduction of the propagation of vibrational modes other than the thickness mode. This property facilitates the design of array transducers because, in numerous cases, simple patterning on the electrode at the surface of the same substrate allows the elements of the array to be acoustically independent. This prop-

erty is also beneficial for avoiding the generation of parasitic radiation lobes.

Thus, piezocomposite technology offers a set of solutions to the designer of transducers for US therapy applications. In the following paragraphs, we review theoretical and experimental results to illustrate examples. The development of transducers for ultrasonic therapy applications using piezocomposite technology has required much research on polymer materials in terms of the stability of their mechanical characteristics over a wide range of temperatures. It has also required the design of multilayered structures (front layer/piezocomposite/backing layer) to facilitate acoustic energy transfer and thermal draining. The results of this work will be the subject of future communications.

APPLICATIONS OF 1-3 PIEZOCOMPOSITE TECHNOLOGY

Recent work performed in different research groups allows us to illustrate the potential of piezocomposite technology in the field of ultrasonic therapy:

Generation of high-intensity ultrasonic waves

Recent progress achieved in the design of high-power transducers permits piezocomposite transducers to deliver acoustic power levels compatible with therapy applications (mean intensity up to 10 to 15 W/cm^2 at the front face of the transducer). To achieve this, transducers must be able to withstand the applied electrical power in a continuous manner for several seconds, or even several minutes. Under such conditions, the internal heating of the transducer due to electrical or mechanical losses constitutes a first source of limitations. Excessive overheating may cause an irreversible loss of performance and the depolarisation of piezoelectric material. To reduce this effect, it is necessary to minimise the losses in the transducer, by either optimising the efficiency of electrical energy conversion into acoustic energy at the excitation frequency or by draining the thermal energy to an exterior environment. Recent tests performed at the University of Michigan (Kluiwstra et al. 1996) on piezocomposite technology transducers manufactured by IMASONIC (Besancon, France) showed efficiencies of around 60% for transmitted acoustic intensity on the order of 10 W/cm^2. These efficiency and acoustic power levels are retained for the duration of the excitation period (60 s) and are reproducible. The acoustic power varies linearly with the electrical input power as is shown, for instance, in Fig.1, with the 64-element array described in Fig.2. Nonetheless, the efficiency, as the excitation voltage is varied, is valid only under a critical power level: a maximum intensity of 15 to 20 W/cm^2 can only be obtained with efficiencies of 40% to 50%. These power levels are quite sufficient for most therapeutic applications.

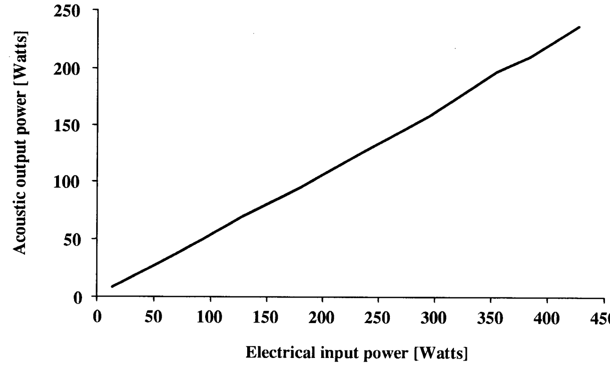


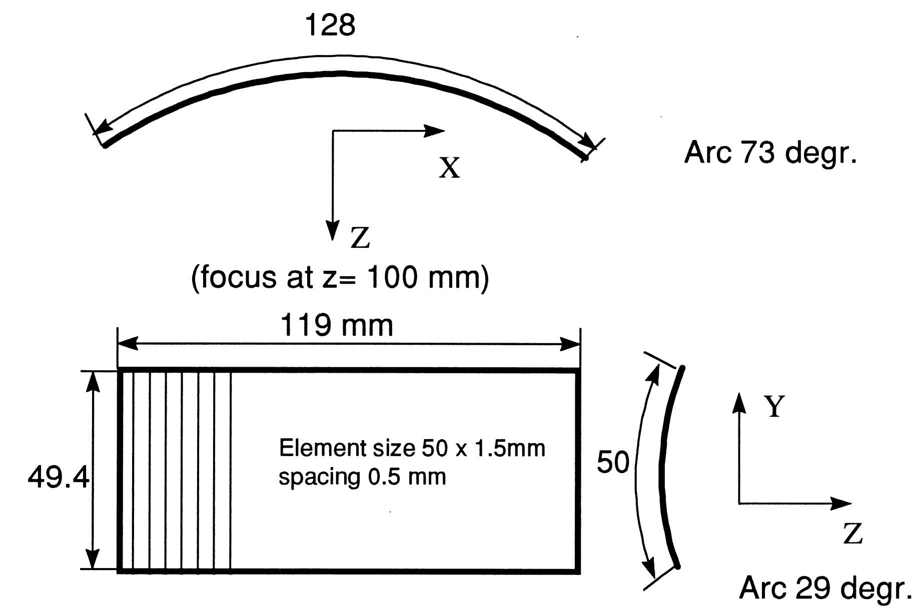
Fig. 1. Graph of the acoustical output power as a function of the electrical input power obtained with the piezocomposite array of Fig. 2 (full array power measurement).

