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Methods for Characterization of Physiotherapy Ultrasonic Transducers

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1. Introduction

Ultrasound (US) is an energy composed of cyclic acoustic pressures with a frequency higher than that of the upper limit of human hearing. This energy is an option to treat many diseases, from healing muscular inflammation to ablating malignant tumors. Ultrasound is an emission coming from a transducer which is chosen depending on the application. There are two main therapeutic applications of the ultrasound in medicine: low intensity ultrasound which uses unfocused transducers with acoustic intensities lower than 3 W/cm^2; and HIFU (High Intensity Focused Ultrasound) which uses focused transducers with acoustic intensities higher than 100 W/cm^2. Each application makes use of different kinds of transducers hence some standards have been established in order to characterize the equipment in accordance with the specific use. For example, in order to characterize a physiotherapy transducer (low intensity ultrasound), it is needed to determine and to validate the Effective Radiating Area (*ERA*), the Beam Non-uniformity Ratio (*BNR*) and the ultrasonic power (related to the effective acoustic intensity). The International Electrotechnical Commission (IEC) and the United States Food and Drug Administration (FDA) have established the methodology to measure all of these parameters.

A comparison of three techniques for characterization a physiotherapy ultrasonic transducer by the determination of two of the mentioned parameters, *ERA* and *BNR,* is presented in this chapter. The ultrasonic power can be measured by using a radiation force balance —a simple and accurate method that is not mentioned here because of the objective of this chapter. The techniques are based on measurements of the acoustic field which are postprocessed in order to get the characteristic parameters of the ultrasonic transducer. This chapter also includes a brief abstract of other techniques that have been used for the same objective. These techniques were not included in the comparison because of their expensiveness and the technological requirements to be implemented. The use of each technique described here depends on the necessities of the application.

2. Ultrasound in Medicine

Ultrasound has been used in medicine for many years. A wide variety of applications have been developed in order to help in diagnosis or even to treat some diseases, and all of them differ in the frequency, the kind of transducer (and therefore the kind of beam), and the acoustic intensities, among other factors. Some medical ultrasound applications that can be mentioned here are the ultrasonic imaging, the flow measurements (Doppler and Transit Time), tissue healing, bone regeneration and cancer therapy (Paliwal & Mitragotri, 2008; ter Haar, 1999; ter Haar, 2007). In this chapter, we will talk about the techniques for characterizing ultrasonic transducers used in the treatment of muscular injuries; however, general information about ultrasound in therapy is needed in order to better understand the specific necessities.

2.1 Ultrasound in Therapy

Therapeutic ultrasound is the use of ultrasonic energy in order to produce changes in tissues through its mechanical, chemical and thermal effects. Depending on the effects in the tissues and the area of application, the ultrasound therapy can have different names. In general, therapeutic ultrasound can be separated in two categories: "low" intensity ultrasound (0.125-3 W/cm^2) and "high" intensity ultrasound (more than 5 W/cm^2) (ter Haar, 1999; ter Haar, 2007). The lower intensities are used when the treatment is expected to propitiate the regeneration of tissues caused by physiological changes. In contrast, higher intensities are used when ultrasound has to produce a complete change in tissue by means of overheating (hyperthermia) or cell killing (ablation) (Feril & Kondo, 2004).

There are two main areas where the last classification is clear: physiotherapy (low intensities) and oncology (high intensities). The therapy in both areas has been called therapeutic ultrasound, but the techniques have significant differences both in the devices as in the results. In general, the effects produced by ultrasound in tissues can be divided in two types:

- *Thermal effects*, which are produced basically because of the absorption of the energy by large protein molecules commonly present in collagenous tissues. Some of these effects are the increase in blood flow, the increase in tissue extensibility, the reduction of joint stiffness, pain release, etc. (Speed, 2001).
- *Nonthermal effects*, which are produced when the therapy is delivered in a pulsed way avoiding the media heating. The first nonthermal effect reported is the "micro-massage" (ter Haar, 1999) whose effects have not been measured yet. Acoustic streaming is another effect that could have important changes in the tissues. Streaming may modify the environment and organelle distribution inside the cells, this in turn can change the concentration gradients near the membrane and therefore it modifies the diffusion of ions and molecules across it (Johns, 2002). These effects are responsible for the stimulation of the fibroblast activity, the tissue regeneration and the bone healing (Johns, 2002; Speed, 2001).

2.1.1 Oncology

Oncologic ultrasound is the therapy that uses thermal effects of ultrasound in order to ablate malignant tumors. This therapy is applied either alone or in combination with radiotherapy or chemotherapy because it has been demonstrated that the effects of these therapies are potentialized when the tumor is ultrasonically heated (Field & Bleehen, 1979). The treatment consists in heating the tumor at temperatures above 42°C, which is the maximum temperature resistance of malignant cells, but avoiding overheating healthy cells around it. Heating is applied approximately for 60 min; during this time, the temperature in the tumor must be between 42-45°C and the temperature of the healthy tissue must be lower than 41.8°C (ter Haar & Hand, 1981).

The main problem of this therapy is the accurate control of temperature in tissues because there are no appropriate methods to measure the temperature in a continuous way and in all the heated volume without damaging tissues. This is the principal reason of why this therapy has not been widely used. Methods that use ultrasound to measure the temperature non-invasively inside a tissue are being developed, but problems with the non-homogeneity of tissues and natural scatters have not been eliminated (Arthur et al., 2003; Arthur et al., 2005; Maass-Moreno & Damianou, 1996; Pernot et al., 2004; Singh et al., 1990). Thermometry by using X-rays, or MRI is another option, but these techniques are expensive (De Poorter et al., 1995; Fallone et al., 1982). It has been proposed that this problem could be avoided by heating the tissues at higher temperatures (about 60°C) so fast that the normal perfusion does not have a significant effect (ter Haar, 1999).

2.1.2 Physiotherapy

The use of ultrasound to treat muscular damage, heal bones, reduce pain, etc. has been called physiotherapy ultrasound. This therapy uses ultrasound in order to induce changes in muscular and skeletal tissues through thermal and mechanical effects. These effects can be changes in the cell permeability (Hensley & Muthuswamy, 2002) or even cellular death when the ultrasonic energy is not controlled correctly (Feril & Kondo, 2004). The desired effect is a light elevation of the temperature into the treated tissue without provoking ablation (cell killing); this phenomenon is called diathermy. This therapy is commonly confused with hyperthermia, but the main difference is that the latter is an elevation of temperature with the objective of producing changes in tissues immediately by means of the overheating. In contrast, diathermy is the phenomenon of heating a tissue in order to induce physiological changes, e.g., an increase of the blood flow rate, activation of the immunological system, changes in the cell chemical interchange among cells and the extracellular media, etc.

The therapy consists in using a transducer to produce ultrasonic waves which are directed to the treated tissue. The transducer is connected to an RF generator that produces a senoidal signal (or approximately senoidal) which has high amplitude and high frequency. The transducer acoustic impedance is relatively small compared to the acoustic impedance of the air. Should the ultrasonic energy travel from the transducer to the air, only a little part would go out and the most significant part would go backwards. This reflected energy, called reflected wave, could damage the transducer and even the RF generator. During the therapy, when the transducer is dry, there is a thin layer of air between the transducer face and the skin; this layer can produce reflected waves. Therefore, in order to avoid this problem, media with acoustic impedances between the transducer and the skin are used to improve the contact between them. The ultrasonic waves are directed to the tissues by means of either using acoustic gel between the transducer and the skin or submerging the desired part of the body in degasified water and applying the energy with the transducer submerged too. Both ways are efficient in getting a correct coupling.

3. Physiotherapy Ultrasonic Transducers

Ultrasonic transducer technology has been improved in the last 50 years. The first transducers were constructed using piezoelectric crystals as ultrasound generator elements (Christensen, 1988). Later on, piezoelectric ceramics (polarized artificially in order to produce the piezoelectric effect) were discovered and developed which allowed designers construct different configurations with many shapes, sizes, frequencies, and at higher efficiencies. New design techniques, and new materials with better properties than their predecessors have contributed to improve the piezoelectric elements (Papadakis, 1999).

The construction of a US transducer is carried out in accordance with its application. The kind of material chosen for the piezoelectric element depends on the acoustic intensity at which the device will be used. However, there is another important parameter to consider: the bandwidth. Some transducers are designed to work in a range of frequencies that allow them to keep a good amplitude either receiving US (like the hydrophones) or both emitting and receiving. Others are good just for emitting ultrasound at a specific frequency. This new consideration allows for another way of transducer classification: wideband and narrowband transducers.

Physiotherapy ultrasonic devices use narrowband transducers because they require high efficiency in the energy conversion. This kind of transducers must work in the resonance frequency to make use of their high efficiency characteristics. When continuous emission occurs through a low efficiency transducer, a great part of the energy is transformed into heat in the transducer and only a little part of the energy is emitted to the media as ultrasound. This fact is not important in some applications, but in a physiotherapeutic treatment, the transducer is in contact with the patient's skin and overheating is an undesired effect. Characterization is an excellent tool to know if a transducer is working properly at nominal values. The incorrect transducer characterization could lead to the lack of results of the treatment or even provoke some injuries to the patient. Some defects in the emission efficiency could be due to a decoupling between the generator and the transducer, so that frequency characterization should be carried out in order to know this efficiency. In this chapter, only the acoustic characterization of a physiotherapy ultrasonic transducer working at its resonant frequency of 1 MHz is shown.

