ULTRASOUND: ITS APPLICATIONS IN MEDICINE AND BIOLOGY

Edited by

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CHAPTER V

ULTRASONIC DOSIMETRY

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SECTION I

1. INTRODUCTION

The objective of this chapter is to describe the general concepts of dosimetry with specific emphasis on ultrasonic dosimetry and relate its applicability to ultrasound in medicine and biology.

Since the emphasis of this volume is on the biological and medical applications of ultrasound, the role of ultrasonic dosimetry will be presented with the view towards radiation protection. This is natural since the purpose of ultrasonic dosimetry is to relate magnitudes of specific ultrasound variables to the likelihood or extent of occurrence of identifiable biological action in the living system, usually located at the same region. This approach is in no way meant to imply that the use of ultrasound in the clinical practice of medicine represents a hazard. But it is commonly known that at sufficient levels ultrasound can produce irreversible biological damage and, as such, the potential does exist to produce damage so it is imperative to develop a common nomenclature to express meaningful radiation protection information.

With the increasing use of ultrasound comes the obligation to properly assess the risk associated with such human exposure. While no statistically based survey has been conducted to document the extent to which ultrasound is being used, a number of indicators do, however, strongly support the view that its use is increasing and that a large fraction of the human population will eventually be exposed.

A market analysis in 1969 predicted that the dollar value of the ultrasonic market would increase 300 percent during the period between 1968 and 1973.[1] This represents an annual increase of 75 percent. More recently, based upon discussions with clinical manufacturers, the U. S.

Food and Drug Administration's Bureau of Radiological Health predicted that in 1976 the industry will grow at an annual rate of 50 percent and that annual dollar sales will be around $40 million.[2]

In 1971, the Bureau of Radiological Health surveyed 301 out of 6306 short-term general hospitals in the United States and found that 12 percent of the hospitals used diagnostic ultrasound.[3] In an editorial in the Journal of Clinical Ultrasound, the editor[4] doubted the 12 percent figure since he believed Doppler ultrasound to be used by over 50 percent of the obstetricians in the United States. In 1976 the same government agency reported that a hospital survey showed 35 percent used ultrasound.[5] While questions may be raised as to the validity of the 1971 and 1976 survey estimates, they do show, however, an approximate annual increase in use of 24 percent.

An international mail survey was conducted under the sponsorship of the IEEE Group on Engineering in Medicine and Biology's Subcommittee on Ultrasound Safety and Standards and with the help of the American Institute of Ultrasound in Medicine, the Biomedical Engineering Society, the Bureau of Radiological Health, the United States Public Health Service and the United Kingdom Medical Research Council.[6] In part, the survey was to determine the extent to which ultrasound is used. On the average, between 1963 and 1971, there was an annual increase in use of clinical ultrasound of approximately 10 percent.

It was estimated that, in the United Kingdom, the number of ultrasonic diagnostic examinations is doubling every three years.[8] This represents an annual 26 percent increase.

The United States National Science Foundation, through its Office of Experimental Research and Development Incentives, conducted an international state-of-the-art survey of diagnostic ultrasound in March, 1973. One of the conclusions of the survey team was that between 1971 and 1973, the number of ultrasonic diagnostic instruments sold in the United States increased by 300 percent, which is an annual increase of 73 percent. Another

conclusion of the NSF survey team was an estimate that the sales of clinical ultrasonic devices will match those of x-ray devices by 1983.[9]

Use and trends of use information for therapeutic ultrasound is as difficult to obtain as with diagnostic ultrasound, even though it has been recognized as a potential hazard since the initial applications in the 1930's.[10,11,12] It is suggested that there is presently little change in the extent of use but information is quite limited. Three surveys have been conducted, all within the last few years, which may provide some guidance, but it is difficult to extract precise information since the surveys were limited in design.[13,14,15] The 1970 Pinellas County, Florida survey[13] indicated approximately 45,000 ultrasonic treatments per month to 6000 patients in a county population of 500,000. Assuming this is typical of the United States, and there is no supportive information for this assumption, in a population of 200 million, this would represent on an annual basis approximately 216 million treatments to 2.4 million patients.

Additionally, although there have been only a few reports of the output parameters, an approximation can be rendered as to the general range of ultrasonic power and intensity outputs from commercial diagnostic instrumentation. The measurements of pulse devices show that the spatial average ultrasonic power ranges from 60 μW to 21 mW, the spatial average, temporal peak intensity ranges from 1 to 95 W/cm^2 and the spatial peak, temporal peak intensity from 2 to 177 W/cm^2. The nominal ultrasonic frequency range is from 1 to 10 MHz with a variation of 400 to 2000 Hz in the pulse repetition frequency.[16,17,18,19] One of the pulse devices is a pulsed Doppler and exhibits a pulse repetition frequency of 48 kHz and a spatial average, temporal average intensity of 296 mW/cm^2.[19] For continuous wave, Doppler devices, the spatial average ultrasonic power ranges from 1 to 37 mW and the spatial average intensity ranges from 1 to 305 mW/cm^2.[16,18,19,20,21]

SECTION II

2. DOSIMETRIC CONCEPTS

Dosimetry is concerned with the quantitative determination of energy interaction with matter, or, in other words, defining the quantitative relationship between some physical agent and the biological effect it produces. In one sense dosimetry is the determination of a dose, or similar type of physical parameter, which characterizes the physical agent as to its potential or actual interaction with the biological material of interest.

In the case of ultrasonic dosimetry, the object is to relate magnitudes of specific parameters - such as intensity, acoustic pressure, particle displacement, etc., or perhaps some parameter yet to be developed - to the likelihood of occurrence of a biological alteration. To accomplish this, it is necessary (1) to quantify the output parameter or parameters of the source, (2) to determine the effect of the material on the propagating energy, viz., reflections, refraction, scattering, absorption, etc., and (3) to relate the first two items to quantitative parameter determination at the site of interest. To this end, Section 4 of this Chapter describes techniques by which the source output can be quantified and Chapters VI, VIII and XIII describe ultrasonic propagation properties of tissue such as absorption, velocity and impedance along with Chapter VII on scattering properties of tissue. Chapters IX, II, X and XIV provide the kinds of biophysical interaction data from which it is potentially feasible to begin developing dose-effective responses.

Thus dosimetry has two important objectives. The first is to define physical quantities which properly reflect an interaction at some site in biological material which may be expressed in units such as joules/kgm, joules/m^3, etc. The second is to develop a concept or concepts of the quantity that is applicable for radiation protection purposes.

At this point it is appropriate to briefly describe two terms which have resulted in some confusion in ionizing radiation dosimetry with the hope that such confusion will not arise or, at least, be minimal in

ultrasonic dosimetry, viz., quantity and unit. Unit represents the amount and quantity represents the thing for which the unit was devised. For example, length, time and mass are quantities for which the meter, second and kilogram are their units, respectively. This distinction is also valid for special units such as hertz (cycle per second) for which frequency is the quantity.