Predictability of radiation pattern

The usual models used to predict the field radiated by the active surface of a transducer make the hypothesis that the latter vibrates in a homogeneous fashion (*i.e.*, the amplitude of displacement is constant at any point of the emitting surface).

Cathignol et al. (1997) have shown that this hypothesis is not valid in the case of a transducer composed of a ceramic spherical shell of PZT material, without matching layer, with air backing, or having a low impedance backing medium. The axial distribution of the pressure amplitude differs noticeably from the modelled results for distances lower than the focusing distance. A theoretical and experimental analysis shows that the vibrating behaviour in thickness of such a transducer is disturbed by the existence of plate waves propagating from the edges of the ceramic layer. Comparative measurements (Cathignol et al. 1995b, 1999) have been taken between such a ceramic transducer and a piezocomposite transducer of the same general makeup, identical dimensions and active geometry. Figure 3 shows that the axial evolution of the pressure differs greatly from the prediction of Rayleigh integral (O'Neil 1949) in the case of the transducer composed of a spherical shell of ceramic material. In contrast, it is very close to the theoretical curve in the case of a piezocomposite transducer. This difference is explained by the fact that the piezocomposite structure does not allow the propagation of plate waves (Fig. 4). It is desirable here to note that the form of the ceramic transducer tested by Cathignol et al. (1999) is typical of many transducers dedicated to ultrasonic therapy.

Kluiwstra et al. (1996) have also observed signifi-



Transducer characteristics	
Center frequency	1.0 MHz
F-number	0.84
-3 dB bandwidth	25 %
-3 dB focal dimension	1 x 2.2 x 12 mm^3
Max. acoustic intensity (surface)	10 W/cm^2
Max. acoustic intensity (focus)	5400 W/cm^2

Fig. 2. 64-element piezocomposite linear array.

cant improvement offered by the piezocomposite approach to eliminate parasitic vibration modes.

Chapelon et al. (1993) demonstrated the predictability of the radiation emitted by a truncated annular array-type piezocomposite transducer where the acoustic field on the axis was measured for several electronic focusing configurations and compared to a theoretical simulation (Fig. 5). In spite of the probable effect of several factors that were not taken into account in the simulation (nonlinear effects of the propagation, directivity of the hydrophone, etc), the good predictability of the radiation is evident.

These examples illustrate the contribution of piezocomposite technology toward improving the radiated acoustic beam. This predictability of the radiation pattern is essential in ultrasonic therapy applications. The development of transducers with complex geometry makes this predictability even more necessary.

Generation of power in a wide frequency range

The significance of the generation of high-intensity US beam with wide frequency bandwidth has been recently illustrated by two works. A research study by Dupenloup et al. (1994, 1996) explored the effect of widening the frequency bandwidth to reduce grating lobes generated by an annular array. This approach creates a sufficiently weak lobe level that a small number of rings may be used in the array and, therefore, an electronic driving system may be used that is neither complex nor costly. The authors have shown that a bandwidth transmitted at around 50%, obtained by the use of piezocomposite transducers, allows for the increase of effective dynamic focusing range of the annular arrays defined in Table 1. A wide-band transmitted signal, in comparison with a single-frequency signal, enables an increase in the dynamic focusing range by nearly 50% for the T1 transducer and 30% for the T2 transducer. The main condition of using such a concept is the possibility of having a radiated acoustic energy level relatively constant in a given frequency range.

A second research study explored an effect of the widening of the frequency bandwidth on acoustic cavitation. The formation of air bubbles by cavitation causes degradation in the efficiency of a treatment and can bring