3.1 Transducer Acoustic Field

When a source of ultrasound emits energy, the ultrasonic waves produced are propagated around all directions of the source. The distribution of this mechanical energy is called acoustic field. The shape of the acoustic field has a distribution of acoustic pressures in accordance with the shape of the emitter. In physiotherapy transducers, the acoustic field shape is, theoretically, cylindrical because of the proportions of the piezoelectric element, i.e., the diameter is more than ten times the wave length (Águila, 1994). The first part of the transducer acoustic field (when the last condition is true) is called near field or Fresnel zone, and the next part is called far field or Fraunhofer zone. The Fresnel zone is composed of symmetrical rings of maximum and minimum pressures along the central edge which cause a non uniformed distribution of the acoustical energy. The Fraunhofer zone is divergent and the acoustic intensity follows the inverse-square law (Seegenschmiedt, 1995):

$$I_x \approx \frac{1}{x_2} \tag{1}$$

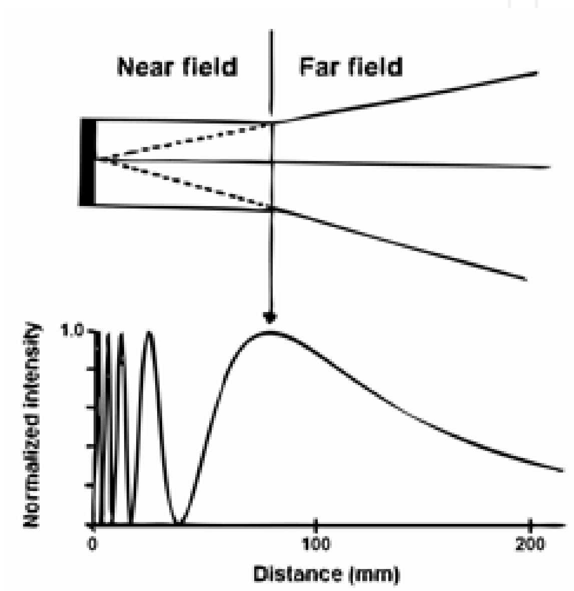


Fig. 1. *Above*, shape of the acoustic field generated by a physiotherapy ultrasonic transducer (D>10λ). *Below*, normalized acoustic intensity versus distance from the transducer (Seegenschmiedt, 1995)

The near field length ($L_{near\ field}$) is directly dependent on the diameter D and inversely proportional to the wave length (Eq. 2). The physiotherapy ultrasonic transducers have diameters bigger than the wave length and therefore they have a long near field. Because of this, when a physical therapy is being carried out, the therapeutic heating is produced inside the near field where the acoustic pressures are the result of a sum of the ones produced at different points in the piezoelectric plate. The near field of the transducer is the most important part to characterize but it is the part where the majority of the non-linearities occur.

$$L_{near\ field} = \frac{D^2}{4\lambda} \tag{2}$$

The divergence angle in the Fraunhofer zone is also dependent on the diameter and the wave length, and it can be calculated with Eq. 3 as follows

$$\sin\theta = 1.22\frac{\lambda}{D} \tag{3}$$

3.2 Transducer Characterization

When ultrasound is used to treat a muscular problem or to heal a fractured bone, it is applied following a protocol for the specific disease. Researchers have developed many protocols to treat some diseases by using different parameters of the ultrasonic device. These parameters differ in the output intensities, time of treatment, duty cycle, and frequency, and it has been considered that all values are correct (Speed, 2001). However, there are reports about calibrating medical ultrasonic devices for therapy in which it has been found that most of them are not working within nominal values (Pye & Milford, 1994).

When a therapist uses a protocol to treat a disease and the gotten results are not sufficient, he/she is going to modify the intensities in accordance with his/her experience. This behavior adds subjectivities to a treatment that nowadays is already subjective. Results dose-response have been gotten (Lu et al., 2008; Nacitarhan et al., 2005; ter Haar, 2007) but there is not a guideline to follow in order to determine the best dose (Watson, 2008). Therapists must calculate the doses base on the results reported in some papers and they must know the characteristics of the radiation produced in order to promote the desired thermal and nonthermal effects. However, the necessity of characterization is still a problem; therefore, new techniques have been developed in order to reduce the time required to make measurements and to reduce the costs.

3.3 Characterization Parameters

There are many techniques for characterizing the emission of an ultrasonic transducer in order to get the parameters of interest. Each technique measures only one magnitude of the ultrasonic beam, but with this result and applying some mathematical calculations it is possible to obtain the others. The transducer acoustic field is composed by a superposition of many waves coming from different parts of the transducer. When the design of the piezoelectric element of the transducer was not right, the generated waves have an undesirable behavior. The parameters of interest of the transducer emission have been developed in order to determine if the transducer has this adverse performance. These parameters are described in the following part of this chapter.

3.3.1 Effective Radiating Area (*ERA*)

There have been many definitions for this parameter. One of these is given by the FDA, which defines the ERA_{FDA} as the area consisting of all points of the effective radiating surface (all points within 5 mm from the applicator face) at which the intensity is 5 percent or more of the maximum intensity at the effective radiating surface, expressed in square centimeters (FDA, 2008). Recently, a new way of measuring and of defining (Hekkenberg, 1998) *ERA* which is written in the IEC standards (ABNT, 1998; IEC, 1991) was developed. This new method consists in measuring and in registering the acoustic intensities (or the proportion in mV, mPa, etc.) in four planes parallel to the transducer face at four distances along the propagation edge z. For each measured plane, the beam cross-sectional area (A_{BCS}) is calculated. This area is defined as the minimum area in a specified plane perpendicular to the "beam alignment axis" which contains 75% of the spatial integral of the "total mean square acoustic pressure" pms_t given by:

$$pms_t = \sum_{i=1}^{N} p_i^2 \tag{4}$$

where p_i is the acoustic pressure in the ith point and N is the total number of points in the scan. After that, it is considered that the near field of the beam is linearly related to z, and hence the A_{ER} (same meaning than ERA_{FDA}, Effective Radiating Area) is calculated with a relation of the extrapolation of the calculated A_{BCS}'s at $z = 0$. The A_{ER} can be calculated with Eq. 5.

$$A_{\mathrm{ER}} = F_{AC} A_{BCS} \tag{5}$$

and

$$F_{AC} = 2.58 - 0.0305ka \qquad \textit{for } ka \leq 40 \tag{6}$$

$$F_{AC} = 1.354 \qquad \textit{for } ka > 40 \tag{7}$$

where a is the effective radio of the transducer and k is the circular wave number in cm^{-1}. In this chapter, we used the ERA_{FDA} definition. The FDA definition gives large uncertainties (more than 20%) if it is compared to the IEC definition (less than 10%). Our calculation of ERA_{FDA} cannot be extrapolated to the A_{ER} of the IEC standards because the measurements and calculations are completely different (Hekkenberg, 1998; Johns et al., 2007).

For characterizing the ultrasonic emission in solid media, there is another definition which considers the Specific Absorption Rate (*SAR*) given by Eq. 8. The ESHO protocols define the *ERA* of an applicator as the 50% *SAR* contour measured at a depth of 10 mm from the surface of a plane homogeneous phantom (Hand et al., 1989). This definition is needed when it is not possible to know the acoustic intensities due to the characterization technique used (Hand et al., 1989).

The *SAR* can be calculated with:

$$SAR = C\frac{\Delta T}{\Delta t} \approx \frac{2\mu_a I}{\rho_0} \tag{8}$$

where C is the heat capacity (J/kg ·°C), ΔT is the change of the temperature (°C), Δt is the change of the time (s), μ_a is the attenuation coefficient (dB/m), I is the acoustic intensity (W/m^2), and ρ_0 is the medium density (kg/m^3).

3.3.2 Beam Non-uniformity Ratio (*BNR*)

This is the relation between the square of the maximum acoustic pressure (p_{max}) and the spatial mean square of the acoustic pressure (pms_t), where the spatial media is taken on the effective radiating area (ABNT, 1998; FDA, 2008; Hekkenberg, 1998). Eq. 9 indicates the process to calculate this parameter (ABNT, 1998)

$$BNR = \frac{{p_{\max}}^2 \cdot ERA}{pms_t \cdot a_0} \tag{9}$$

where a_0 is the area per the global raster. If the transducer is for physiotherapy, *BNR* must be in the range of 1:6 because of the patient's security (Hekkenberg, 1998). When the value is close to 1, the transducer is safer than the case when the *BNR* is close to 6.

3.3.3 Penetration Depth (P_D)

It depends on the properties of the medium where the ultrasound is passing through. In this chapter, it is used to calculate the t_{max} in the IR thermography (Eq. 12), but it can also be used to determine whether the treatment has an adequate depth. By definition, the penetration depth is the distance from the transducer where the Specific Absorption Rate (*SAR*) magnitude is 50% of the maximum magnitude at the *ERA* (Hand et al., 1989).

4. Characterization Techniques

The objective of characterizing the devices is to prevent patients' injuries because of either non-uniformities of the beam, commonly called hot-spots, or an effective radiating area different to the reported one, which modifies the total power emitted. Manufacturers deliver the devices with measurements of their characteristic parameters but with high tolerance in the measurements. For example, they tell us that the value of the *ERA* is about 10 cm^2 but with a tolerance of ±20%, which means that *ERA* could be between 8 cm^2 to 12 cm^2; the rest of the reported parameters have this kind of tolerance. Needless is to say that the sum of these uncertainties can result in an ineffective treatment or in injury to patients.