Typically, dose connotes something that is given or imparted in a quantitative manner. The history of other forms of radiation has documented that defining dose, or dose-like concepts, is difficult, especially when the object is to include all possible physical and biological variables. Otherwise, and the more common, special quantities are developed for the specific case or biological action under consideration. In ionizing radiation, for example, dose generally refers to the quantity absorbed dose which has been specifically defined as the energy imparted to matter by ionizing radiation per unit mass of irradiated material at the site of interest.[22] But other dose quantities have been defined for specific purposes such as genetically significant dose which is the gonad dose from medical exposure, or cumulative dose, dose equivalent, threshold dose, etc.[23] In photobiology, dose sometimes refers to the quantity dose of ultraviolet radiation which has been defined as the energy per unit surface area applied to an object.[24,25] Other quantities which have been used to characterize ultraviolet radiation were chosen to quantify a specific bioeffect. These included minimal erythema dose, minimal perceptible erythema, subvesicular dose, minimal color dose, etc.[26] There is currently much discussion regarding microwave dosimetry. Terms such as specific absorption rate,[27,28] absorbed power density and specific absorption density[29] and energy dose-rate[28] have all been either used or suggested as a basic quantity to describe absorbed electromagnetic energy. Appropriateness of units has also not been agreed upon.[30]

It is useful, for purposes of illustration, to examine in some detail the history of another form of radiation with a view towards its dosimetry development. For this purpose, ionizing radiation is chosen because it represents a well developed history.[31,32,33,34,35,36] However, this examination is not meant to indicate that ultrasonic dosimetry should take this route. The mechanisms of interaction between ultrasonic and ionizing radiation are quite different. Rather, the rationale for ionizing radiation dosimetry, as with other radiations, is one of developing an acceptable and reasonable nomenclature by which researchers in various fields can communicate and, if necessary, by which radiation protection guidelines can be developed. Knowledge of the energy disposition of the tissue site of interest is one of the critical elements in understanding the interaction between the radiation form and matter.

One of the earliest dosimetric concepts for ionizing radiation was the <u>skin unit dose</u>, more commonly known as the <u>skin erythema dose</u> or <u>threshold erythema dose</u>. One skin unit dose was the amount of radiation which just produced reddening, or erythema, of the skin. The reddening followed the exposure within a period of about one week. The detector, human skin, was very imprecise but the skin unit dose was, nevertheless, used as a basis for the first radiation protection guideline in the mid-1920's. The <u>tolerance dose</u> was suggested to be a small fraction, around one percent, of a skin unit dose, averaged over a one month period.

Not until sensitive and reproducible ionizing radiation measurement devices were developed was there a physical measurement of ionizing radiation. The concept and value of the unit <u>roentgen</u> was established in 1928 and defined in terms of the ionization, or interaction, of x-ray radiation in air.[37] It was a special unit of exposure but no specific quantity was defined at that time for which the roentgen was its unit.

In an effort to relate the <u>tolerance dose</u> to a physical parameter, radiotherapists were polled as to the number of roentgens required to produce one skin unit dose. Thus, based upon a rough value of 600 R for one skin unit dose, the tolerance dose worked out to be 6 R on a monthly basis, or 0.2 R/day. In the early to mid-1930's national and international organizations endorsed a tolerance dose of 0.2 R/day but neither gave a description of their concept of a tolerance dose. Later this value was reduced to 0.1 R/day and remained at this level for 12 years. The term tolerance dose created many problems because it was impossible to predict just what level was tolerable over a long period of time. With the realization in the mid-1930's that ionizing radiation effects may not be threshold type reactions, the term <u>maximum permissible dose</u> was substituted for tolerance dose.

As a result of biophysical and biological effect studies with various types (qualities) of ionizing radiation, it was recognized that broader dosimetric concepts were required to define and describe quantitatively ionizing radiation fields. This was especially important when applying dosimetry to radiation protection in that the roentgen was inadequate because of its limitation to x- and γ-rays and because it was not a measure of absorbed energy. In the early 1930's, it was shown that the biological effect of ionizing radiation depended not only upon the exposure intensity and time but also upon the quality of radiation since differences between x-rays and γ-rays were observed in growth reduction and mortality studies.[38] This was termed the <u>relative biological effectiveness</u> (RBE) concept and became even more important in the 1940's with the production and discovery of other ionizing radiation particles. In terms of the absorbed dose unit rad, which will be discussed shortly, this meant that the same number of rads of neutrons, for example, would produce a greater biological effect as compared to x- or γ-rays.

In the late 1930's a unit, termed the <u>energy unit</u>, was suggested as a dose of γ-rays delivered to tissue in terms of absorbed energy per gram of tissue.[39] Also, around this time, another unit, the <u>gram-roentgen</u>, was suggested[40] as the amount of energy absorbed by one gram of air when irradiated by one roentgen. In the late 1940's another unit was suggested to describe energy absorption, <u>viz</u>., the <u>rep</u> for roentgen-equivalent-physical.[41] Originally, one rep was defined as that dose of ionizing radiation which produced an energy absorption of 84 $ergs/cm^3$ in tissue. It was based upon the roentgen in that this was meant to be the energy absorbed by tissue when exposed to one roentgen. That meant that the definition depended upon a calculation of energy absorption and upon other tissue parameters which were subject, in some cases, to wide uncertainty. These difficulties were reflected in redefining one rep from the original value to other values in order to reflect actual tissue absorption properties and to remove dependence of tissue density.

The rep concept led to what is currently the quantity <u>absorbed dose</u> with the unit <u>rad</u> for radiation absorbed dose. The advantage of the unit rad over the rep is that the one rad has been arbitrarily defined as 100 ergs/gm and thus is independent of material properties.[42]

At the same time the rep was being suggested as a unit to describe dose, the unit <u>rem</u>, for roentgen-equivalent-man, was also being suggested for radiation protection purposes. The rem was defined as the product of energy absorption (in reps) and the relative biological effectiveness (dimensionless) of the energy under consideration. If there were energies of different RBE, the rem was then the sum of each respective product.[41]

In the mid-1950's, the rem concept was adopted, using the rad instead of the rep. The quantity <u>RBE dose</u> in rems was equal to the product of absorbed dose in rads and the RBE and, in the case of multiple RBE's, the sum of each product.[43]

A few years later, RBE was changed to <u>quality factor</u>, QF, and assigned fixed values which were closely representative of actual RBE's for specific conditions and energies. This was done because RBE itself was dependent upon a large number of variables and for radiation protection purposes, the quality factor values chosen were representative of RBE's. Thus, the quantity <u>dose equivalent</u> was adopted, its unit the rem, and was equal to the product of absorbed dose, quality factor and other dose modifying factors to account for spatial and temporal dose distribution.[44]

In the history of ionizing radiation dosimetry is reflected the rationale

for which national and international commissions have labored to develop concepts and define units and quantities. Initially, the threshold dose was defined as a monthly fraction of the skin unit dose and later, following proper instrumentation development, in terms of an exposure in roentgens. The term threshold dose was later called maximum permissible dose because of the realization that risk from ionizing radiation may not be represented by a threshold. Because of the desire to express the effect of ionizing radiation in terms of the interaction with or absorption by tissue at the site of interest, the absorbed dose concept was defined. Eventually the dose equivalent quantity evolved to embody both physical and biological parameters.

