ULTRASONICS METROLOGY I. THE HISTORY OF THE MEASUREMENT OF ACOUSTIC PRESSURE AND INTENSITY USING HYDROPHONES

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I. Introduction

The calibrated piezoelectric hydrophone is now the bedrock of practical ultrasound metrology for medical applications. It is a small receiving transducer, consisting of a piezoelectric element that converts variations in acoustic pressure to electric charge, which may then be detected electronically.

Piezoelectricity was discovered in 1880 by the Curie Brothers and is a property of a select group of materials, such as quartz, which generate electricity in response to external applied pressure and vice versa. Piezoelectric materials lack a centre of symmetry in the unit cell of the crystalline structure. The application of stress to such materials generates electrical polarisation (an electric field) in the direction of the applied stress. The reverse effect (the motor effect) also occurs; application of an electric field causes a mechanical strain. These two effects can be used to detect and transmit ultrasound.

The central place now held by piezoelectric hydrophones was not always so. Optics, calorimetry and radiation force were the dominant approaches to metrology during the initial decades of the development of ultrasound, and it was not until the 1970s that serious attention was addressed to the design and construction of high-fidelity hydrophones for medical ultrasound equipment evaluation.

This first section will trace the early development of hydrophones for ultrasound metrology, leading up to the initial uses for medical ultrasound.

II. Hydrophones for marine use

The need to detect submarines during the 1914-18 war gave the impetus for Paul Langevin's invention of an ultrasonic pulse-echo system using resonant quartz transducers. Working with the engineer Charles-Louis Florisson, Langevin continued after the war to develop a commercial echo-sounding device, with the company *Société de condensation et d'applications méchaniques* (SCAM). He protected his inventions using patents and in one, filed in 1926, he described a quartz hydrophone probe that was small enough to investigate the spatial variations in the ultrasonic field (Fig. 1) [1].

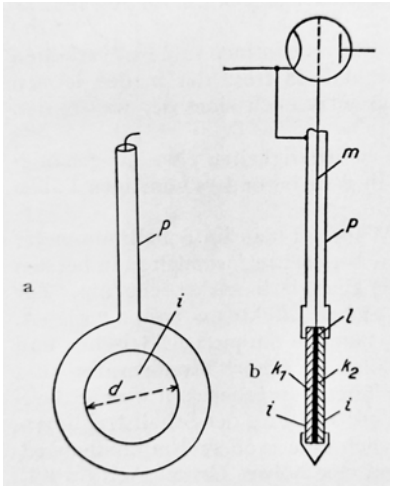

Fig. 1 Langevin's quartz crystal probe (1926). a) Front view. b) cross section. k_1 k_2 quartz plates; l metal plate; p metallic housing; m insulated cable; i electrodes. (From Langevin 1926) [1]

The diameter of the probe, d, was small compared with the wave-length, so approximating to a point receiver. Working at 40 kHz, this suggests that d was about 1-2 cm. He introduced two notable innovations. Two quartz plates were mounted with opposing electrical axes and the probe was placed with its face aligned along the direction of the field, not facing into the beam. In this way he minimised reflections, beam disturbance and the potential for standing waves. The second innovation was to remain an essential feature of all successful subsequent designs. The small charge generated by the quartz plates when exposed to an ultrasound beam was fed directly to an amplifier with very high input impedance, in his case the grid of a triode. This avoided the loading effect of an interconnecting cable.

Langevin had been investigating the use of piezoelectric probes for measurement before he came up with this design. A brief publication in the *Journal de Physique et Radium* in 1923 is the first reference to the use of a piezoelectric probe to measure intensity. The work was carried out with a Japanese physicist and seismologist, Mishio Ishimoto, visiting from Tokyo Imperial University [2]. The derivation of intensity from hydrophone measurements of acoustic pressure is now an integral part of the methodology of acoustic exposure estimation.

Crystals other than quartz were investigated for ultrasonic transducers during the 1920s and 1930s. In tourmaline, the polar axis coincides with the optical axis, so it was suitable as a hydrophone receiver for studies of the transmission of high-amplitude pressure waves from underwater explosives, where the three-dimension strains could result in charge cancellation in quartz [3]. The piezoelectric effect in Rochelle (Seignette) salt is considerably greater than in either quartz or tourmaline [4], and it was used widely in hydrophone construction once a reliable method of manufacture was established.