There are different transducer characterization techniques that can deliver accurate results. Most of these techniques were designed in order to improve a specific characteristic of the measurement. Some techniques are faster or cheaper than others, but they are not so accurate; there are some which are more accurate but they are too expensive or slow; with some of them it is possible to measure some magnitudes that with others is not possible, and vice versa. In this chapter, three techniques: C-scan with hydrophone, IR thermography, and Thermochromic Liquid Crystals (TLC) are going to be compared. A brief review of other techniques that could help in the characterization will also be included in order to have a better picture of the different solutions to this task.

4.1 C-scan

This technique consists in moving a small microprobe into the ultrasonic beam in order to measure the acoustic pressure levels punctually (Papadakis, 1999). The measurements are carried out into a tank filled with degasified water where all the elements (transducer and sensor) are immersed. The microprobe dimensions depend on the magnitude of interest, i.e., the C-scan technique can be used to measure the acoustic pressures instantly or the absorbed energy during a known interval of time. It is required that the sensor be as small as possible to get a good resolution. Also, the system for positioning the sensor must allow very small steps to prevent affecting the overall resolution. According to the literature, the sensors that have been used with this purpose are the hydrophones, the thermistors or the thermocouples (Marangopoulos et al., 1995), and even a reflecting ball as it will be explained later (Mansour, 1979).

The setup for carrying out the measurements has many common components among the variants mentioned. In general, the C-scan technique uses a tank, a base to fix the transducer, a system for positioning the sensor, an oscilloscope, an electronic card to excite the transducer, and the computer to register and process data. The tank must be made using ultrasonic absorbent material in order to avoid (or reduce) wave reflections (Selfridge, 1985). The water where the measurements are carried out must be degasified so the bubbles caused by the acoustic vibrations are eliminated thus avoiding the error in the results because of cavitation. A base with adjustable grips is required for fixing and centering the transducer. The sensor is fixed on the positioner XYZ which will move it transversally along the ultrasonic beam.

The setup of the experiment has some initial steps. At first, the transducer is fixed, and then a sequence of measurements aimed at finding the center is carried out. The sequence is composed by sweeps in each axis of the transducer transversal section in order to find the maximum acoustic pressure level which corresponds to the center of the piezoelectric plate. This procedure is repeated at different distances from the transducer until this one is completely centered which is determined when the movement of the sensor along the direction of the beam propagation occurred without losing the center at each distance (Vera et al., 2007). The measurements are started after the installation and the centering, and they are carried out in accordance with the problem necessities: characterization, data processing, modeling, etc. A system for 3D positioning is used in this technique.

4.1.1 Using a point reflector

This technique uses the same transducer to emit and receive the ultrasonic beam (Mansour, 1979); it works with the concept of pulse-echo. We have to know, initially, the sensitivity to ultrasound of the transducer to characterize at each point of the area of the transducer front face. This is because the ultrasound will arrive at the transducer and the energy will be changed to an electrical signal; the relation between the arriving ultrasonic energy and the electric signal generated is needed. The C-scan with point reflector, also called ball target (Papadakis, 1999), consists in positioning a small ball into the acoustic field by means of a positioner XYZ which will move the ball transversally along the beam. Although the transducer emits a cylindrical beam with a relatively large transversal section (approximately equals to *ERA*), the measured signal corresponds only to a small area just in the direction of the ball target (Fig. 2).

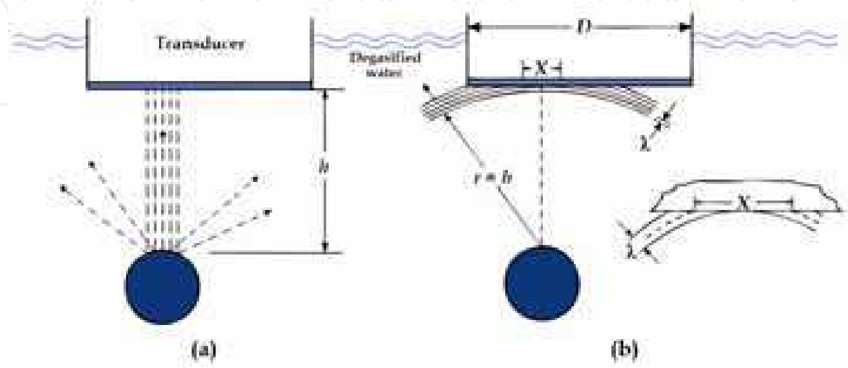


Fig. 2. C-scan with ball reflector. (a) The waves out of the pole of the ball go out of the transducer. (b) A small area above the ball performs the sampling. Destructive interference prevents the sampling of the other waves (Papadakis, 1999)

The magnitude measured represents the product of the acoustic intensity arriving at the transducer and the sensitivity of the small area above the ball of the transducer (*X* in Fig. 2b). When the plane wave returns to the transducer after the reflection, it is converted into a spherical wave. The measurement is taken only in the transducer area nearest to the ball because the wave arrives first at this part. Other wave segments are lost due to the cone-like reflection with a large angle (Fig. 2a). Even if some wave segments reach the transducer, waves beyond a specific radius are lost because of destructive interference (Papadakis, 1999). The wave measured, converted into voltage, is the result of the product of the acoustic pressure and the sensitivity at the point where the wave reaches the transducer. This characteristic could result in a problem because there is another unknown parameter that can influence the measurement.

4.1.2 Using a hydrophone

The most accurate technique until now, in accordance with IEC standards (Hekkenberg, 1998; IEC, 1991), is the C-scan with hydrophone, which uses a hydrophone as the sensor element of the C-scan system. This technique consists in moving a hydrophone inside the acoustic field while it registers the acoustic pressure at each point. This is a through-transmission technique which means that the ultrasonic transducer to characterize emits the energy and the hydrophone measures the signal, and no-reflection is considered. It is more acceptable to use this element as the sensor because it can register a time-dependent signal that can be used to get most of the required parameters not only used for characterization, but also for other applications. The utilization of a sensor for measuring directly the acoustic pressures, independently of the element to be characterized, eliminates unknown variables as the sensitivities required for the C-scan with ball reflector. The hydrophone sensitivity can be determined with a calibration, and the transducer gain per unit of area (if it would be required) could be determined by using the C-scan with a calibrated hydrophone.

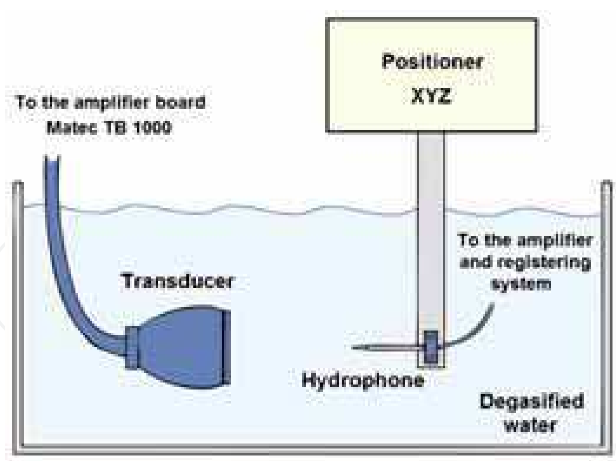


Fig. 3. Setup of C-scan with hydrophone. A more detailed diagram is shown in Fig. 10

The transducer is excited with a pulsed sinusoidal signal using either the ultrasonic equipment or a special amplifier board that produces a standard signal. The excitation signal is not continuous in order to allow the ultrasound wave to die before emitting another signal, which is required to avoid the addition of reflected waves. Therefore, a sinusoidal signal modulated by a square short pulse is used. Some parameters are required to know before applying the excitation:

Output Voltage: it is the voltage that excites the US transducer. It must be adjusted in accordance with the desired ultrasonic output power (acoustic intensity) at which the transducer is going to be characterized.

Pulse width: it is the length of the electrical square pulse that modulates the sinusoidal signal in order to excite the transducer. For example, inside an excitation with pulse width of 10 µs, there are 10 cycles of a sinusoidal signal of 1 MHz (period of 1 µs).

Repetition rate: it is the repetition period of the excitation pulses. For example, an excitation sinusoidal signal of 1 MHz modulated with a square signal of pulse width of 10 µs has a repetition rate of 13 ms because the pulse is repeated (or initiated) each 13 ms. It could be transformed into a repetition frequency which is, in this case, of $1/13e^{-3}$ Hz.

4.1.3 Using a temperature sensor

The increment of temperature in an ultrasonically irradiated medium is directly related to the acoustic intensity in the medium for a unit of time. Considering this, it is possible to use a temperature sensor as the element that registers the signal inside the acoustic field provided that the acoustic intensity is quite elevated to produce a thermal change in the medium. However, the sensor by itself is not sufficient since it must be covered by an ultrasonic absorber material which is going to be heated (Marangopoulos et al., 1995). Therefore, the temperature measured by the sensor is related to the acoustic intensity, the radiation time, and the material parameters by Eq. 10; the material parameters are the ones of the material that covers the sensor. This modality of C-scan has an overall resolution given by the size of the temperature sensor covered by the absorber material.

In contrast to the C-scans which use the above described sensors, this technique measures the energy applied during a period of time. This feature gives a relation of the measured magnitude to the applied signal which is equal to the integral of effective acoustic intensities in the media with respect to the time. The sensor cover absorbs each wave and increases its temperature as it is indicated in Eq. 10. The temperature (T) generated by the absorption of the acoustic energy for a specific time (t) is given by:

$$T = \int \frac{2\alpha I_x}{\rho C} dt \tag{10}$$

where α is the absorption coefficient of the ultrasonic energy in the medium, I_x is the acoustic intensity at a depth of x (W/cm^2), ρ is the medium density (kg/m^3) and C is the heat capacity of the medium (J·kg^{-1}·K^{-1}).