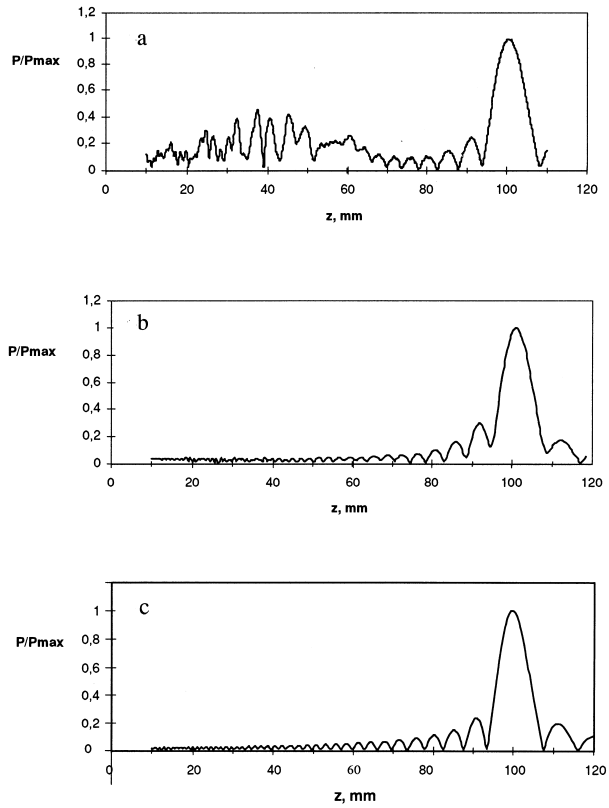


Fig. 3. Axial evolution of the pressure with ceramic and piezocomposite materials. Frequency: 1.6 MHz; diameter: 100 mm; radius of curvature: 100 mm. (a) Ceramic transducer with air backing; (b) piezocomposite transducer with backing; (c) theory.

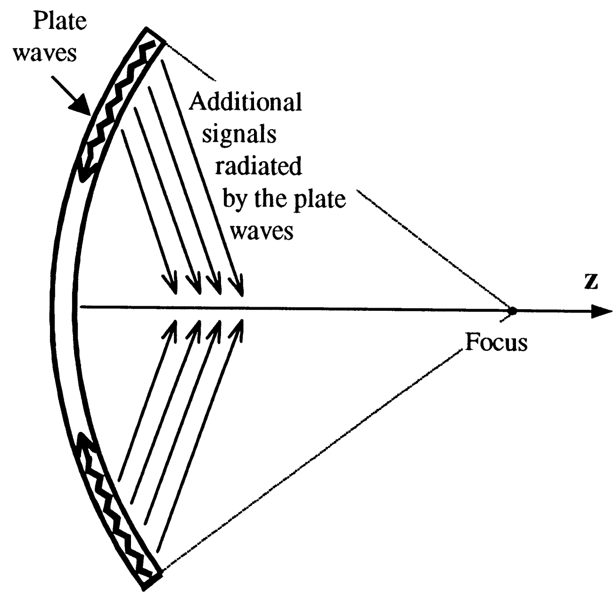


Fig. 4. Drawing showing the generation of plate waves.

about undesired lesions outside of the zone to be treated. To reduce cavitation, the only immediate solution is the limitation of the acoustic intensity produced at the focal point, and this implies an increase of the duration of the treatment. The approach proposed by Chapelon et al. (1996a) consists of using a signal phase modulated by a pseudorandom code instead of a single-frequency emission. This is, once again, made possible through the use of a wide bandwidth transducer (55% of the central frequency to -6 dB), taking advantage of piezocomposite technology. The results are impressive because cavitation was successfully reduced by a factor of 50 for a mean acoustic intensity of 4.6 W/cm^2. The intensity limitation due to the cavitation threshold is in this way relaxed and high-intensity focused US treatments may be performed with more precision in aim and better protection of surrounding tissues.

Flexibility of geometric design

The shaping of composite materials allows the realisation of complex geometric structures and, therefore, permits a large adaptability to the constraints of form. VanBaren et al. (1996) have used a linear array composed of 64 elements with a strong geometric focusing following the two directions perpendicular to the propagation axis of the beam. The total active area of the array measured along the curved surface is 128 mm by 50 mm (Fig. 2). The resonance frequency of this transducer is 1.0 MHz. Different studies showed the capacity of the array to generate peak acoustic intensity up to 5400 W/cm^2 at the focal point.

Piezocomposite technology enables the production of miniaturised therapy transducers for endoscopic applications. Transducers used by Chapelon et al. (1996b) include, for instance, an annular array with a frequency of 3.0 MHz, an active diameter of 50 mm, a truncated width of 18 mm and a focal distance of 40 mm. The f-number is 0.8 in one direction. This transducer includes a tissue thickness control function for appropriate damping of its electroacoustical response. This transducer and another one with a 36×11 mm^2 active area have been applied in ablating tumours of the pancreas or metastases of the liver. The different studies performed on animals have shown (Prat et al. 1997) that small dimension transducers of piezocomposite material are capable of generating *in situ* intensity higher than 900 W/cm^2 at the focus, with a 4-s pulse duration repeated every 5 s. The technology used satisfied the electric insulation requirements inherent to intracavity applications.