By comparison, the field of ultrasonic dosimetry has not developed to the extent of ionizing radiation dosimetry. The most widely used dosimetric parameter in ultrasonic bioeffect and biophysical studies is intensity in the mixed unit of W/cm^2. The principal reason for the use of intensity is, perhaps, convenience since it is an easily measured parameter as will be discussed in Section IV. As a dosimetric quantity, intensity represents many of the same problems as does the ionizing radiation quantity "exposure" in that it is not a measure of dose, or the like. Yet the majority of bioeffect and biophysical reports use intensity as the measured physical parameter of the ultrasonic field. This extensive literature documents the actions of ultrasound but, in most cases, lacks the necessary characterization of the field at the site of interest. An ideal situation would be to know the instantaneous particle velocity, the instantaneous acoustic pressure and the phase between these two field parameters at the site or sites of interest.[45]

Through both calculations[46] and experimentation[46,47,48,49] attempts have been made to determine <u>in utero</u> ultrasonic intensity in both the gravid and nongravid human uterus. A model of the tissue layers between the skin surface and fetal sac have yielded a total attenuation of 23 dB at a frequency of 2.5 MHz.[46] One of the earliest <u>in vivo</u> experiments[47] showed an average loss between the skin and uterus to be around 2.5 dB at

2.25 MHz but more recent studies have shown this to be higher, in the range from 9 to 20 dB[46,48] or from 6 to 14 dB.[49]

There have been ultrasonic dosimetric quantities which are noteworthy of comment in that they represent, in concept, the basic approach to dosimetry. The cataract-producing unit, CPU, was a quantity defined as the length of exposure necessary to produce a grossly observable cataract and expressed in units of seconds.[50] The dosimetric concept damage ability index with the unit $second^{-1}$ is a quantity intended to describe the effect of ultrasound on spinal cord hemorrhage.[51]

Quite recently it has been suggested[52] that a universal dosimetric response to ultrasonic exposure may exist for different tissues but the response has only been demonstrated, in a limited manner, in mammalian brain tissue. The response is in terms of the "energy absorbed per unit volume" (J/mm^3) for histologically observable lesions at superthreshold levels[53] as a function of the "delivered intensity." It is shown that at two different ultrasonic frequencies, 3 and 4 MHz, identical constant volume curves result even though there are two different "threshold levels."[53]

A step or two removed from knowledge of the in situ ultrasonic field parameters, but yet an essential part of field characterization, is a thorough characterization of the ultrasonic field at the site where the specimen is placed but without the specimen in place. This provides a basis from which in situ field parameters can be calculated and a basis whereby duplication of the experimental arrangement is made possible. As will be discussed, there are techniques available to measure many of the ultrasonic field parameters when the medium of interest is macroscopic, homogeneous, isotropic and low-loss, such as water. But this does not hold for in vivo measurements and, as a result, both instrumentation and dosimetric concepts need to be developed.

To properly develop ultrasonic dosimetric concepts, a much greater understanding of the processes by which ultrasonic energy interacts with biological material is required. This point can be illustrated by a number

of examples. Whenever ultrasonic energy is absorbed by any biological material, heat results. In fact, biological tissues absorb ultrasound at a relatively high rate (see Chapter VI). If temperature rise, or for that matter the rate of temperature rise, is the response under study, then the amount and rate of ultrasonic energy being absorbed per unit volume or mass will be important. To determine this, the spatial and temporal intensity, the exposure time and the material absorption properties will be required to calculate the absorbed energy. Additionally, geometric beam considerations affect the temperature response since the initial rate of rise of temperature has been shown to be quite similar for both plane and focused ultrasonic beams but diffusion properties of the tissue markedly affect the temporal temperature development.[54,55] The importance of thermal diffusion has also been shown to explain the apparent frequency independence of histological lesions in mammalian central nervous system tissue under focused ultrasonic conditions.[56] If intensity is to be the ultrasonic quantity at the site of interest, then the specific heat will have to be considered in order to calculate the temperature response.[57,58] Thus, to know the intensity in situ and, in most cases where the irradiation would be in the near field, would require detailed information of the acoustic pressure and particle velocity.

If cavitation, as another example, is the mechanism of action, then the acoustic pressure may be the primary physical parameter which should be specified, or, at least, the parameter from which a dose-related term is derived. An approach to develop a dosimetric concept for considering cavitation as the causative mechanism could begin by defining the parameters responsible for the onset of bubble activity in tissue (while this is an extremely difficult and, perhaps, not yet an achievable goal, the approach is what is being emphasized). These parameters would include, but are not necessarily limited to, the ultrasonic field quantities, such as acoustic pressure frequency and amplitude, the environmental conditions such as ambient pressure and the tissue properties such as nuclei sites, degree and state of gas, etc. Following the formation of the bubble(s), the rate of growth could be considered. If the ultrasonic forces were insufficient to cause the bubble to grow to the "dynamic threshold radius,"[59,60] the bubble would remain stable and radially oscillate. This has been termed stable cavitation.[61] Under this condition, a relatively small amount of energy is extracted from the ultrasonic wave but relatively high and localized energies are established in the bubble vicinity as it functions as a secondary source. Examples of the types of resultant forces from such oscillatory bubbles can be found in the literature[62,63,64,65] and in this Volume. Should the bubble grow greater than the dynamic threshold radius, transient or collapse type cavitation would result, causing extremely high energy densities within its immediate vicinity with the certainty of adversely affecting biological material. The temporal exposure conditions would be an essential parameter to consider in view of the fact that an exposure time appears to be necessary to elicit a biological effect.[66,67] This, along with other biological observations[53,68,69] resulting from cavitation-type phenomenon have all been reported in terms of ultrasonic intensity.

Other phenomena which need to be at least considered in any ultrasonic dosimetric approach, especially when aimed at radiation protection, includes the question of whether or not ultrasonic biological actions are cumulative, the role of synergism, frequency dependence of an effect, critical organ or tissue concept and, perhaps, others. Although ultrasonic energy does not have an analogy to ionizing radiation's "quality of radiation," the relative biological effectiveness represents an important radiology concept and thus should be kept in mind.

There appears to be, at least, a reasonable doubt whether or not cumulative effects occur from exposure to ultrasound. Summation of subparalytic doses of ultrasound, with sufficient time for temperature equilibrium to be re-established between pulses, produced paralysis in frog hind limbs.[70] It has also been demonstrated that, under pulsed ultrasonic exposure conditions, by varying only the duty cycle with a constant pulse width, spinal

cord hemorrhage occurred only when the total sum of on-time of pulses reaches the same value.[51]

Synergism between ultrasound and other agents should properly be considered for any ultrasonic dosimetric concept. There have been both positive[71] and negative[72] synergistic findings with ionizing radiation and positive findings with both hypoxia[51] and one chemotherapy drug.[73]

Structural lesions in mammalian adult brain were initially thought to be frequency independent over the range from 1 to 9 MHz[53,56,69] but after further examination of the data, a slight but oscillatory frequency dependence was shown.[74] The mechanism responsible for this dependence has been identified as a capsular layer surrounding the brain.[75] It is interesting to observe that most organs possess some type of capsular layer and thus its dependence upon ultrasonic energy transmission should be considered.[76] By compensating for the capsular layer effect, the intensity threshold responsible for lesion production appears to be frequency independent.