4.2 Schlieren technique

This technique applies the Schlieren effect, discovered by Robert Hook (Rienitz, 1975). It uses a two candle system to visualize the ultrasonic beam; the first explanation of the phenomenon in ultrasonic waves was made by Raman-Nath in 1935 (Johns et al., 2007).

Schlieren techniques make density gradients in transparent media visible based on the deflection of light that passes through it. This characterization technique consists in sending a beam of light normal to the ultrasonic beam. When the longitudinal ultrasonic beam travels through a medium, the medium local densities are changed because of the compressions and rarefactions of the beam. These changes in density modify the optical index. Hence, the light passing through the ultrasonic beam changes the direction in accordance with the acoustic intensities (Hanafy & Zanelli, 1991).

The system is composed by a source of light (emitter) which is normally a laser or an arc lamp which produces high intensity uniform light. The light has to be collimated by using a system of lenses as shown in Fig. 4. The refracted light is sensed by the camera at the other side of the emitter. The acoustic beam is covered mostly by the light beam, which allows relating the collected light intensity and the acoustic radiation pressure (Hanafy & Zanelli, 1991). The light is strobed at a fixed delay after emitting the ultrasound pulses. This does not affect the image formation at the video camera because the image appears to be static, but this permits to avoid taking the image of the ultrasound reflected wave. The ultrasound absorber does not avoid reflections; it just reduces them significantly.

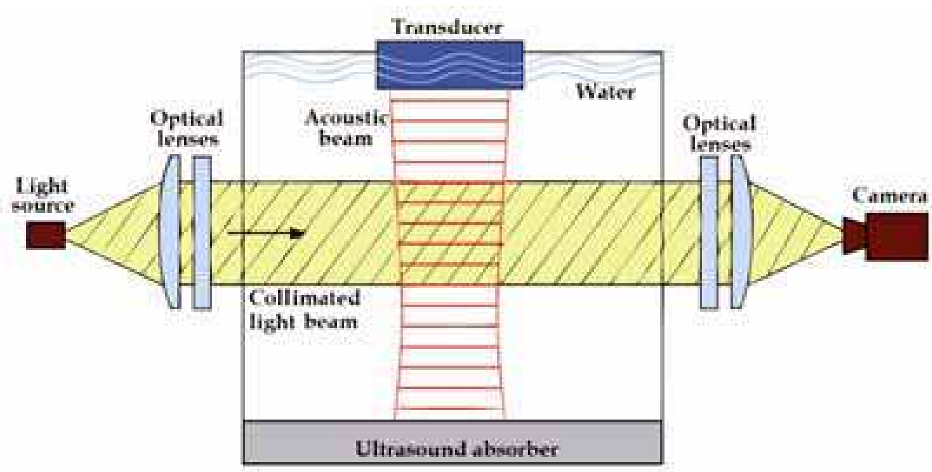


Fig. 4. Schlieren system

Optical intensity at each pixel is proportional to the acoustic intensity integrated along the line where the light passed through. This statement is true provided that

1. the acoustic wave fronts are quasi-planar and normal to the light beam, and
2. the acoustic intensity is low enough to avoid acousto-optic nonlinearities.

Both conditions are satisfied if some considerations are taken. Condition 1 is satisfied if the transducer is aligned by measuring the acoustic intensities at each point of the transversal section. Condition 2 is satisfied by adjusting the acoustic intensity in order to be sure that the changes are within 5% of linearity, which is commonly true at low acoustic intensities, less than 0.2 W/cm^2. When the system does not satisfy these conditions, the optic intensity is not linearly proportional to the acoustic intensity; hence, Schlieren system cannot be used quantitatively.

This technique is very useful because it does not affect the acoustic emission and it permits to have the acoustic beam without the previous knowledge of the shape. However, the system is expensive and it requires some critical adjustments: lense alignment, high intensity light, transparent propagation media, ultrasound low intensities, high quality optics, etc. Moreover, it is not possible to get a punctual acoustic intensity, but an acoustic intensity integrated along the optical path. Researches continue in order to find a way to eliminate these disadvantages, e.g., reducing the cost of the lamp for emitting high intensity light (Gunarathne & Szilard, 1983).

4.3 Sarvazyan technique

This method was proposed by Sarvazyan et al. in 1985. It is a simple and rapid method that consists in mapping the ultrasound fields using a white paper and an aqueous solution of methylene blue dye. The paper is an Astralux 200 µm card (Star Paper Company, Blackburn, Lancashire) that has demonstrated being suitable in characterizing the ultrasound emission at frequencies around 1 MHz. The field is directed at a sheet of paper through the blue solution during 1 minute. After this exposition time, there is a pattern of dye formed in the paper which is related to the intensity distribution of ultrasound (Watmough et al., 1990). The patterns obtained along the ultrasound beam are processed in order to get the acoustic intensities at each point in the card.

The dye diffusion is because the paper has microbubbles on the surface due to the microscopic irregularities. The size of the microbubbles depends on the paper, and because of that, it is not possible to use any kind of paper. Astralux card has microscopic holes of 3 µm of diameter which are resonant at frequencies around 1 MHz. Resonant gas bubbles are related to microstreaming of the liquid surrounding them, and this is the phenomenon that causes the increment of dye diffusion in high acoustic intensity areas (Shiran et al., 1990; Watmough et al., 1990). The resolution technique depends on the distance between the gas bubbles.

This technique has some disadvantages, e. g. the gas bubbles cause ultrasound reflections to the transducer; this can affect the radiation pattern and consequently the characterization results. Because of gas bubbles, Astralux paper is not ultrasound "transparent" and stationary waves could be formed that could even cause damages to the ultrasonic transducer. Also, it is not a reversible technique, hence the paper must be changed before the next measurement and this can result in errors because of differences in the position of the papers.

4.4 Holography with Flexible Pellicle

This technique was proposed by (Mezrich et al., 1975) to display the ultrasonic waves by using a flexible pellicle and the Michelson interferometer. The pellicle is located into the ultrasound field perpendicularly to the ultrasound propagation in order to have movement in the pellicle proportional to the acoustic intensities. As pellicle (M_2) moves, the relative phase between M_1 and M_2 varies and produces intensity changes at photodiode D. The laser beam is moved in order to scan the displacement at every point of the pellicle. The interferometer is shown in Fig. 5, and in this system, the pellicle thickness is 6 µm.

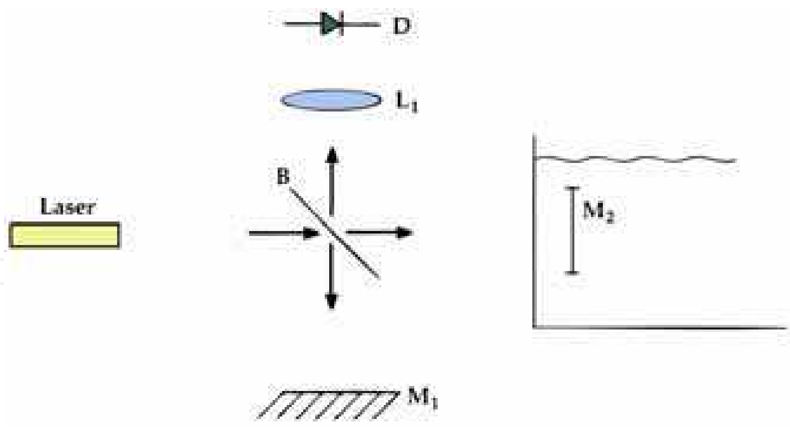


Fig. 5. Michelson interferometer used by (Mezrich et al., 1976) to detect the acoustic displacement amplitude

4.5 Optical Computerized Tomography

This technique uses a Michelson interferometer in order to visualize the ultrasonic beam based on the modified index of refraction gradient caused by the ultrasound pressures. It is similar to the Schlieren technique but it uses the Michelson interferometer to visualize the transducer beam. The light passes through the ultrasound beam and it is compared to the reference light in order to determine the optical intensity that has the information about the acoustic intensity. This method compares the light phase as well as the light intensity of each beam. The light is detected by an avalanche photodiode and the data are postprocessed in order to reconstruct the beam (Obuchi et al., 2006).

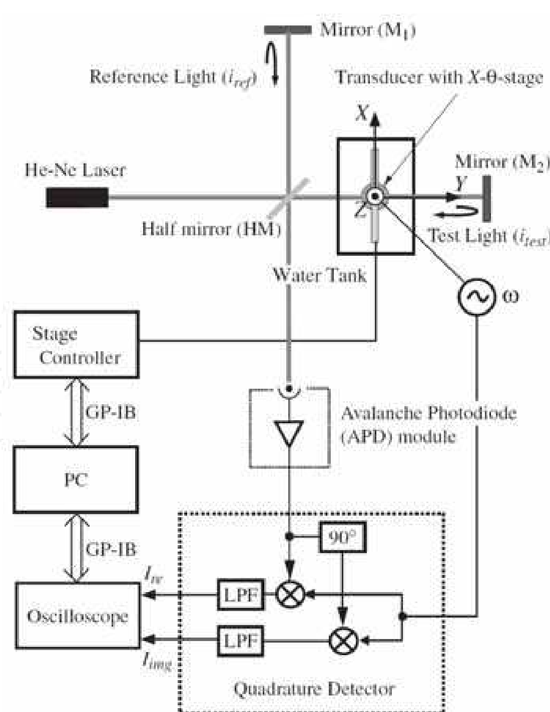


Fig. 6. Schematic diagram and experimental setup for the system proposed by (Obuchi et al., 2006)

The setup of the system for visualization of the ultrasonic beam designed by (Obuchi et al., 2006) is shown in Fig. 6. The light beam is divided in two: the reference light and the test light. When both beams are reflected by their respective mirror, they go to the avalanche photodiode which generates a signal with all the information. The signal is processed by a quadrature detector considering the driven frequency of the ultrasound transducer, ω. The optical intensity resulted is I_{out} which is given by

$$I_{out} = I_{test} + I_{ref} + 2\sqrt{I_{test} I_{ref}} \cos\phi(x, z, t) \tag{11}$$

where I_{test} is the test light intensity, I_{ref} is the reference light intensity and ϕ is the phase difference between these lights.