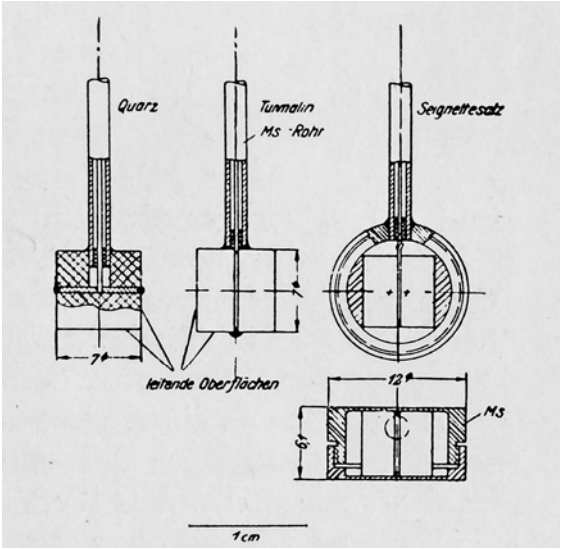

Fig. 2 Three alternative hydrophones, using quartz, tourmaline and Rochelle (Seignette) Salt. (From Meyer and Tamm 1939) [5]

Meyer and Tamm included examples of hydrophones made from these three alternatives when reporting a laboratory study into acoustic cavitation in 1939 (Fig. 2) [5]. Each hydrophone had an overall dimension of about 1 cm, less than one wavelength for the low ultrasonic frequencies under investigation. In use, the amplified signal was rectified and smoothed for measurement using a meter or, once the technology was available, displayed on a cathode ray oscilloscope. Resonant hydrophones, operating at the frequency of the transmitting transducer, were only used when sensitivity was a critical factor. Usually, they were designed to operate away from and below resonance, where the frequency response was flatter. Frequency-compensating amplifiers were introduced during the development of SONAR in the 1939-45 war. By 1946, a wide variety of hydrophones had been developed for naval use to operate at frequencies up to about 100 kHz [6].

Some piezoelectric materials are also ferroelectric. Ferroelectric materials such as barium titanate (BaTiO3) in ceramic form were investigated in secret in several countries, including the US, UK, USSR and Japan, during the 1939-45 war in an effort to develop high dielectric constant materials for capacitors. It was only after the war that it was established that the high dielectric constant of $BaTiO_3$ was due to its ferroelectric properties [7]. Unlike simple piezoelectric materials, which produce a

polarisation only when under stress, ferroelectric materials develop polarisation spontaneously and form permanent dipoles in their crystalline structure. Local ferroelectric domains are formed in which the direction of polarisation is aligned. When manufactured, the polarisation domains within the material are randomly oriented, resulting in zero net polarisation. In 1945, it was discovered that an external electric field could orient the ferroelectric domains within the polycrystalline ceramic grains, thus producing a material that acted like a single ferroelectric crystal. This "poling" process, which was carried out above a critical temperature, the Curie point, turned an inert ceramic into an electromechanically active material with a multitude of uses. The electromechanical response of barium titanate was found to be about 2 orders of magnitude greater than that of quartz. In 1954, it was reported that another ferroelectric material, lead zirconate titanate (PZT), had useful piezoelectric transducer properties [7]. PZT in various forms soon became the ferroelectric material of choice for ultrasound transducers for diagnostic and therapeutic applications. However, piezoelectric ceramic materials have proved to be less satisfactory for hydrophones, as will be shown.

III. Hydrophones for medical ultrasound

A. Hydrophones with miniature piezo-ceramic cylinder elements

By 1950, companies such as the Brush Development Company, founded in 1919 to utilise piezoelectric crystals, were manufacturing barium titanate ceramic elements in a wide range of forms. By the time Physikalisch-Technische Bundesanstalt (PTB) needed to establish procedures for type-testing German therapeutic ultrasound equipment in 1952, piezoelectric transducers made from barium titanate ceramic had become available [8]. The primary output measurement in PTB was acoustic power, using a radiation force balance. But, additionally, PTB measured the mode of operation of the therapy equipment 'with the help of a barium titanate probe microphone and a high-frequency cathode ray oscilloscope'. This initial use of a hydrophone for medical equipment testing was limited to the determination of the operating frequency and of the pulsing regime. It did not include an estimation of intensity from acoustic pressure, so calibration was unnecessary and the frequency response was not critical.

