Acoustical Imaging

Edited by A. J. Berkhout, J. Ridder, and L. F. van der Wal

Volume 14

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PREFACE

The representation of 'material' parameters using acoustic waves, i.e. acoustical imaging, still proves to be an interesting field of research, showing rapid development with respect to transducer technology, signal processing and imaging hardware.

The application of acoustical imaging techniques can be observed in various fields, e.g.

- low- and high-frequency seismic exploration
- seabottom investigation and underwater robotics
- medical diagnostics and tissue characterization
- acoustical microscopy and nondestructive testing

Regarding each of these different applications, it is quite remarkable that each application seems to be limited to a well-defined range of frequencies. This is explained from the fact that, given the attenuation of the medium and the desired depth of penetration, each acoustical imaging method tends to be applied within that specific range of frequencies, which produces optimum spatial resolution. However since all applications may be traced back to a joint wave-theoretical basis, quite often the differences between them can be expressed by a mere scale-factor with respect of frequency range, depth of penetration and transducer dimensions. Given the proper framework, a meeting of researchers from various acoustical imaging fields may therefore prove a very fruitful enterprise. Seismic modeling with respect to layered media may for instance also be applied within the field of nondestructive evaluation of integrated circuits, tissue characterization methods -known in medical echography- may inspire those working in the field of seabottom investigations and seismic exploration, etc.

As mentioned earlier, the achievement of optimum resolution is a common factor in most acoustical imaging techniques.

In seismic exploration the large gas and oil fields have already been discovered and the remaining small ones can only be found with high-resolution techniques. In medical diagnostics there is a large need for more detail in acoustic images and high-resolution systems are highly in demand. In the expanding field of nondestructive testing there is, apart from the detection of anomalies, an increasing interest in quantitative characterization. The latter is only feasible with high-resolution systems.

Therefore current efforts are increasingly devoted to acoustic inversion techniques, i.e. acoustic focussing, seismic migration. inverse scattering, computerized tomography and deconvolution, in order to eliminate image distortion and improve spatial resolution.

For this reason 'spatial resolution' was chosen as the key subject of the 14th International Symposium on Acoustical Imaging.

The manuscripts in this publication cover all the material presented at the symposium in both lectures and posters. We feel sure that the contents will provide useful information to those who are professionally involved with acoustical imaging.

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J. Ridder
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The 15th Symposium on Acoustical Imaging will be organized by H.W. Jones during July 14-16, 1986 at Halifax, Nova Scotia, Canada. We wish him both courage and a successful meeting.

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ULTRASONIC THERAPY AND IMAGING IN OPHTHALMOLOGY

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I. INTRODUCTION

The term "medical ultrasound" usually brings to mind the diagnostic ultrasonography systems that have become an internationally accepted modality for examining virtually all organs of the body. However, some of the earliest research with medical ultrasound was directed at studying how intense ultrasound could be used to modify tissue structures for possible therapy. For example, early work, reviewed by Kremkau (1), investigated how ultrasound might be used to treat cancer, and pioneering efforts in the laboratories of Fry, Dunn, and Lele (2,3) showed how focused ultrasound could be used to produce focal lesions for treating brain tumors and other disorders. Recently, there has been a renewed interest in therapeutic ultrasound, especially as a modality for inducing hyperthermia for cancer therapy (4).

Our interest in therapeutic ultrasound for treating diseases of the eye started in the late 1960s (5). In addition to therapeutic applications, we were motivated to examine ultrasonically induced lesions to provide background data relating to the safety of the high frequencies applied in ophthalmic diagnostic systems (6-8). Our research has continued to the present and has identified a variety of conditions that can benefit from ultrasonic therapy (7-14). After collecting animal data related to safety and efficacy, clinical trials have started in a number of areas, and equipment and procedures have been refined to support applications in the clinical setting. A key feature of our system (15-17) is the incorporation of diagnostic ultrasonography as an adjunct for aiming the treatment beam and, with new tissue characterization images (19), for planning tumor treatments and monitoring therapeutic results over time.

In ophthalmology, ablative and hyperthermia modes of applying therapeutic ultrasound are possible. In the ablative mode, a tightly focused beam is aimed at the tissue to be treated and a short (e.g., 1-sec), high-intensity (e.g., 2000 W/cm^2) exposure is used to destroy or modify tissue within a well-defined region. The ablative mode causes brief, intense heating within a tissue volume whose size is comparable to the dimensions of the focal zone of the beam. The ablative use of ultrasound is similar to that employed with therapeutic lasers. However, ultrasound does not require an optically transparent path to the target tissue, and it is not affected by optical properties that limit most laser treatments to the surface layers of tissues. In certain applications (e.g., vitreous membrane treatments (9)) the ablative mode can be employed to produce mechanical effects that are useful for therapy.

The hyperthermia mode uses broader beams, lower intensities (e.g., on the order of 1 W/cm^2), and longer exposure times (e.g., 10 min) to achieve sustained, lower-level heating (e.g., to 45 deg C). This mode is being investigated for tumor treatments in the eye (12, 13, 18) and in other organs (20, 21). We have emphasized its conjoint use with radiotherapy to treat a variety of ocular tumors: here, the synergism between hyperthermia and radiotherapy is exploited so that radiation doses can be lowered while still achieving successful tumor therapy.