4.6 Thermography in liquids and solids

As it was described before, the increment of temperature is directly related to the acoustic intensities in a specific medium (Eq. 10). It is possible to get the characteristic parameters by measuring this temperature directly in the heating media. Next, three techniques of temperature measurement that can be used in characterization of ultrasound transducers are described.

4.6.1 Invasive Thermography

This technique consists in measuring the temperature in the irradiated solid medium (phantom) using a temperature sensor inserted in it. The sensor must not be affected by the ultrasonic radiation and it must be as small as possible in order to have a punctual measurement. The data are registered in a matrix containing all the information about the measurements and the place where they were taken with respect to the transducer. Measurements are carried out upwards using as many sensors as possible because the values are going to be processed in order to get the parameters of study. The measurement is carried out in this direction because the phantom is destroyed when the sensor is inserted; if measurements are performed in the opposite direction, this destruction can cause problems with the ultrasonic propagation. When the measurements are made starting at the bottom of the phantom, we can be sure that the propagation is correct from the transducer to the inserted sensor, and that the destroyed part is left behind. Postprocessing is required to relate the measurements to the characteristic parameters or to reconstruct the thermal field to calculate the penetration depth, the absorption (*SAR*), etc.

Even though this technique has some interesting advantages, the disadvantages could be even more important. This technique requires little specialized equipment and relatively simple postprocessing, and its temperature sensors (the thermocouples or the thermistors) are not affected by ultrasound. However, whichever sensor is inserted into the media will produce a hot-spot caused because the media and the sensor have different acoustic impedances. This difference causes backward wave reflections and therefore the addition of the arriving and the returning waves; this can be observed as an increment of temperature in that point. Another disadvantage is the time used for the measurements. For each line, it is required to heat the phantom while the sensors are measuring, and to wait until the phantom reaches the original temperature before taking another line.

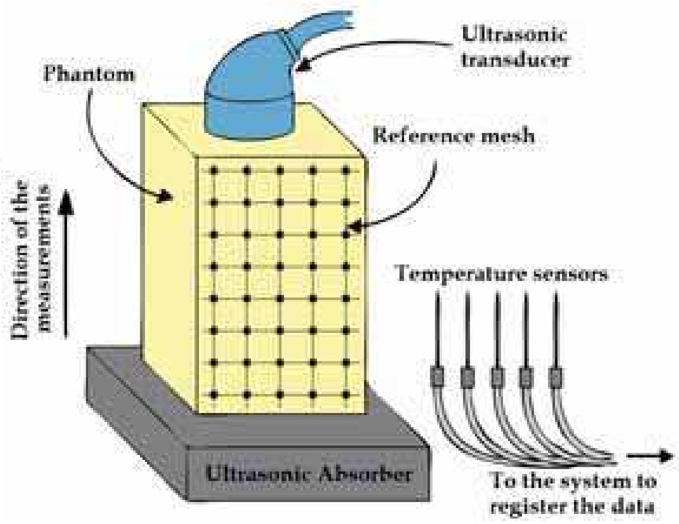


Fig. 7. Invasive Thermography setup. All the elements are fixed. The mesh is the reference for inserting the sensors. The distance between the sensors is the spatial resolution of the system

4.6.2 IR Thermography

The use of devices to capture IR images is another alternative for the characterization of the ultrasound effects (Guy, 1971). The IR radiation caused by any material depends directly on its temperature. Nowadays, there are cameras that can be used to detect the IR radiation and convert it to a thermal image related to a temperature color scale; these cameras are called IR cameras. However, for making the measurement, it is necessary to have the temperature of interest at the surface of the material. A modification of the setup proposed in 1991 for electromagnetic applicators is shown in Fig. 8 (Andreuccetti et al., 1991b). It consists in a phantom cut by the half and an IR camera that takes the picture. The phantom is heated by the US transducer during a period of time at which it is separated to make visible its internal part. The picture is taken before the complete temperature dissipation occurs. After each picture, the phantom in cooled and the transducer is moved in order to get another plane of the beam. The displacements are part of the resolution because the overall resolution is limited mainly by the IR camera and the non-homogeneities of the phantom.

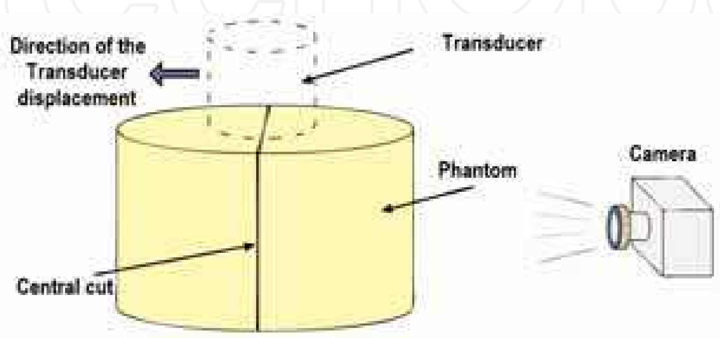


Fig. 8. IR thermography system setup. The transducer is moved after taking each picture in order to get the complete beam

Andreuccetti et al. established the maximum time to get the IR image in relation to the penetration depth; they got this empirical formula:

$$t_{\max} = 13.2 P_D^{\ 2} \tag{12}$$

where P_D is the penetration depth (section 3.3.3) in centimeters and $t_{\max}$ is the maximum time for taking the picture in seconds. Disadvantages of this technique are the difficulties to fix the transducer and the phantom and the difficulty in being sure that the displacement of the transducer is the desired one. Methods of registering data can solve this problem by taking some planes and rotating the transducer in order to measure another plane.

4.6.3 Thermography with Thermochromic Liquid Crystals (TLC)

This thermographic technique uses a TLC sheet to create a colored image which corresponds to the heat produced by the ultrasonic absorption in a medium. The sheets that contain TLC (sometimes called thermochromic sheets) are used as sensors and they have to be in contact with the medium heated (Gutierrez et al., 2008). The ultrasound generates heat when the media are highly absorbent; therefore, a phantom with a large absorption coefficient must be used. The technique can be applied when the characterization of the transducer emission in either liquid or solid media is required, but modifying the system setup for each situation in order to have a coherent measurement.

Thermography with TLC in solid media is made by using a setup as the one shown in Fig. 8 but by placing in the middle, where the cut is made, the TLC sheet. The heat is transmitted from the phantom to the TLC sheet which creates the thermal image. The picture is taken by using a normal camera because the image is in the visible range (Andreuccetti et al., 1991a; Andreuccetti et al., 1991b). The disadvantage of this application is that the TLC sheet has to be implemented in contact with the medium during all heating and this produces distortion of the transducer radiation pattern because of the acoustic differences between the TLC sheet and the phantom.

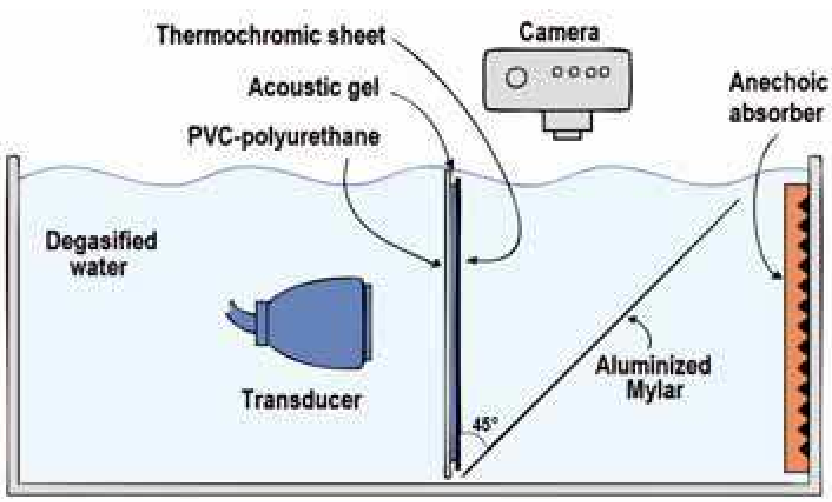


Fig. 9. Thermography with TLC setup. The color image is related to the effective acoustic intensities

The other application is in the characterization of the acoustic emission in transparent liquid media, commonly degasified water (Martin & Fernandez, 1997). In this technique, the transducer is placed inside a container as it is indicated in Fig. 9. The transducer radiates through the water in direction to the TLC sheet and the absorbing layer, which are placed perpendicular to the acoustic propagation. The ultrasound will continue beyond the layers and it will reach an absorber material at the end of the tank in order to avoid ultrasound reflections. The image is taken with a common camera through a Mylar mirror; the distributions of color obtained are related to the energy absorbed converted punctually into heat. The transducer has to be moved by a positioner along the direction of propagation in order to capture the required images along the acoustic field. Postprocessing is required in order to get the characteristic parameters or to reconstruct the complete acoustic beam.