One form of ceramic hydrophone made use of a tiny cylindrical element whose outer diameter and length were only 1.5 mm. In 1954, Ackerman and Holake of Pennsylvania State University described ceramic probe microphones which made use of this small $BaTiO_3$ cylinder mounted on the end of a long thin tube, which could be used in air or water [9]. The hydrophones could be used at frequencies up to 100 kHz in small liquid volumes in which micro-organisms and red blood cells were exposed to acoustic fields. However, they were susceptible to mechanical pickup along the mounting tube which could be coupled to the element. To minimise this, the element was mounted on latex rubber washers and rubber bonding insulated it from the main stem (Fig. 3). This design recognised the importance of mechanical decoupling between the sensing element and the mount, an aspect of hydrophone design that was often missing in later, simpler, designs. Theodor Hueter was sufficiently impressed with this design of hydrophone to include a detailed description in his classic textbook on acoustics [10].

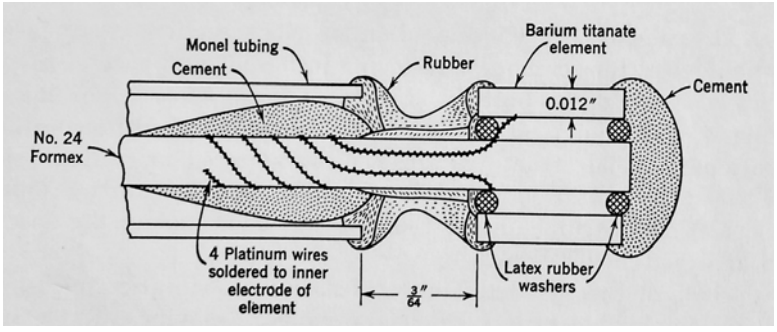

Fig. 3 A miniature piezoelectric ceramic hydrophone based on a small barium titanate cylinder element. (From Hueter and Bolt 1955) [10]

Mellen, of the US Navy Underwater Sound Laboratory, used the same type of barium titanate cylinder to construct a similar device in 1956 (Fig. 4) for the study of the collapse of spherical cavities

in water [11]. The inside and outside of the cylinder were plated with silver. The plating was then removed by sanding, and polarisation was accomplished by heating to 130 °C (above the Curie point) and slow cooling while maintaining a potential of 300 V between inner and outer electrodes. The high dielectric constant of the ceramic material gave a high capacitance for the small element size and wall thickness, allowing it to be connected via a coaxial cable without too much loss of sensitivity. The frequency response of the probe extended to 1 MHz.

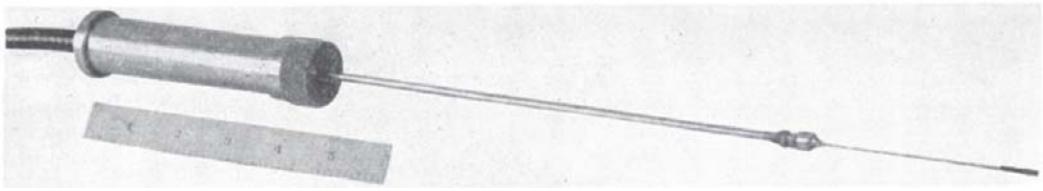

Fig. 4 The probe hydrophone of Mellen (1956) [11]. (Reprinted with permission from Mellen RH. An experimental study of the collapse of a spherical cavity in water. *J Acoust Soc Am* 1956; 28: 447. Copyright 1956, Acoustic Society of America.)

During the next couple of decades, as interest in medical applications of ultrasound focussed largely on therapeutic and surgical uses, little attention was paid to hydrophone development or use. The absence of any reference to hydrophones is very noticeable in reports from three workshops on medical ultrasound held at the University of Illinois in 1953, 1957 and 1965. Even in 1967, a review of ultrasonics applied to medicine limited the discussion on piezoelectric hydrophones to those with cylindrical shape, noting that these should only be used at frequencies below resonance, where the frequency response is flat, effectively limiting their application to below 1 MHz [12].