Other types of frequency dependent examples, all based upon intensity as the ultrasonic parameter, include greater damage to liver at lower frequencies over the range 0.5 to 6 MHz,[51,77] greater change in the electrophoretic mobility of irradiated cells at lower frequencies over the range 0.5 to 3.2 MHz[78] and greater susceptibility to the production of cataracts at higher frequencies over the range 5 to 15 MHz.[79]

SECTION III

3. MEASUREMENT CONSIDERATIONS

3.1 <u>The Non-Idealized Field</u>

The typical ultrasonic use of diagnostic ultrasound in the clinical practice of medicine suggests that the source is circular in shape and exhibits a uniform velocity distribution of simple harmonic motion.[80,81,82,83,84,85] Most standard texts[86,87,88] describe the resultant ultrasonic field from the plane piston source. Generally, it is divided into two regions, the near or Fresnel region and the far or Fraunhofer region. This is an idealized model, one which is highly convenient for instructional purposes but one which may not in fact describe the field of real transducer sources.

It is, at least, prudent to make a few observations regarding the ultrasonic field from a flat, circular piston source operating with a uniform velocity distribution. It is known that as the mechanical back loading of a ceramic crystal and quartz plate is increased, and back loading is widely used for diagnostic transducers, the surface displacement becomes more uniform and, hence more piston-like.[89,90,91] The dividing point between the near and far field is commonly taken at the last axial maximum which occurs at a distance of a^2/λ from the transducer surface (a is the transducer radius and λ is the acoustic wavelength).[82,88] Other distances have also been used to designate this dividing point such as $2a^2/\lambda$ when the axial intensity just begins to behave as r^{-2} (r is the distance from the transducer surface)[87] or $0.75\ a^2/\lambda$ where the beam width is the narrowest.[92]

Even with the idealized piston source, the near field is extremely complex. Numerical procedures have been employed to construct detailed perspective images of the near field and demonstrate quite a complicated field parameter distribution.[92] Also, it has been demonstrated that a reactive component of axial intensity exists near the transducer surface.[93] Thus, even with the idealized piston source, plane wave assumptions are not valid in the near field.

Non-uniform velocity distribution on the transducer surface markedly affects the propagated field distribution pattern.[94,95,96] For example, a Gaussian velocity distribution on the transducer surface results in an axial intensity distribution free of oscillations and a directivity pattern Gaussian in shape and thus free of side lobes.[94] Clamped and simply supported boundary conditions of a circular transducer result in axial intensity distributions more typical of the Gaussian case but with small

oscillations, the clamped case exhibiting oscillations of a lesser amplitude than the simply supported case.[96] Also, the clamped and simply supported disks have a greater divergence than that of the piston source.[96]

At the spatial peak, temporal peak intensity levels at which some of the ultrasonic diagnostic instrumentation operates, the effect of finite amplitude effects upon the measurement process must be considered. Finite amplitude effects simply mean that a fraction of the propagated ultrasonic energy is transferred to its higher harmonics as a function of distance.[97,98,99] For example, at an ultrasonic frequency of 2.5 MHz and with an initial intensity of 3 W/cm^2 at the source, as the energy propagates away from the source in water 4 percent of the energy has been transferred into the second harmonic, 5 MHz, at 10 cm and 12 percent at 20 cm. Also about 4 percent has been transferred into the third harmonic, 7.5 MHz, at 20 cm. With an initial intensity of approximately 12 W/cm^2, 12 percent of the energy is transferred into the second harmonic at 10 cm from the source.[97]

On the other hand, in Chapter IX of this Volume, at extremely high intensities, it is argued that finite wave effects are negligible under certain experimental conditions.

3.2 Absolute and Relative Measurement Process

There does not appear to be a standard definition as to what is an absolute measurement process for ultrasonic field parameter determination. So a working definition which will be employed here is one in which the measurement process does not need to be calibrated in a known field in order to yield quantitative information about the ultrasonic field parameter which is being measured. Therefore, a relative measurement process must be one which requires calibration in a known field. It should be noted that the total measurement process is either absolute or relative, not the measurement itself.

The piezoelectric technique developed at the National Bureau of Standards[100,101] and described in the next section under Piezoelectric Techniques qualifies as a National Bureau of Standards reference standard for total ultrasonic power. The National Bureau of Standards is also developing a calorimetric method for determining total ultrasonic power[102] but no description is currently available. This may also be available as an NBS reference standard.

3.3 Accuracy and Precision

The process of taking and analyzing measurements requires both human skills and judgment along with instrumentation which is subject to uncertainties. As such the reporting of analyzed measurements are commonly represented by numerical qualifiers which are intended to convey a degree of confidence. One of the more typical statements is "the accuracy of... is ___ percent." It has been suggested[103] that such a quasi-absolute statement represents poor practice and should be avoided. An understanding of this statement is embraced in the terms error, true value, systematic error, precision and accuracy.

Error is the difference between the value obtained and the true value. In order to quantify the error of a measurement, it is necessary to first quantify the true value. In concept, true value seems to be one of those terms which is implicitly understood but in practice it may never be exactly determined. For example, in measuring, say, ultrasonic intensity, the system will yield a number. But to what is it compared for the determination of the error. If the United States National Bureau of Standards (or some other standards organization) had an ultrasonic intensity standard, the measured value would be compared to it, and this would be explicitly stated in reporting the measurement, but there would be no guarantee that the NBS standard truly represented the true value.

Whereas error can be defined in terms of a measurement, precision and accuracy are defined in terms of a measurement process, that is a method of measurement which can be described in statistical terms or, as Eisenhart[104] states, has "attained a state of statistical control." Precision has been defined as "... the typical closeness together of successive independent measurements of a single magnitude generated by repeated ap-

plications of the process under specified conditions."[103] High precision is indicative of a measurement process which exhibits small experimental or random errors.[105] Accuracy, on the other hand, has been defined as "... the closeness to the true value characteristic of such measurements,"[103] and a highly accurate measurement process is indicative of a small systematic error, or bias.[105] In other words, precision is concerned with how close together the measurements are to each other, irrespective of how close the measurements are to the true value, and accuracy is concerned with how close the measurements are to the true value.

Now it is, perhaps, clearer why specifying the accuracy of something in terms of some percentage represents poor practice. How, in fact, do you specify the accuracy unless you know the true value. It is, of course, possible to indicate all possible sources of systematic errors, and make an informed judgment as to the degree of bias. Systematic errors are those which introduce a constant bias and are affected by such items as an error in instrumentation calibration, a flaw in the fundamental theory being imposed, improper use of an instrument, observer error resulting from habit, etc. If the measurement process is calibrated against a reference standard, such as maintained by the National Bureau of Standards, then it is recommended that the systematic error be considered negligible and that an explicit statement to this effect be indicated when reporting the results.[103]

Precision, or more properly, imprecision, can be numerically qualified through statistical tests, that is the precision of a measurement process can be known and measured. This is accomplished by computing and reporting quantities such as the standard deviation.

A much more comprehensive discussion of those concepts can be found in the literature.[103,104,105] Specific recommendations are provided by Eisenhart[103] for reporting experimental uncertainties for each of the four possible situations of whether or not systematic error or imprecision are negligible.