5. Measurements

The techniques described before could be used to get characteristic parameters of an ultrasonic device. This chapter presents a comparison of the characterization results gotten with three techniques with different features: time consuming, accuracy, and kind of media where ultrasound passes through. The characterized device was a physiotherapy ultrasonic equipment Ibramed from Brazil; its principal features are shown in Table 1.

Manufacturer	Ibramed, Brazil
Model	Sonopulse
Frequency	1 or 3 MHz (+/- 10%)
ERA	3.5 cm² or 1 cm² (+/- 20%)
Output power	0.1 to 2 W/cm² (+/- 20%)
BNR	<8:1 (+/- 30%)

Table 1. Principal features of the characterized physiotherapy ultrasonic equipment

5.1 Methodology

5.1.1 C-scan with hydrophone

The C-scan used for these measurements is composed by a PZTZ44-0400 hydrophone as the sensor element. The hydrophone signal is amplified 17 dB before sending the signal to the oscilloscope, which has a 150 MHz bandwidth (TDS-340, Tektronix, USA). The transducer is excited with an amplifier/receiver board TB 1000 (Matec Instruments, USA) using the next parameters: repetition rate of 13 ms, frequency of 1.05 MHz, pulse width of 6.22 μs, and output voltage of 120 V. The generator board and the oscilloscope are synchronized and the computer registers a new value each time the oscilloscope is ready to measure, i.e., when the signal is stable. The block diagram of the C-scan system can be seen in Fig. 10.

During the characterization, the positioner XYZ and the oscilloscope are controlled by a computer with the software Scan 340 and through a GPIB interface. For these measurements, the software was configured to read transversal planes of 40 mm X 40 mm separated from the transducer by a distance that varied from 2 mm to 100 mm until completing 51 planes. These data were saved in a matrix of 40 X 40 X 51 so they could be taken in order to calculate the ultrasonic beam parameters.

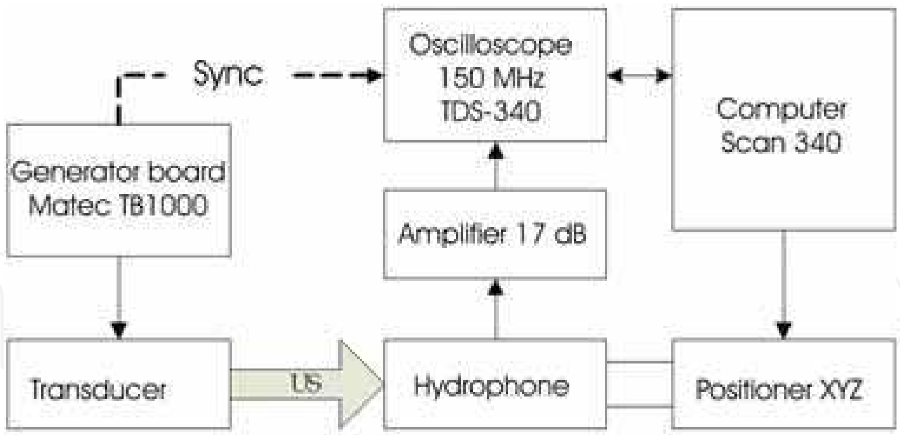


Fig. 10. Block diagram of the C-scan with hydrophone. The setup of the system is described in section 4.1.2

5.1.2 Thermochromic Liquid Crystals (TLC)

The thermochromic sheet used for the experiments (R25C5W, Hallcrest, USA) has a "red start to blue start" range of 25-30°C. The medium that absorbs the ultrasound is a sheet of PVC-polyurethane with 0.4 mm of thickness, a sound reflection coefficient in water of 0.06 (±0.01) and an attenuation coefficient of 23.5 (±1.02) $dB \cdot cm^{-1} \cdot MHz^{-1}$. All the systems are mounted in a degasified-water-filled tank as it is shown in Fig. 9. The anechoic absorber has an attenuation of 45 dB/cm at 1 MHz and 25°C (Zeqiri & Bickley, 2000). The thermal image was observed through the aluminized Mylar which is practically "transparent" for the ultrasound because of its small acoustic reflection coefficient.

The pictures were taken after 30 min of ultrasonic radiation with an acoustic intensity of 2 W/cm^2; this time was chosen because it was the moment when the stabilization of the temperature pattern took place. The transducer was moved backward the absorbing layer in steps of 2 mm starting at 2 mm from the absorbing layer until it reached 13 mm from the layer.

5.1.3 IR thermography

The temperature in a phantom of soft-tissue was obtained and a protocol to take the photographs was used. Initially, the phantom was made with a cylindrical form; then, it was cut at the middle in order to separate the two parts. After heating, the phantom was divided and a photograph was taken as fast as possible in order to avoid thermal conduction effects; before taking another thermal image, the phantom was cooled and the transducer was displaced perpendicularly to the surface where the picture had been taken in order to have images at different axial cuts of the acoustic field. The images were taken with an IR camera Mikron model TH5104. The penetration depth was measured to know the maximum time to get the image.

The phantom elaborated was agar-based with a combination of ingredients to simulate the acoustic properties of the muscle: 2.5% agar, 5% graphite, 10% glycerol and distilled water. The agar is used to solidify the resulted mixture; the graphite, to simulate the acoustic attenuation; the glycerol, to get the desired acoustic velocity; and the distilled water, to dissolve the materials. The resulted phantom had the acoustic properties shown in Table 2.

Magnitude (20°C)	Measured	Muscle
Attenuation (dB/cm)	0.807	0.5-1.3
Velocity of propagation (m/s)	1537.51	1520-1580

Table 2. Acoustic properties of the phantom compared to those of the muscle

5.2 Radiation pattern

In order to compare the results, the characteristic parameters were obtained with each technique by using the radiation pattern. Although the transducer was always the same, the radiation patterns are different because the measured magnitudes were different for each technique. However, the results can be compared because every magnitude is related to the acoustic intensity as it has been explained.

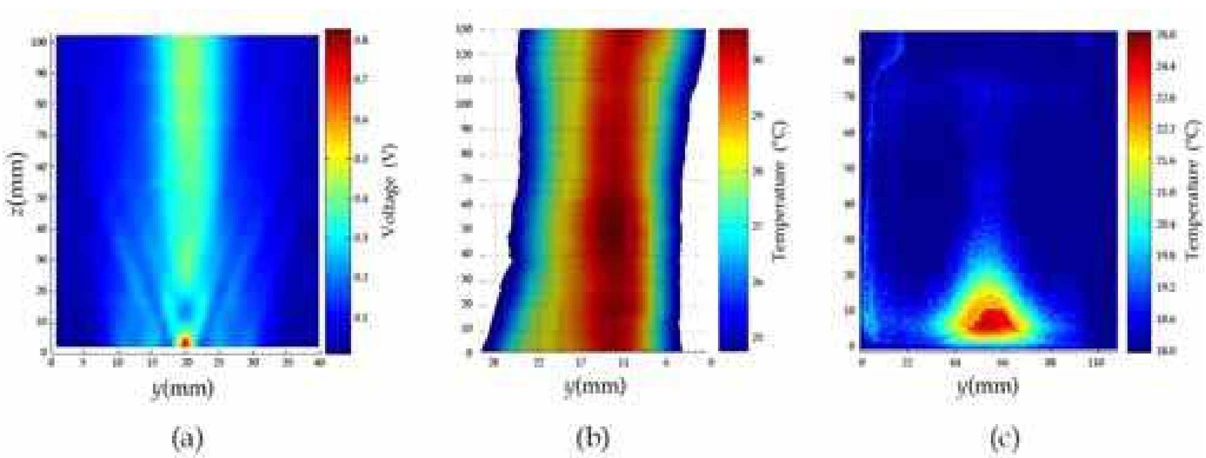


Fig. 11. Radiation pattern of the characterized transducer. (a) C-scan with hydrophone, (b) thermography with TLC, (c) IR thermography

The radiation patterns are shown in Fig. 11. The acoustic pressure distribution around the ultrasonic beam when the technique C-scan was used is shown in Fig. 11a. We can notice that the acoustic pressure magnitude represents the “free ultrasonic field” since the attenuation is negligible. This is because the medium where the ultrasound passes through was water which has an attenuation coefficient of $2.5x10^{-4}$ Np/cm at 1 MHz. The increments near the transducer are caused by the wave cancelations (and additions) in the near field zone.

The radiation pattern of the transducer was obtained by using the thermography with TLC; the pattern was composed by the temperatures measured, as it is shown in Fig. 11b. There was a visible effect in the inclination of the figure due to the internal inclination of the piezoelectric element with respect to the transducer face. The initial sequence to find the center (as in C-scan) was not enough because of the dependence on the system angles. The Fig. 11b shows the radiation pattern resulted from the interpolation among every image taken.

Finally, the radiation pattern of Fig. 11c was obtained using the IR thermography. This picture is completely different from the pictures showed in Fig. 11a and 11b because of the differences among the techniques. In Fig. 11a, the radiation pattern was formed with instantaneous acoustic pressures, and in Fig. 11c the magnitude measured was the quantity of the ultrasound absorbed (effective acoustic intensity was transformed into temperature). In spite of the different pattern, it is possible to calculate the characteristic parameters using this results; moreover, this is the most appropriated technique, of the three compared here, for calculating the penetration depth in the media, if it were required.