In 1970, Kit Hill, of the Institute of Cancer Research, reported on the construction of a hydrophone probe based on a 1.6 mm x 1.6 mm cylindrical element for the measurement of beam shape and pulse shape from bio-medical ultrasound sources [13]. This cylindrical element was made of lead zirconate titanate (PZT). The directional response of the probe was symmetrical about its axis but was found to have a strong resonance at 1 MHz when the beam direction was parallel to the probe axis. Hill used the device to plot the beam profile, in terms of output voltage, from a 1 MHz, 3 cm diameter transducer and noted that for absolute measurements at 1 MHz or above, it would be necessary to calibrate the device at each frequency and pulse length of interest.

B. Hydrophones with small piezo-ceramic disc elements

Finally, during a workshop held at the Battelle Seattle Research Center in November 1971, several presentations started to re-consider the place of hydrophones in the methodology of medical ultrasound metrology. A committee report proposed that a quartz or lithium niobate transducer should be used to determine (a) the point of maximum acoustic pressure, from which the maximum instantaneous intensity should be derived, (b) the radial distribution of intensity in a plane parallel to the sound source at this point and (c) the temporal waveform at this point. From these measurements, the maximum average intensity could be derived, given a number of not unreasonable assumptions [14]. The committee supposed that the transducer should be acoustically shielded so that only a small area of the order of the wavelength is exposed.

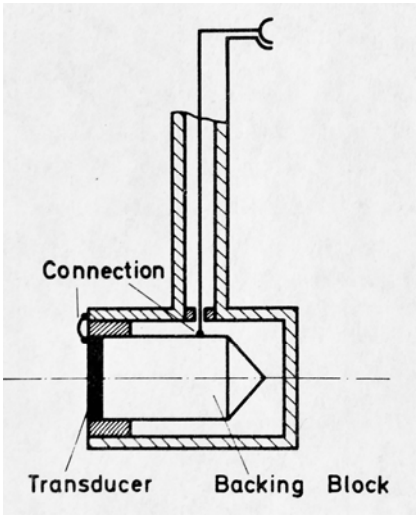

Fig. 5 A general hydrophone design for pulsed diagnostic beams. (From Brendel 1972). [15]

During the same workshop, Klaus Brendel, from PTB, described a general design of hydrophone in which a piezoelectric ceramic plate, of natural resonant frequency above 15 MHz, and of diameter between 1 mm and 5 mm, was mounted on a backing block (Fig. 5). Such a device would satisfy his criteria of time-independent properties, small size compared with the wavelength (at least for frequencies up to 1 MHz) sufficient sensitivity and sufficient frequency range [15].

Dennis Newman, from Battelle Northwest, Washington, who was working in high-frequency materials testing, addressed some of the challenges that would have to be overcome in the design of a practical hydrophone that could operate up to 10 MHz. He identified the most serious of these to be the sensitivity to orientation, the lack of wideband response and the need for calibration. His hydrophones were quite simple in design, consisting of PZT piezo-ceramic plates with active areas of about 1 mm diameter, mounted directly on the end of short tubes. Schlieren photography demonstrated the directionality of such small elements, especially at frequencies above 5 MHz [16].

One of the challenges in constructing probe hydrophones was in achieving a small enough element to give good directionality while maintaining sufficient sensitivity to measure low intensity areas of the beam. By 1974, Harold Stewart, from the Food and Drug Administration in the USA, was able to report a commercially available ceramic hydrophone, 460 μm in diameter [17]. By the late 1970s, Weight and Hayman, of City University, London, had constructed an even smaller, wideband receiving probe with a diameter of 150 μm using a PZT disc element of 40 μm thickness [18]. The small piece of PZT was soldered to a brass wire which acted as the back face connection. The PZT was then shaped to the required dimensions. The front face connection was made by first coating the probe tip with insulating epoxy then carefully abrading it to expose the front face of the PZT, which was coated with conductive paint and then recovered with epoxy. Due to the small size of the device and low capacitance, the electrical source impedance was high, requiring a close coupled preamplifier. The hydrophone was used to investigate the acoustic fields from transducers driven by single cycle excitation.