Thus far, most of our clinical trials have involved treatments of glaucoma (11), a disease that elevates pressure in the eye and can thereby induce blindness. To treat glaucoma, an ablative mode is used to produce a series of lesions in the sclera and underlying ciliary body. These lesions promote the diffusion of excess aqueous humor (through the modified sclera and under the overlying, intact conjunctiva) and they also suppress aqueous humor production (in the treated ciliary body segments). The treatments have now been applied to approximately 150 patients and have proven effective in lowering intra-ocular pressure, even in cases that had proven resistant to drugs, lasers, and surgery.

In another clinical series, we have treated ocular tumors with a hyperthermia mode used with conjoint radiotherapy. Preliminary findings indicate that this approach may offer a welcome alternative to surgical removal of the eye in cases that resist conventional techniques.

This report describes portions of the research we have conducted into the nature and applications of ultrasonically induced lesions and discusses some of the design considerations that pertain to practical systems. First, it describes a therapeutic system that permits precise aiming and control of the therapeutic beam. Second, it briefly summarizes a mathematical simulation for ultrasonic treatments that guides experimental studies and treatment planning. Third, it describes a number of animal and clinical studies related to the safe and effective treatment of specific diseases.

The role of diagnostic ultrasound in practical applications of therapeutic ultrasound is also briefly outlined in the paper. We have found that high-resolution A- and B-mode techniques are important for precision lesion placement. Furthermore, new techniques in tissue characterization offer an important means of detecting subtle changes induced within treated tissues. This information can be useful for treatment monitoring and adjustment. As experience grows, this combined use of diagnostic and therapeutic ultrasound is expected to become particularly valuable.

II. THERAPEUTIC ULTRASOUND SYSTEM

The therapeutic ultrasound system (15-17) has provisions for delivering a controlled amount of ultrasonic energy to a well-defined tissue volume. The key element in the system is the therapeutic transducer assembly shown in Fig. 1. The assembly houses a therapeutic piezoelectric transducer which is a spherically curved ceramic segment; typically, the diameter of the transducer is 80 mm, its focal length is 90 mm, and it is operated at a harmonic frequency between 4 and 10 MHz. For adjusting stand-off distances, a coaxial diagnostic ultrasound transducer is situated within a central aperture in the therapeutic transducer. Additionally, for

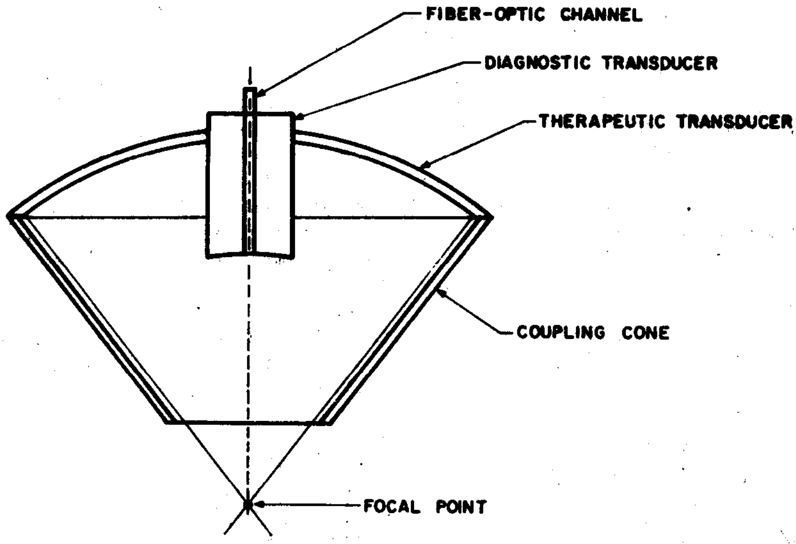

Fig. 1. Cross-section of Therapeutic Transducer Assembly.

visual indication of the treatment zone, a light spot is projected along the center of the therapeutic beam by means of a fiber-optic module contained within a central channel in the diagnostic transducer. The assembly also incorporates a conical coupling chamber that is filled with degassed, distilled water; a thin rubber membrane seals the tip of the cone. In treatment sessions, the patient lies supine and a body-temperature saline bath is configured in a steri-drape about the eye to be treated, using the same technique applied in diagnostic immersion scanning. The tip of the coupling cone is immersed in the saline bath and the transducer assembly is positioned for treatments.

The transducer assembly is aimed at the target tissue by using the projected light beam (that indicates the point of entry into the globe) and the diagnostic transducer (that measures the axial or stand-off distance to a selected tissue structure). The dimensions of the assembly components have been designed so that the focal point of the therapeutic transducer is situated 25.5 mm beyond the end of the coupling cone. The diagnostic transducer senses an echo from the membrane at the cone's tip as well as an echo from the target tissue. These echoes are processed with an A-scan biometry unit that provides a numerical display of the distance between the cone tip and a selected tissue structure. The position of the transducer assembly is adjusted until this distance is 25.5 mm so that the therapeutic beam is focused on the target tissue.

The dimensions and operating frequency of the therapeutic transducer are selected to provide beam characteristics suitable for specific applications. Frequencies above 11 MHz have been studied, with -3dB beamwidths as small as 0.15 mm. Before using the transducers for animal or human exposures, a series of calibration data are obtained. Total acoustic output is measured by radiation-force techniques. Beam profiles are measured by a pulsed reciprocity method using a small sphere mounted on a 0.005-mm diameter wire (e.g., a 0.05-mm diameter sphere may be used to profile a focused beam whose -3dB width is 0.4 mm). These data are then used to compute intensities produced by specific excitation-voltage levels. (This report specifies intensities in terms of spatial average values over the -3dB spot size of the focused beam.)