5.3 Effective Radiating Area

This was calculated using the FDA definition given above. Considering the results of the C-scan, Fig. 12a, the region with the 95% of the intensity starts at the sample 11 and ends at the sample 28. Because each sample was equal to 1 mm, we have a diameter of 17 mm, and then an *ERA* of 227 mm^2.

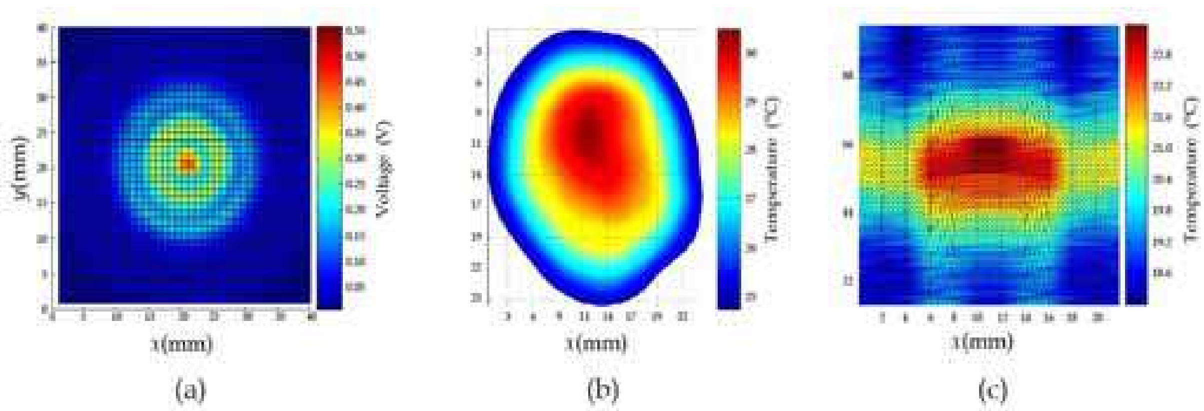


Fig. 12. Radiation pattern used to calculate the *ERA*. (a) C-scan with hydrophone, (b) thermography with TLC, (c) IR thermography

In order to calculate the *ERA* using the TLC, a MATLAB algorithm was used (see the Fig. 12b). This algorithm takes the color of the image, creates a relation of the color and the temperature, transforms the color image to a temperature data matrix, finds the maximum *SAR*, and determines the area with the 50% of the maximum *SAR* in order to have the *ERA* in agreement with the ESHO protocols (Hand et al., 1989). Same calculations were carried out to find the *ERA* using the measurements with the IR thermography. The results are shown in Table 3.

Technique	*ERA* (mm^2)
C-scan with hydrophone	227
Thermography with TLC	201
IR-thermography	154

Table 3. *ERA* calculations using each technique. C-scan is the most accurate in accordance with the IEC standards. Thermography with TLC gave a result close the one of the C-scan

5.4 Beam Non-uniformity Ratio

This parameter was calculated by using the definition given above, which corresponds to Eq. 9. The results are quite similar and they are in the safe range of 1:6. The calculations for the techniques that use the temperature as the magnitude measured were the same as the C-scan. The acoustic intensity is related to the acoustic pressure and to the increment in temperature, as it can be seen in the next equations:

$$I_x = \frac{p_{rms}^2}{\rho c} \tag{13}$$

where c is the sound velocity. Using Eq. 10 for the temperature increment and substituting in Eq. 13, we have

$$p_{rms}^2 = \frac{C\rho^2 c}{2\alpha}\frac{dT}{dt} \tag{14}$$

Hence, the *BNR* adjusted for thermographic techniques is

$$BNR = \frac{\Delta T_{\max} \cdot ERA}{\Delta T_{average} \cdot a_0} \tag{15}$$

where $\Delta T_{\max}$ is the maximum temperature increment and the $\Delta T_{average}$ is the average of every temperature increment measured inside the *ERA*. The results of the calculation of the *BNR* with the three techniques are registered in Table 4. Again, the similarities of the results of using the thermography with TLC and the C-scan with hydrophone are clear. However, the IR thermography was different.

Technique	*BNR*
C-scan with hydrophone	1.52
Thermography with TLC	1.69
IR-thermography	1.29

Table 4. Calculations of the *BNR*. The three techniques had close results. IR thermography was, again, a little different

5.5 Penetration Depth

This parameter was obtained with the IR thermography in order to know the $t_{\max}$ to take the pictures. The penetration depth was calculated by using the definition described in section 3.3.3; the radiation pattern of Fig. 11c was used for the calculations. The axial *SAR* was determined along the beam, and the reduction of 50% of the one calculated at 10 mm was the P_D. Another technique that could be used and which gives accurate results is the invasive thermography by using either thermocouples or thermistors. The thermography in solids with TLC is another option but with the disadvantage of the effect on the acoustic beam caused by the TLC sheets.

The P_D calculated for the transducer radiation on the muscle phantom was 2.086 cm. This indicates that the maximum time to take the picture, in accordance with Eq. 12, is given by

$$13.2(2.086^2) = 57.44 \text{ s.}$$

6. Conclusion

Physiotherapy ultrasonic transducer characterization is required because the patient must be safe. Effects of hot-spots produced by an inhomogeneous beam (large *BNR*) could injure tissue because of the physiological implications: ablation and cavitation. Moreover, the efficacy of the treatment depends on the device acoustic intensity and this is related to the ultrasonic power and the transducer *ERA*. Ignorance of these characteristic parameters can generate undesired effects and even tissue damage, as it has been mentioned before.

In this chapter, measurements using each technique were carried out in order to determine the mentioned parameters. By using these techniques, it is possible to determine both the *ERA* and the *BNR* of the physiotherapy transducer. The resolutions and the time of acquisition are different for each technique; it was determined that the TLC was the fastest one and the C-scan was the slowest. Even though this comparison does not include other techniques, their use is not excluded. Some improvements are still required so these techniques can be applied confidently in the characterization of ultrasonic transducers, as the reduction of the costs, the uncertainties, and the technological requirements.

The comparison of three of these techniques gave us the possibility to know their characteristics in order to plan the future measurements in accordance with the specific requirements. The C-scan with hydrophone is recommended when a good resolution and high reproducibility of the measurements are required. With this technique, it is also possible to determine the acoustic intensity in every point in the acoustic field, and therefore, to know most of the parameters required in experiments in liquids, i.e., the acoustic pressure, the total power, the attenuation, etc. IR thermography can be used to characterize the US devices when they emit in solid media. It is possible to calculate the desired parameters mentioned before, but taking into consideration that the measurements are the result of the absorption of the same acoustic intensities during a period of time (integrated in time). The technique with TLC is a new proposal to characterize a device quickly. It can be used when fast and high resolution measurements are required. The configuration is similar to that of the C-scan, but this technique uses a sheet to measure the heat produced in an acoustic absorber material. This technique also uses time-integrated magnitudes, but with water between the transducer and the absorber; therefore, it is used to characterize the emission in liquid media. The other variant of the thermography with TLC can be used in solid media, but its application was not mentioned here.

Most of the techniques described along this chapter work adequately under certain conditions, and their use in characterization is chosen in accordance with the necessities. The Schlieren technique has been widely used since their application for quantifying the ultrasound emission. The use of a Michelson interferometer could be another good option, but expensive. However, the challenge is to find an accurate technique that allows characterizing transducers quickly and inexpensively. According to this comparison, the most accurate technique, in regards to IEC standards and FDA definitions, is the C-scan with hydrophone, but as it was mentioned before, it is the slowest one. Techniques as the thermography with TLC and the Schlieren method could be improved to increase its accuracy and to reduce its costs, respectively. It is also required to find a portable system to characterize the transducer *in situ* which does not require a lot of time in the all the process.

7. References

ABNT (1998). "Ultra-som - Sistemas de fisioterapia - Prescriçoes para desempenho e métodos de mediçao na faixa de freqüências de 0,5 MHz a 5 MHz.," 1998, ABNT - Associaçao Brasileira de Normas Técnicas

Águila C. d. (1994). Ultrasonido Terapéutico. In: *Electromedicina*, 291-302, Hispano Americana S.A. Hasa

Andreuccetti D., Bini M., Ignesti A., Olmi R., Priori S. & Vanni R. (1991a). Characterization of hyperthermia applicators by semi-automatic liquid crystal dosimetry. *Physica Medica*, 7, 4, 145-151

Andreuccetti D., Bini M., Ignesti A., Olmi R. & Vanni R. (1991b). Phantom Characterization of Applicators by Liquid-Crystal-Plate Dosimetry. *Int J Hyperther*, 7, 1, 175-183, 0265-6736

Arthur R. M., Straube W. L., Starman J. D. & Moros E. G. (2003). Noninvasive temperature estimation based on the energy of backscattered ultrasound. *Med Phys*, 30, 6, 1021-1029, 0094-2405

Arthur R. M., Straube W. L., Trobaugh J. W. & Moros E. G. (2005). Non-invasive estimation of hyperthermia temperatures with ultrasound. *Int J Hyperther*, 21, 6, 589-600, 0265-6736

Christensen D. A. (1988). *Ultrasonic bioinstrumentation*, John Wiley & Sons, 0471604968