In 1981, Peter Lewin and Bob Chivers, of the Danish Institute of Biomedical Engineering and the University of Surrey respectively, constructed a miniature ceramic disc probe hydrophone for acoustic pressure measurements in liquids at low megahertz frequencies [19]. The device used a 0.5 mm diameter, 0.1 mm thick PZT disc polarised in the thickness direction mounted on the end of a 0.5 mm diameter glass tube. It had reasonable directional response and a flat frequency response from 0.5 to 6 MHz, well below the resonant frequency (19 MHz) of the disc element. They compared its performance to that of a hydrophone consisting of a 1.5 mm x 1.5 mm PZT cylinder mounted on the end of a 2 mm hypodermic needle. The cylinder element was positioned on rubber washers to provide acoustic insulation from the needle. The hydrophone showed very uniform directivity in the plane perpendicular to the cylinder axis, but a number of resonances at frequencies near to 1 MHz, limiting its use to frequencies up to 0.7 MHz [19]. Filmore and Chivers, of the University of Surrey, constructed and tested batches of miniature ceramic needle hydrophones with disc elements of diameters in the range 0.5 – 1.0 mm [20]. These were shown to have quite non-uniform frequency

responses up to 18 MHz and directivity patterns that did not always follow the pattern expected for a plane circular piston receiver.

Achieving a uniform frequency response with a small PZT ceramic hydrophone was challenging. To meet the requirement for adequate spatial resolution and good directional characteristics, the disc element had to be small. Also, the disc needed to be no thicker than a few hundred μm to ensure that the fundamental thickness mode resonance was beyond the range of interest. The constraint on disc diameter led to the presence of radial modes of resonance at frequencies of a few MHz affecting the overall frequency response. Coupled with inevitable manufacturing tolerances, these resonance effects made it difficult to produce ceramic hydrophones with consistent properties. In particular, the non-uniform frequency response could cause severe distortion of ultrasound short pulse voltage waveforms typical of imaging systems, leading to potentially large errors in estimating peak pressure values. To minimise measurement uncertainties it was necessary to calibrate ceramic hydrophones at small frequency intervals to ensure that their frequency and directivity responses were well known.

Despite the limitations of ceramic probe hydrophones, they had their uses in characterising the acoustic frequency and pulse regime from sources such as physiotherapy transducers, whose output typically consists of long time-duration, low amplitude narrowband pulses. The small surface area that the miniature probe presents to the beam results in minimal reflection back to the source and hence avoids problems with standing waves.

C. PVDF (polyvinidilene fluoride) membrane hydrophones

Perhaps the single most important development in ultrasound metrology in recent decades was the development of hydrophones manufactured from the piezoelectric polymer material PVDF. The piezoelectric properties of this material were discovered by Kawai in 1969 leading to a wide range of applications in transducer technology [21]. PVDF had been developed mainly as a dielectric material for capacitors and was available in rolls or sheets ranging in thickness from a few to several hundred microns [22, 23]. The piezoelectric properties were enhanced by stretching or drawing the sheet unaxially or biaxially. This increased the number of dipoles which could couple to the polarising electric field during the subsequent poling process. Before its application to ultrasound hydrophones, PVDF had been used in a number of pressure sensing applications, such as shock sensors for measuring deceleration in vehicle crash studies [23].

A membrane version of the PVDF hydrophone was first described in 1978 by De Reggi et al.[22, 24] The membrane hydrophone consisted of a sheet of the material stretched across an annular frame, large enough to allow the acoustic beam to pass through its aperture. Electrodes were vacuum deposited on opposite surfaces of the membrane and used to pole a small active region or spot at elevated temperature to define the spatial characteristics of the device. Early versions of these “spot poled” membrane hydrophones in single layer and bilaminar forms were developed in the US in the late 1970s at the National Bureau of Standards and the Bureau of Radiological Health and shown to have very useful properties for measurements at diagnostic ultrasound frequencies [25, 26].