SECTION IV

4. MEASUREMENT TECHNIQUES

4.1 Capacitance Probe

When an ultrasonic wave is normally incident upon a boundary and the characteristic acoustic impedance mismatch is sufficient to assume total reflection such as with a solid-air interface, the boundary displacement is twice that of the wave's particle displacement. This is the principle behind the capacitance probe wherein one surface of a capacitor is free and the other fixed such that when an ultrasonic wave is incident at the free surface boundary, the capacitor thickness is modulated. One of the first applications of the capacitance probe was the absolute determination of the fundamental and second harmonics in distorted ultrasonic waves to study third order elastic constants.[106,107,108] This principle has more recently been applied to the absolute determination of the instantaneous particle displacement of plane[17,109,110,111] and focused[112] ultrasonic pulses in liquids. The capacitance probe is very desirable for measuring the ultrasonic output from clinical pulse-echo diagnostic equipment in that it represents one of two techniques (the other is the electrodynamic probe) which is absolute, operates over a wide frequency range and yields the response of pulses. The major disadvantage is that frequency compensation is required to convert the instantaneous particle displacement to intensity. The maximum width of the pulse is a function of the distance the pulse traverses in a buffer rod which provides the free capacitor surface. The buffer rod also introduces an impedance mismatch with the liquid, and thus must be acoustically characterized to determine the fraction of signal transmitted into it.

Filipczynski[110] used a gap thickness of 49 μm, an aluminum buffer rod and a polarization voltage across the gap of 208 volts and obtained a free surface displacement of 6.4 Å for a temporal peak intensity in the liquid of 76 mW/cm^2.

4.2 Chemical

The ultrasonic vibration potential results by virtue of the interaction

between the ultrasound and ions in solution. As a compressional or vibrational wave is propagated in an ionic fluid, the differing mobilities between the cations and anions cause a local electrical deviation from neutrality. Debye[113] developed the first theory which predicted and described the phenomenon. Later[114] a more exact form of the Debye effect equation was developed. Some sixteen years after Debye first predicted the effect, the suggestion was put forth that it may prove useful and practical as a means for determining absolute ultrasonic intensity.[115] Around that same time the Debye effect was first experimentally observed.[116] Later, vibration potentials were reported in pure water[117] and in pure organic liquids[118] but these findings, which contradicted theory, were attributed to the probe design.[119]

While the mechanism responsible for the velocity potential has not been precisely defined (so that this technique can be used as a primary method), there appears to be no disagreement that the potential is directly proportional to the amplitude particle velocity.[117,118,119,120,121,122,123] Specifically, in a moderately dilute solution where the diffusion forces are negligible, the ultrasonic vibration potential Φ_o of the $C^{Z+}A^{Z-}$ electrolytic is given by[122]

$$\Phi_o = 0.155\, a_o \left[\frac{W_+}{Z_+} t_+ - \frac{W_-}{Z_-} t_- \right]$$

where Φ_o is the vibration potential in microvolts, a_o is the amplitude particle velocity in cm/sec and W, Z, and t are, respectively, the apparent molar mass, the number of unit charges and the transport number for cations (+) and anions (-). In a 0.1 M solution of NaCl, for example, at an ultrasonic frequency of 220 kHz, an amplitude particle velocity of 1 cm/sec yields a potential of 1 μV.[122] Assuming idealized plane wave propagation, this corresponds to an ultrasonic intensity around 86 mW/cm^2.

4.3 Electret Transducer

An electret is a material, or dielectric body, which possesses separate electric poles of opposite sign of a permanent or semi-permanent nature. At the macroscopic level, it can be considered the electrostatic analogue to a permanent magnet.[124] Electret transducers, and specifically the foil-electret transducer, have been employed over the extensive frequency range from 10^{-3} to 2×10^8 Hz.[125,126,127]

The applicability of foil-electret transducers to megahertz ultrasound in liquids has been demonstrated,[128,129] wherein an N x N array of electret microphones sampled the ultrasonic field. Two different two-dimensional arrays of 16 x 16 were designed, one which consisted of a foil electret and a 16 x 16 element backplate and the other which consisted of the foil electret and backplate each with 16 stripes in an overlapping matrix fashion. In the former design 16^2, or 256, switches are required and in the latter 2 x 16, or 32, switches are required to sample each electret receiver. When the ultrasonic signal is incident upon the array, each element yields a voltage proportional to the instantaneous acoustic pressure. The useable frequency range of these arrays is from 0.3 to 2 MHz. The extension to arrays of 200 x 200 are suggested as feasible.

4.4 Electrodynamic Probe

The electrodynamic probe is similar in concept to the air loud-speaker wherein the incident sound wave on the diaphram moves the coil positioned in a permanent magnetic field and thus generates a voltage. For this probe wires are wound around an insulator, and the surface which the ultrasound is incident upon is positioned in a permanent magnetic field with the wires normal to the flux lines. The ultrasound vibrates the windings yielding a voltage directly proportional to the instantaneous particle velocity.

Filipczynski[130] was the first, and perhaps still the only one, to apply this absolute method to the measurement of ultrasonic pulses from clinical diagnostic equipment. The electrodynamic probe has been developed for the measurement of both plane[17,110,130] and focused[112] ultrasonic pulses. For plane waves, as an example, Filipczynski[110] used an effective length of 2 μm aluminum wire of 8.0 cm in a magnetic field of 4200 Gauss and yielded

a particle velocity of 3 cm/sec which is equivalent to a pulse intensity of 58 mW/cm^2.

4.5 Optical Techniques

The principle advantage of the acousto-optic interaction techniques for the measurement of acoustic pressure is the avoidance of placing any measuring device in the sound field. It is assumed that the interaction between light and sound perturbs the sound field negligibly. Brillouin[131,132] first predicted that elastic waves in liquids and solids diffracted light. The phenomenon was experimentally verified by Debye and Sears[133] and by Lucas and Biquard.[134] Bär[135] conducted rather extensive experimentation which, in part, led to a fundamental theory for describing the distribution of light in the different diffracted orders.[136,137,138,139,140,141] Thus Brillouin scattering, Debye-Sears effect, Raman-Nath diffraction or scattering and Bragg diffraction have all been terms used to describe the general phenomenon. Further discussion of the early work can be found in references 142,143,144.

The diffraction problem, as it relates more directly to the interaction phenomenon between laser light and megahertz ultrasound has been reduced to a set of coupled difference-differential equations.[147] Subsequently, Klein and Cook[148] numerically examined the two limiting cases termed the Raman-Nath and Bragg regions and also the intermediate region with the view towards establishing criteria for quantitative measurements of acoustic pressure amplitude. The three regions, in fact, are basically determined by the parameter Q which is governed completely by the geometry of the experiment. Mathematically,

$$Q = \frac{k^* L}{\mu_o k}$$

where k^* and k are the respective wave numbers for sound and light, μ_o is the undisturbed optical index of refraction for the material in which the sound is propagating and L is the distance over which the light interacts with the sound, the interaction length.