De Poorter J., De Wagter C., De Deene Y., Thomsen C., Stahlberg F. & Achten E. (1995). Noninvasive MRI thermometry with the proton resonance frequency (PRF) method: in vivo results in human muscle. *Magn Reson Med*, 33, 1, 74-81, 0740-3194 (Print)

Fallone B. G., Moran P. R. & Podgorsak E. B. (1982). Noninvasive thermometry with a clinical X-ray CT scanner. *Med Phys*, 9, 5, 715-721, 0094-2405 (Print)

FDA (2008). "Performance Standards for Sonic, Infrasonic, and Ultrasonic Radiation-Emitting Products ", April 1 2008, International Electrotechnical Commission

Feril L. B. & Kondo T. (2004). Biological effects of low intensity ultrasound: the mechanism involved, and its implications on therapy and on biosafety of ultrasound. *Journal of Radiation Research*, 45, 4, 479-489

Field S. B. & Bleehen N. M. (1979). Hyperthermia in the Treatment of Cancer. *Cancer Treat. Rev.*, 6, 2, 63-94, 0305-7372

Gunarathne G. P. P. & Szilard J. (1983). A New Stroboscope for Schlieren and Photo-Elastic Visualization of Ultrasound. *Ultrasonics*, 21, 4, 188-190, 0041-624X

Gutierrez M. I., Leija L. & Vera A. (2008). "Therapy Ultrasound Equipment Characterization: Comparison of Three Techniques," *30th Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, Vancouver, British Columbia, Canada, Aug 20-24 2008, IEEE

Guy A. W. (1971). Analyses of Electromagnetic Fields Induced in Biological Tissues by Thermographic Studies on Equivalent Phantom Models. *IEEE T Microw Theory*, Mt19, 2, 205-&, 0018-9480

Hanafy A. & Zanelli C. I. (1991). "Quantitative real-time pulsed Schlieren imaging of ultrasonic waves," *IEEE 1991 Ultrasonics Symposium Proceedings. (Cat. No.91CH3079-1)*, 1223-1227 vol.1222, Orlando, FL, USA, 8-11 December 1991, IEEE

Hand J. W., Lagendijk J. J. W., Andersen J. B. & Bolomey J. C. (1989). Quality Assurance Guidelines for ESHO Protocols. *Int J Hyperther*, 5, 4, 421-428, 0265-6736

Hekkenberg R. T. (1998). Characterising ultrasonic physiotherapy systems by performance and safety now internationally agreed. *Ultrasonics*, 36, 1-5, 713-720, 0041-624X

Hensley A. B. & Muthuswamy J. (2002). "Ultrasound induced permeabilization of cell membranes as a therapy for cytotoxic neuronal edema," *Conference Proceedings. Second Joint EMBS BMES Conference 2002 24th Annual International Conference of the Engineering in Medicine and Biology Society. Annual Fall Meeting of the Biomedical Engineering Society. Houston, TX, USA. 23 26 Oct. 2002.* 2002, IEEE, Piscataway, NJ, USA

IEC (1991). "Measurement and characterization of ultrasonic fields using hydrophones in the frequency range 0.5 MHz to 15 MHz," 1991, International Electrotechnical Commission

Johns L. D. (2002). Nonthermal effects of therapeutic ultrasound: The frequency resonance hypothesis. *J Athl Training*, 37, 3, 293-299, 1062-6050

Johns L. D., Demchak T. J., Straub S. J. & Howard S. M. (2007). The role of quantitative schlieren assessment of physiotherapy ultrasound fields in describing variations between tissue heating rates of different transducers. *Ultrasound Med Biol*, 33, 12, 1911-1917, 0301-5629

Lu H., Qin L., Cheung W., Lee K., Wong W. & Leung K. (2008). Low-intensity pulsed ultrasound accelerated bone-tendon junction healing through regulation of vascular endothelial growth factor expression and cartilage formation. *Ultrasound Med Biol*, 34, 8, 1248-1260, 0301-5629

Maass-Moreno R. & Damianou C. A. (1996). Noninvasive temperature estimation in tissue via ultrasound echo-shifts. Part I. Analytical model. *Journal of the Acoustical Society of America*, 100, 4, 2514-2521

Mansour T. M. (1979). Evaluation of Ultrasonic Transducers by Cross-Sectional Mapping of the near-Field Using a Point Reflector. *Mater Eval*, 37, 7, 50-54, 0025-5327

Marangopoulos I. P., Martin C. J. & Hutchison J. M. S. (1995). Measurement of Field Distributions in Ultrasonic Cleaning Baths - Implications for Cleaning Efficiency. *Physics in Medicine and Biology*, 40, 11, 1897-1908, 0031-9155

Martin K. & Fernandez R. (1997). A thermal beam-shape phantom for ultrasound physiotherapy transducers. *Ultrasound Med Biol*, 23, 8, 1267-1274, 0301-5629

Mezrich R., Etzold K. F. & Vilkomerson D. (1975). Interferometric Measurement and Display of Ultrasonic-Waves. *IEEE J Quantum Elect*, 11, 9, D9-D10, 0018-9197

Mezrich R., Vilkomerson D. & Etzold K. (1976). Ultrasonic-Waves - Their Interferometric Measurement and Display. *Appl Optics*, 15, 6, 1499-1505, 0003-6935

Nacitarhan V., Elden H., Kisa M., Kaptanogglu E. & Nacitarhan S. (2005). The effects of therapeutic ultrasound on heart rate variability: A placebo controlled trial. *Ultrasound Med Biol*, 31, 5, 643-648, 0301-5629

Obuchi T., Masuyama H., Mizutani K. & Nakanishi S. (2006). Optical computerized tomography for visualization of ultrasonic fields using Michelson interferometer. *Jpn J Appl Phys 1*, 45, 9A, 7152-7157, 0021-4922

Paliwal S. & Mitragotri S. (2008). Therapeutic opportunities in biological responses of ultrasound. *Ultrasonics*, 48, 4, 271-278, 0041-624X

Papadakis E. P. (1999). *Ultrasonic Instruments & Devices: Reference for Modern Instrumentation Techniques, and Technology*, Academic Press

Pernot M., Tanter M., Bercoff J., Waters K. R. & Fink M. (2004). Temperature estimation using ultrasonic spatial compound imaging. *IEEE Transactions on Ultrasonics, Ferroelectrics and Frequency Control*, 51, 5, 606-615, 0885-3010

Pye S. D. & Milford C. (1994). The performance of ultrasound physiotherapy machines in Lothian Region, Scotland, 1992. *Ultrasound Med Biol*, 20, 4, 347-359, 0301-5629 (Print)

Rienitz J. (1975). Schlieren Experiment 300 Years Ago. *Nature*, 254, 5498, 293-295, 0028-0836

Seegenschmiedt M. H. (1995). *Thermo-radiotherapy and Thermo-chemotherapy*, Springer,Atlanta, GA

Selfridge A. R. (1985). Approximate material properties in isotropic materials. *IEEE Transactions on Sonics and Ultrasonics*, SU-32, 3, 381-394

Shiran M. B., Quan K. M., Watmough D. J., Abdellatif K., Mallard J. R., Marshall D. & Gregory D. W. (1990). Some of the Factors Involved in the Sarvazyan Method for Recording Ultrasound Field Distributions with Special Reference to the Application of Ultrasound in Physiotherapy. *Ultrasonics*, 28, 6, 411-414, 0041-624X

Singh V. R., Chauhan S., Yadav S. & Chakarvarti S. K. (1990). Ultrasonic velocity as a non-invasive measure of temperature in biological media. *Applied Acoustics*, 29, 1, 73-80

Speed C. A. (2001). Therapeutic ultrasound in soft tissue lesions. *Rheumatology*, 40, 12, 1331-1336, 1462-0324

ter Haar G. (1999). Therapeutic ultrasound. *European Journal of Ultrasound*, 9, 1, 3-9

ter Haar G. (2007). Therapeutic applications of ultrasound. *Prog Biophys Mol Bio*, 93, 1-3, 111-129, 0079-6107

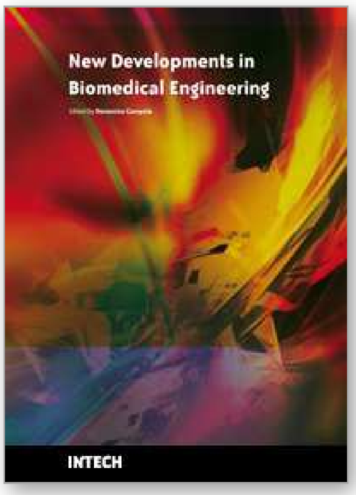
ter Haar G. & Hand J. W. (1981). Heating techniques in hyperthermia. III. Ultrasound. *The British journal of radiology*, 54, 642, 459-466, 0007-1285 (Print)

Vera A., Moreno E., Leija L. & Vazquez M. (2007). Optimization of hydrophone centering in circular ultrasonic transducer during field characterization using edge waves: A feasibility study. *Jpn J Appl Phys 1*, 46, 7A, 4321-4323, 0021-4922

Watmough D. J., Quan K. M. & Shiran M. (1990). "The Sarvazyan method of mapping ultrasound fields," *IEE Colloquium on `Ultrasound Instrumentation' (Digest No.072)*, 4/1-4/3, London, UK, 10 May 1990, IEE

Watson T. (2008). Ultrasound in contemporary physiotherapy practice. *Ultrasonics*, 48, 4, 321-329, 0041-624X

Zeqiri B. & Bickley C. J. (2000). A new anechoic material for medical ultrasonic applications. *Ultrasound Med Biol*, 26, 3, 481-485, 0301-5629



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