At about the same time in the UK, collaborative work between the National Physical Laboratory (NPL) and GEC Marconi led to the development of similar devices. NPL is the UK’s national measurements standards laboratory and became involved in medical ultrasound in the late 1970s in response to requests from medical physics departments and manufacturers for help in making reliable measurements of the acoustic output of medical ultrasound equipment. NPL soon discovered the limitations of ceramic hydrophones, due to their non-uniform frequency response, and was aware of work at the Marconi Research Centre at Chelmsford on hydrophones based on PVDF for underwater acoustics applications [23]. Marconi had significant expertise in constructing multi-layer devices using 25 μm PVDF sheets and collaborated with NPL to produce a membrane hydrophone suitable for medical ultrasound measurements. In 1980, Shotton et al. described a prototype membrane device consisting of a 25 μm sheet of PVDF stretched over a 100 mm diameter perpsex ring [27]. A 4 mm active element was defined by the overlap of the vacuum deposited gold/chromium electrodes and activated by poling at over 100 °C with an electric field of approximately 1 MV cm^{-1}. The unshielded metal film leads to the active element were well separated to minimise load capacitance. Following the successful performance of the prototype, a range of devices was produced with element sizes down to 1 mm made from 25 μm and 9 μm film.

PVDF membrane hydrophones were manufactured by Marconi from the early 1980s for the following 20 years or so and are still used by NPL and many other laboratories (Fig. 6). Many hundreds were calibrated and supplied to customers around the world. The stability and predictable performance of PVDF membrane hydrophones soon led to their adoption as the gold standard

hydrophone for the characterisation of medical ultrasound fields and led to their embodiment in international measurement standards.

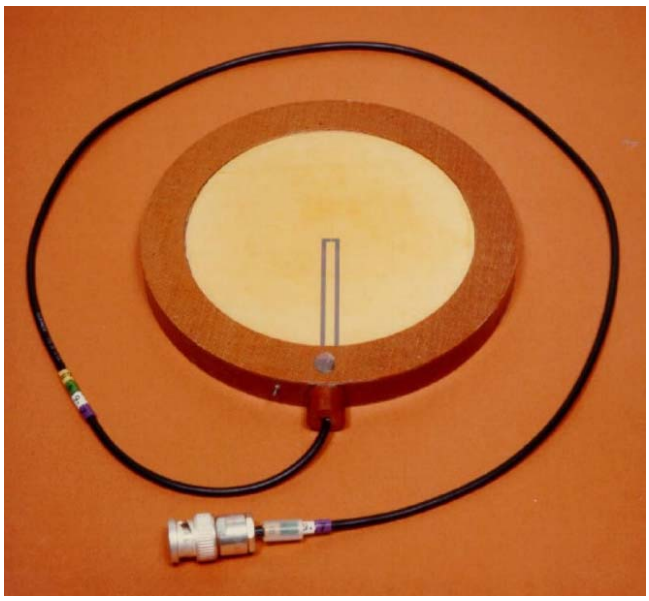

Fig. 6 An early NPL/Marconi bilaminar membrane hydrophone. (Image courtesy of Dr B Zeqiri, NPL, Teddington, UK)

For use in diagnostic fields with short pulse exposures, the early PVDF membrane hydrophone had a number of advantages over the probe hydrophone. The thin membrane (typically 9 – 25 μm) allowed through transmission of the beam with little perturbation, so that the free field acoustic pressure was sensed at the central element. The low acoustic impedance of the material, which is close to that of water, resulted in only weak reflection at its surface; the close acoustic match to water resulted in a low Q resonance and a broad frequency response which rose slowly towards the resonant frequency [28]. The shape of the active spot defined by the electrodes was close to the ideal shape of a plane disc, leading to a predictable directional response which conformed closely to that of a plane piston. For measurements in diagnostic ultrasound fields, the active spot would normally be 1 mm or less in diameter. The main limitation of the membrane hydrophone was that its bulk could prevent measurements being made in close proximity to the transducer face.

The ability of the membrane hydrophone to reject external electrical noise was affected by the arrangement of electrodes deposited on the PVDF film. Preston et al. of NPL, described three possible electrode arrangements [29]. The coplanar shielded type (Fig. 7) consisted of a single sheet of PVDF with an active element surrounded by shielding electrodes. On one side, the shielding electrode connected to the active element electrode. The bilaminar type (Fig. 8) used two PVDF membranes bonded together. The active element electrode and its connecting lead were deposited on the inner surface of one of the layers. The outer surfaces of both membranes were almost entirely covered in shielding. The differential configuration used matched connections to the active element on opposing sides of the membrane surrounded by shielding. The electrical terminals were connected to a differential amplifier. The bilaminar design gave improved signal to noise ratio over the coplanar shielded type due to the extra shielding. However, the additional thickness also resulted in a higher reflection coefficient, which could cause problems with standing waves with long pulse or continuous waves. The differential design gave even better signal to noise ratio that the bilaminar type without the disadvantage of the thicker membrane [29].