In the Raman-Nath region, where $Q << 1$, the ultrasound beam is typically narrow and the frequency is low. Under this condition, when the light beam enters normal to the sound beam, the parallel acoustic wavefronts function equivalently to an optical grating by virtue of the sound field modulating the material's index of refraction. The resulting Fraunhofer diffraction pattern is symmetrical and equally spaced about the primary (zeroth order) light beam. The normalized nth order light intensity, I_n, is described by

$$I_n = J_n^2 \ (v)$$

where J_n is the nth-order cylindrical Bessel function and v is the Raman-Nath parameter, which can be determined uniquely by measuring three adjacent diffraction orders.[149,150,151] The acoustic pressure amplitude can be determined from

$$p = \frac{v}{kL\left(\frac{\partial \mu}{\partial p}\right)_s}$$

where $\left(\frac{\partial \mu}{\partial p}\right)_s$ is the adiabatic piezo-optic coefficient. Raman and Ventakaraman[152] and Riley and Klein[153] have determined values for the adiabatic piezo-optic coefficient for different materials, including water.

This procedure has been suggested in order to obtain the acoustic pressure amplitude.[151,154,155,156] However, in order for this technique to be absolute, the interaction length must be known. Breazeale and Hiedemann[154] used a schlieren technique.[157,158,159] Cook[160,161] has shown, however, that transducer orientation is critical in a non-uniform field. A principle assumption is plane wave propagation. In the Fresnel region of the ultrasonic beam, the interaction length may be assumed to be equivalent to the active surface of the transducer. But, theoretical analyses between the Raman-Nath parameter and the strength of the effective optical phase grating have been reported for circular[162] and square[162,163,164] transducers and indicate problems associated with determining the Raman-Nath parameter by assuming a transducer configuration.

Haran and colleagues[151] have described an experimental procedure by which the total acoustic power can be obtained without the need to make assumptions about the interaction length.

The intermediate and Bragg diffraction regions[148] have not been utilized for acoustic field measurement owing to the fact that the predominant clinical ultrasonic instrumentation operates at frequencies and possesses beam diameters which result in low values of Q.

An imaging broadening method for the determination of acoustic pressure amplitude was suggested by Hueter and Pohlman[165] based upon work by Lucas.[166] Also, a decrease in the zeroth order light intensity method was developed by Loeber and Hiedemann.[167] Both of these techniques, along with the above Raman-Nath region technique were used a number of years ago but little has been done with these two procedures since that time.[154,155]

The above acousto-optic methods are applicable in determining the acoustic pressure amplitude and continuous wave conditions. Newman[159] describes a schlieren technique by which an image of individual acoustic wavefronts can be obtained. Cook and Berlinghiere[160,161,168,169] have described a method by which a cross-sectional map of the acoustic pressure amplitude and phase can be obtained from a progressive sound wave. The procedure involves measuring the optical phase retardation of a light beam which traverses the sound beam laterally as the sound source, in effect, is rotated about its axis. They have shown[169] that the data requirements are not too excessive to yield an appropriate cross-sectional map. Cook[170,171] further describes a method by which the optical retardation, both amplitude and relative phase, can be determined and thus yield absolutely the Raman-Nath parameter. This latter procedure is applicable for both continuous wave and pulsed ultrasonic waves and is capable of yielding average and peak acoustic pressure.

An acousto-optic technique which requires a "probe" in the field and yields the absolute spatial particle displacement amplitude is based upon the Michelson interferometer whereby a thin, flexible membrane (metalized plastic film) is placed in the sound field and exhibits a particle displacement of the sound wave.[172,173] The interferometric arrangement is designed to detect the amplitude displacement of the membrane as a function of position on the membrane, hence yielding the spatial distribution of the particle displacement amplitude of ultrasonic pulses. The reported sensitivity of this technique is approximately 0.1 Å which is comparable to an intensity of 1 $\mu W/cm^2$ at 1.5 MHz in water.

4.6 Piezoelectric Techniques

Piezoelectric techniques are generally applied to measure the instantaneous acoustic pressure. They are, perhaps, the oldest techniques employed to measure underwater sound, following the discovery of the piezoelectric effect in 1880.[174,175] A most interesting and readable history of its discovery and development has been prepared by Cady[176] and Hunt.[177] In this section, the term piezoelectric techniques refers to all the associated phenomena which embody the transduction of the instantaneous acoustic pressure to an electrical signal in the megahertz frequency range. This includes not only piezoelectricity but also magnetostriction, ferroelectricity, etc. There are a number of excellent books[86,176,177,178,179] and monographs[180,181,182] which aptly describe these phenomena and their association with transduction.

One class of piezoelectric materials not included are the polymer films which have yet to be successfully applied to megahertz ultrasound. Yet they appear to have a potential application and thus are mentioned for completeness.[183,184,185,186]

Brendel[187] gives a brief summary of the basic requirements for hydrophones used to measure clinical ultrasonic instrumentation. These include being (1) temporally stable, (2) small relative to the wavelength, (3) adequately sensitive, and (4) broad-banded. While no one hydrophone satisfies all these criteria,[188] reasonable attempts have been made to construct hydrophones suitable for diagnostic ultrasound measurements.[187,189,190,191,192,193] Most were designed so that calibration

could be accomplished with reciprocity. Only three have quantified the hydrophone response with sensitivities of 0.004 - 0.007 $\mu V/bar$ (0.04 - 0.07 $\times 10^{-6}$ $\mu V/Pa$),[189] 0.2 mV/Wcm^{-2} (1.2 $\times 10^{-3}$ $\mu V/Pa$ at 1 W cm^{-2})[192] and -109 db re 1 V/Pa (3.6 $\mu V/Pa$)[193] over the approximate bandwidth of 1-10 MHz with approximate transducer surface areas, respectively, of 0.0003, 0.17 and 38 mm^2.

The application of piezoelectric techniques are well suited for mapping the acoustic pressure distribution. In one report[194] a PZT-5H ceramic cylinder, 1.6 mm by 1.6 mm by 0.25 mm, positioned and attached coaxially on the end of a 1.6 mm diameter stainless steel tube, was used in an automated measuring system to obtain field plots of the acoustic pressure. The hydrophone exhibited an approximate sensitivity of 2.8 $\mu V/Pa$ at 1 MHz.[195] Another piezoelectric probe, constructed with the same type of ceramic cylinder in basically the same fashion exhibited a sensitivity of approximately 10 $\mu V/Pa$.[190]

Two other experimental procedures are described to map out the ultrasonic field distribution with piezoelectric probes but little detail is provided regarding the probe calibration.[196,197] However, both do yield detailed and highly descriptive perspective maps of the ultrasonic field distribution.

The Underwater Sound Reference Division (USRD) of the Naval Research Laboratory USRD type E8 transducer is a calibrated ultrasonic transducer over the frequency range from 150 kHz to 2 MHz.[198] It is used as a projector, hydrophone and reciprocal transducer and is available on a rental basis.[199] It has a 2 cm diameter piezoelectric disk with a coupling media of castor oil between the disk and rubber window. It is designed to operate over the temperature range from 5 to 35°C which may represent some problems. The beam pattern closely follows the theoretical shape in the megahertz frequency range.