Generation of shock waves

The piezoelectric generation of shock waves is currently used for the destruction of kidney stones. This

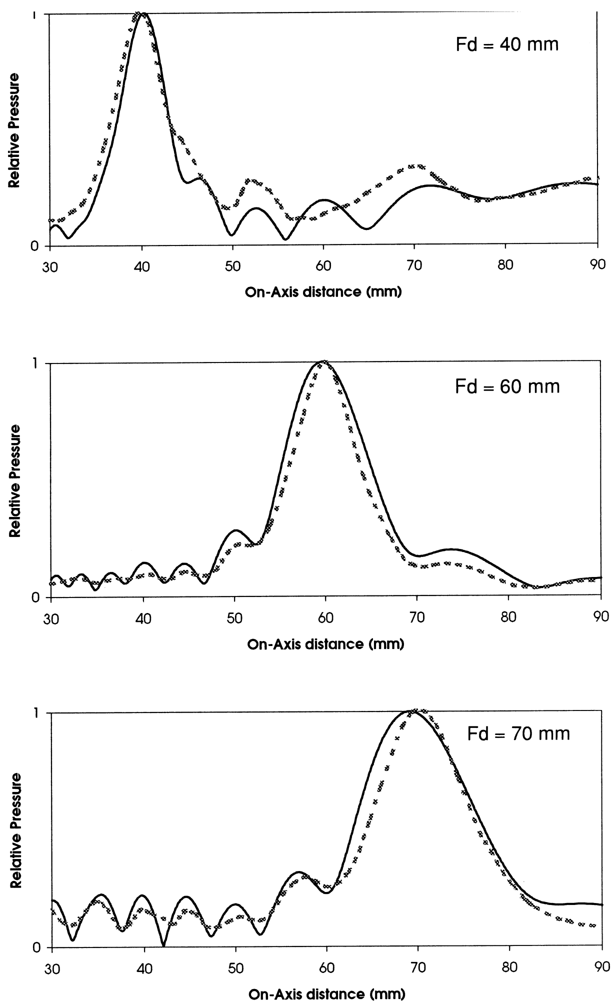


Fig. 5. Simulated (—) and measured (· · · · · ·) on axis pressure field of an 8-ring annular array, driving the focus electronically at 40, 60 and 70 mm, respectively. Frequency: 2.25 MHz; diameter: 50 mm truncated at 31 mm; geometric focal length: 60 mm.

technique allows for an excellent optimisation of the spatial distribution of the energy, as well as reproducibility of the wavefront in the time domain. The use of piezocomposite technology in the production of shock waves opens the way to new developments. Several prototype bidimensional arrays shaped with a 190-mm

Table 1. Characteristics of the transducers used by Dupenloup et al. (1996)

Transducer	Frequency (MHz)	Bandwidth (MHz)	Diameter (mm)	Focal length (mm)	Number of rings
T1	2.25	1.2	35	35	6
T2	1.10	0.5	150	150	8

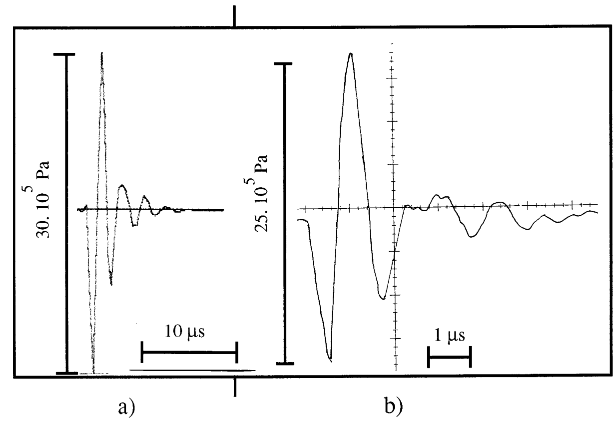


Fig. 6. Comparison of the (a) theoretical and (b) experimental pressure time waveforms at the front face of one element of the array used by Cathignol et al. (1995a).

radius of curvature have been produced. The elements are arranged in the form of concentric rings around a central element of diameter 8.3 mm. The rings are cut in sectors in such a way that all of the sectors have a surface that is constant and equal to that of the central element. Two prototypes of 121 elements have been used (Cathignol et al. 1995a; Thomas et al. 1994) to test different principles of electronic focusing under conditions of shock waves. These prototypes have demonstrated excellent aptitudes for the electronic focusing of shock waves. Figure 6 shows the excellent predictability of the waveform transmitted by one element of the array. Peak-to–peak pressure reaches 3.10^6 Pa with an applied voltage of 5500 V and the performance is stable after 10^6 shocks.

CONCLUSIONS

The evolution of ultrasonic therapy and the growing interest that it elicits, makes it increasingly necessary to take into account the specific nature of each application. The applications of piezocomposite technology that have been discussed show the significance of this technology. This is particularly evident if we consider the entirety of the needs to be satisfied in the design of a transducer for therapy applications. The specific properties of composite materials and the combination of these properties bring to the design of transducers a flexibility and a degree of optimisation that are often lacking in the case of designs based on ceramic materials. This flexibility and this degree of optimisation are the foundation of new solutions; hence, responding to the diversity of the needs of therapeutic US.

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