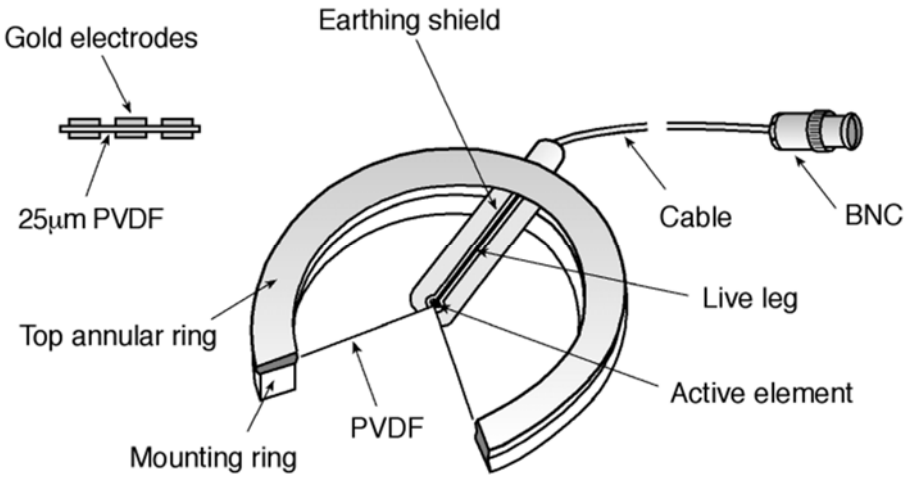

Fig. 7 The electrode configuration of the coplanar shielded membrane hydrophone. (Image courtesy of Dr B Zeqiri, NPL, Teddington, UK)

The Marconi hydrophones were made in 3 thicknesses; these were 9 μm, 25 μm and 50 μm (bilaminar). The resonant frequency, and hence the useful frequency response of the membrane hydrophone, was determined by the thickness of the membrane. A 25 μm PVDF hydrophone has a natural resonance at about 50 MHz and a frequency response that rises slowly over most of the diagnostic frequency range (0-25 MHz) towards this value. A 50 μm bilaminar hydrophone has a resonance at about 25 MHz, whereas the resonant frequency of the 9 μm hydrophone was over 100 MHz, well outside the diagnostic range [30]. The sensitivity of the hydrophone was determined almost entirely by the area of the active element.

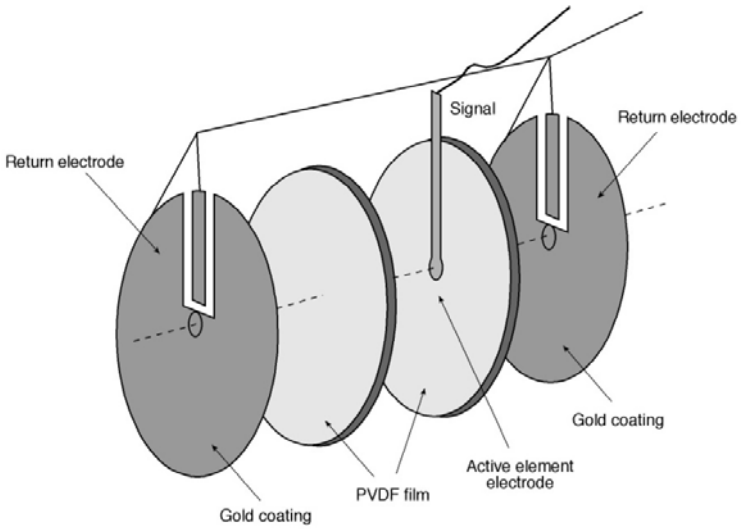

Fig. 8 Construction of the bilaminar shielded membrane hydrophone. (Image courtesy of Dr B Zeqiri, NPL, Teddington, UK)