The National Bureau of Standards has developed a standard method to determine ultrasonic power[101] and is now offering a calibration service.[100] The technique employs a quartz transducer, operating at its fundamental frequency as a plane piston source. The total ultrasonic power is determined by the product of the applied voltage squared and the resonant radiation conductance, viz., $P = V^2 G_r$. The resonant radiation conduction is determined by measuring the input admittance of the transducer at resonance both in water (loaded resonant condition) and in air (unloaded resonant condition) and at twice the resonant frequency in air (clamped condition), thus yielding, respectively, G, G' and G" from which

$$G_r = \frac{(G' - G)\ (G - G'')}{(G' - G'')}$$

Because of the difficulty in determining G and G' directly, an experimental procedure has been established, using sophisticated NBS equipment, such that G_r can be determined within an uncertainty of about ± 2%. Typical value of the resonant radiation conductance for an air-backed, 2 MHz, 6.35 mm radius active surface quartz transducer radiating into water is 4.92 microsiemens. Taking into account additional uncertainties, including about a ± 1% in the applied voltage, the total radiated power can be known to within ± 5%.[101]

4.7 Radiation Force

Radiation force techniques are very desirable methods for determining ultrasonic power and spatially averaged intensity under both continuous wave and pulse conditions in that they are absolute, capable of sensitivities into the microwatt range, relatively simple in design, and generally do not require frequency compensation. When an object is placed in an ultrasonic beam, a force is exerted on the object by virtue of the time-independent pressure, called radiation pressure. This force is a direct result of the transport of energy by the ultrasonic wave and is equal to the time rate of change of momentum per unit area and, thus, is directly related to ultrasonic intensity and power. Chapter I provides mathematical and theoretical details of the phenomenon.

There have been two basic types of objects, or targets, used to measure radiation force, one which intercepts the entire beam and one which

intercepts only part of the beam, thus yielding, respectively, the total ultrasonic power and spatially averaged intensity. In the case where the target intercepts the entire beam, is normal to the beam and is a perfect absorber (that is, all the energy incident upon the target is dissipated into it), a one milliwatt ultrasonic beam exerts a force equivalent to a weight of 67 micrograms upon the target, should the target be in a medium such as water with a speed of sound of 1500 m/s. Rooney[20] utilized a perfect absorber target at normal incidence in conjunction with a sensitive electrobalance to measure total power. The sensitivity of the balance was 0.1 micrograms, which is equivalent to 1.5 microwatts, but the system's reported minimum capability was 30 microwatts with a 5 microwatt standard deviation. Kossoff[200] earlier used a modified analytic balance with a sensitivity of 10 micrograms (0.15 milliwatts) and a totally absorbing target at normal incidence to the beam and achieved a minimum detectability of 80 microwatts. Kossoff[201] cautioned that an absorbing target is subject to Archimedes upthrust force because target temperature could increase due to energy absorption. At normal incidence, it should also be noted that if the target is not a "perfect" absorber, a standing wave develops between target and source which would affect the results.

To minimize thermal problems, a perfect reflecting target can be used. The target surface is typically positioned at an angle of 45° with respect to the beam which redirects the energy perpendicular to the beam and into an absorbing and/or scattering material. Under these conditions in water, the force exerted by a one milliwatt ultrasonic beam is equivalent to a weight of $134 \cos^2 \theta$ micrograms where θ is the angle of incidence between the beam direction and the normal of the target surface, which is 67 microgram at 45°. Kossoff[200] used the same balance for a totally reflecting target as with absorbing target reported above and achieved a minimum detectability of 40 microwatts. Hill[190] described the construction of a balance system which was a modification of Kossoff's[200] design. This design exhibited a precision (variance) of ± 300 microwatts with a sensitivity of 300 microwatts. A less sensitive balance system was also described for application at therapeutic power levels. Wells and coworkers[202] described two radiation force balances and compared them at the one watt level to a calorimeter. Agreement was reported to be within 6 percent. The torsion balance principle has been utilized to measure radiation force incident upon a reflecting target.[203,204,205] The minimum detectability reported has been 30 microwatts. Zeidonis[206] also reported on the use of an analytic balance to measure ultrasonic power and obtained a sensitivity of 10 microwatts.

Another experimental set-up to measure ultrasonic power with a totally reflecting target has not yet shown itself sensitive enough to measure diagnostic levels. The bouyant float technique, first described by Oberst and Richmann,[207] is used to measure the radiation force of a downwardly directed ultrasonic beam by measuring the displacement of a float bouyant between two liquids, say water and a denser material such as carbon tetrachloride. The float target is an inverted conically shaped surface which permits the float to be self centering when the ultrasonic power is incident upon the target. The sensitivity of the technique depends predominately upon the diameter of the float stem which traverses both liquids and density difference between the two liquids. This technique has mainly been used to measure ultrasonic powers in the high milliwatt to tens of watts range[15,194,208] although Reid[209] has suggested that this technique may serve as a field standard and indicated that a sensitivity of 1 milliwatt per millimeter of float displacement may be achievable.

The horizontal deflection of a target has been used to measure both power and intensity. Wells and colleagues[210] suspended a totally reflecting target at an angle of 45° with respect to the direction of the sound beam and measured its deflection. In this arrangement, a power of one milliwatt caused a horizontal deflection of 200 micrometers. The minimum detectability of this arrangement was 10 microwatts. Using the deflection of a spherical target mounted on a bifilar suspension to measure radiation force

was suggested as early as 1949 by Fox and Griffing.[211] The contemporary techniques generally consist of a steel ball connected to a filament (either with glue or by threading the filament through the finely drilled hole in the ball) and suspended vertically into a liquid such as water. The horizontal displacement of the suspended ball is determined for small angular deflections and yields the radiation force

$$F_r = \frac{d m_b g}{l}$$

where m_b is the bouyant mass of the sphere, g is the gravitational constant and l is the suspension length.[194,212,213] The relationship between the radiation force and the ultrasonic intensity is given by

$$I = \frac{F_r c}{\pi a^2 Y}$$

where c is the speed of sound in the fluid, a is the ball radius and Y is the "acoustic radiation force function"[214,215,216] which depends upon the elastic properties of the ball, the density of the fluid, ball size, and acoustic frequency. The suspended ball technique has not been applied to measuring clinical instrumentation because it typically is not sensitive enough. For example, if a stainless steel 440C ball, radius of 0.794 mm, has a suspension length of 11 cm, a 1 mm deflection is equivalent to 111 mW/cm^2.[194] A comparison of this technique with that of two other measurement techniques, optical and thermal, in 1958 showed agreement to within ± 10 percent.[217] Also, a recent evaluation of the suspended ball technique using elastic spheres showed that by comparing the acoustic radiation force function to experimental data, the ultrasonic intensity can be determined with an accuracy of about ± 3 percent.[218]

Zieniuk and Chivers[219] in a reasonably complete review have compared the radiation force and thermal techniques for the measurement of ultrasonic power and intensity.

Experimentally, both acoustic streaming and radiation force effects occur simultaneously.[220,221] As a result, Sokollu[221] argues that since the two phenomena cannot be differentiated, the radiation force techniques have a built-in bias or systematic error. There is no effective procedure to separate out the acoustic streaming except with the insertion of a membrane which, ideally, would serve as a barrier to the flow and permit the sound wave to be transmitted undisturbed. Rooney,[20] for example, used a 25 μm thick stretched membrane to shield the target from acoustic streaming.

The phenomenon of acoustic streaming, on the other hand, can be utilized to measure the absolute ultrasonic energy density.[220,222] Here an experimental arrangement permits the measurements of acoustic streaming flow rate and with known vessel dimensions and acoustical properties of fluid the energy density can be evaluated.

4.8 Reciprocity

On their own MacLean[223] and Cook[224] applied the principle of reciprocity[225] to the development of a method for absolute calibration of electroacoustic transducers. The complete range of reciprocity techniques are extensively covered by Hunt,[177] Ackerman and Holak[226] and Bobber[198] as they predominately apply to transducer calibration at frequencies typically lower than those employed in medicine; however, the principles are the same. To apply conventional reciprocity to a single absolute calibration, three transducers are required and three sets of voltage measurements must be made. The three transducers are a projector (source, loudspeaker, etc.), a hydrophone (receiver, microphone, etc.) and a linear, passive, reversible electroacoustic transducer. The latter transducer is reciprocal, viz., the ratio of its receiving sensitivity to its transmitting response must be equal to a constant, the reciprocity parameter. This parameter is a function of the transmission medium, the acoustic frequency, and boundary conditions. Bobber[227] derived a unifying concept for all reciprocity parameters in terms of a general reciprocity parameter. The only requirement was that the medium satisfy the acoustical theorem. The reciprocity parameter for conventional (spherical-wave) reciprocity is[87]

$$J = \frac{2d\lambda}{\rho c}$$

where d is the distance between transducers, λ is the acoustic wavelength and ρc is the characteristic acoustic impedance of the medium.

The two-transducer reciprocity techniques, a special case of conventional reciprocity, require that both the hydrophone and reciprocal transducer have the same sensitivity which, without a third transducer, cannot be verified. Carstensen[228] devised a self reciprocity technique in which both the hydrophone and reciprocal transducer exhibited the same sensitivity. This was accomplished by reflecting the transmitted signal off a perfect reflector and back into the same transducer. It is possible to obtain an absolute calibration on a single transducer without the need for additional transducers. Koppelmann and colleagues[229] and Reid[230] applied the self-reciprocity technique to calibrate transducers. In the former, a two transducer procedure was used wherein one was calibrated via the self-reciprocity technique and then the unknown transducer was calibrated in the known sound field. Reid[230] developed the theory for and described the use of the self-reciprocity technique for calibrating medical pulse-echo transducers. He developed theoretical expressions for both near and far field conditions under both continuous wave and pulsed regimes.

4.9 Thermal Techniques

Both transient and steady state thermal techniques have been employed as primary methods for the measurement of ultrasonic intensity and power. It has been pointed out[221] that thermal techniques have been employed since early in the century. Two reasonably comprehensive articles dealing with early work have been reported by Richards.[231,232]

Palmer[233] was among the first to demonstrate the feasibility of a thermocouple instrument to measure the minute intensity variations in an ultrasonic field. For this device, the thermocouple junction was coated with various absorbing materials. It was necessary to calibrate this device.

Fry and Fry,[57,58] on the other hand, demonstrated how the thermocouple probe was an absolute method for the determining ultrasonic intensity. The transient thermoelectric technique consists of a thermocouple junction embedded in a liquid of known ultrasonic absorption.[57,58,213,234,235,236] The absorbing liquid is contained in a holder with polyethylene, as an acoustic window, providing minimal reflection. A transient ultrasonic pulse, usually of one second in duration, is incident upon the junction. The time rate of temperature change, dT/dt, is related to the ultrasonic intensity by the expression

$$I = \frac{\rho C J}{2\alpha} \left(\frac{dT}{dt}\right)_o$$

where ρ, C, J and α are, respectively, the density, the heat capacity per unit volume, the mechanical equivalent of heat and the amplitude ultrasonic absorption coefficient, of the absorbing liquid in which the junction is embedded. A comparison with optical and radiation force techniques in 1958 showed agreement to within approximately $\pm$ 10 percent.[217] This transient thermoelectric technique also provides the opportunity to determine the absorption coefficient in tissue (see Chapter III in this Volume).

Zienuik[237] employed the principal used by Palmer[233] whereby the difference in voltage was determined between sets of thermocouples in and out (shielded) of the sound field.

Thermistors have also been employed to measure the fine details of an ultrasonic field.[238] In this technique, the thermistor probe senses the intensity of a continuous wave ultrasonic field but must be calibrated.

The principal attraction of calorimetric techniques lies in the fact that when ultrasonic energy is completely absorbed, a precise determination of the temperature change assures a precise determination of ultrasonic power. This, of course, requires that the thermal losses to the environment be at a minimum or, at least, determinable and under control. Also, calorimetric techniques are preferred for determining ultrasonic power in those cases where nonlinear effects, including cavitation, exists since these methods are ideally suited to transfer energy into heat, regardless of its spectral distribution.[221,239]

Three types of calorimetric techniques have been reviewed[240] and include a steady flow system, a transient system and a substitution system, all of which require the conversion of ultrasonic energy into thermal energy <u>via</u> a suitable high absorbing material. Thus each yields an absolute measure of ultrasonic power. In the steady flow system the change in temperature between the input and output of a flowing absorbing liquid is determined. In the transient system, the time change of temperature of the absorbing medium is determined and, in principle, this is similar to the transient thermoelectric probe with the exception that the total power is the determinable parameter. The substitute system duplicates the thermal history in the absorbing medium between the absorbed ultrasound and an embedded electrical heater and thus equates the electrical power to the ultrasonic power.

Sokollu[241] described a miniaturized calorimeter designed with the purpose of measuring total ultrasonic power from a focused source. The system's sensitivity restricted its application to levels above one watt. Its accuracy was highly dependent upon the accuracy of the electrical power measurement comparison. van den Ende[242] also described a similar calorimeter but it was not necessarily designed for a focused signal. Comparison with a radiation force technique over the power range from about 0.3 to 1.2 watts showed agreement to within $\pm$ 10 percent. Comparison of a spherical calorimeter within the same power range to a radiation force technique agreed to within $\pm$ 6 percent.[202]

Zienuik[243,244] developed the theory for and argues that the non-adiabatic, non-isothermal calorimeter does not represent two drawbacks of quasi-adiabatic calorimeters, viz., corrections of heat exchange are extremely difficult and high thermal inertia.

Mikhailov[245] developed a calorimeter which made use of the thermal expansion of the sound absorbing liquid. The device is calibrated by applying a known electrical power through the heating coil, embedded in the absorbing liquid. The thermal expansion of the absorber is determined by measuring the rise of the liquid in a small capillary tube.

A calorimetric technique has been employed to yield ultrasonic intensity by providing a finite size aperture, diameter of 5 wavelengths, for the ultrasonic energy to enter and hence to be absorbed.[246] This device has been used to measure the peak ultrasonic intensity (over limited spatial extent).

Szilard[247] reported on an instrument which could be used to measure the ultrasonic energy delivered to patients during therapeutic treatment. A plate, which is affixed to the therapeutic transducer surface during treatment, has a series of thermistors embedded to monitor the temperature change and, hopefully, the ultrasonic energy delivered to the patient.

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