D. PVDF probe and needle hydrophones

In the early 1980s, companies in the US were manufacturing PVDF probe hydrophones [26]. Similar work at the Danish Institute for Biomedical Engineering led to the development and sale of needle probe hydrophones with thin disc PVDF elements between 0.6 mm and 1 mm in diameter. Their characteristics were described by Lewin [31]. They showed much better performance than ceramic hydrophones in terms of directional response, and the frequency response extended to about 10 MHz, with some variations of up to 3 dB in the 1 – 3 MHz range. These variations were shown to be due

mainly to radial modes of resonance in the material backing the PVDF element and to diffraction phenomena at the edges of the hydrophone front face [32]. Although such traits might affect measurements of short pulse ultrasound waveforms, needle probes have advantages over membrane hydrophones in characterising field distributions where long pulses or continuous wave fields are used, such as with physiotherapy devices. The small area of the probe limits the formation of standing waves, especially close to the source. Needle hydrophones have also been used for in situ measurements within tissues. The frequency response limitations have been addressed in modern PVDF probe hydrophones by careful compensation of the frequency characteristics (see later).

In 1990, a new company, Precision Acoustics, was formed in Dorchester, UK by medical physicist Joe Aindow and radiographer Terri Gill. The main focus of the company was the manufacture of PVDF needle hydrophones for the characterisation of medical ultrasound fields. The needle probes were made with diameters in the range 0.2 – 2 mm and were mounted interchangeably, directly into a submersible preamplifier (Fig. 9). Electrical power to the submersible preamplifier was supplied via the sealed-in coaxial cable by a DC coupler module outside of the water tank, which also coupled to the hydrophone voltage signal.

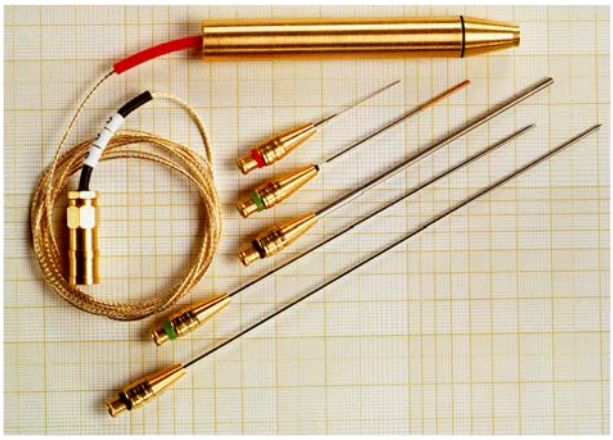

Fig. 9 A set of early Precision Acoustics needle hydrophones. The hydrophones could be connected interchangeably into the submersible preamplifier shown at the top of the picture. (Image courtesy of Dr A Hurrell, Precisions Acoustics, Dorchester, UK)

Precision Acoustics supported the marketing of the Marconi membrane hydrophones in the late 1990s until production ceased in 2000, and by September 2001 had begun the manufacture of its own membrane hydrophone design (Fig. 10).

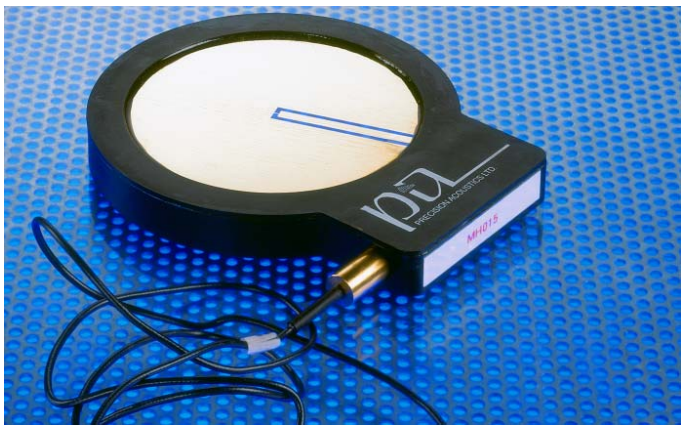

Fig. 10 An early model of Precision Acoustics bilaminar membrane hydrophone. (Image courtesy of Dr A Hurrell, Precisions Acoustics, Dorchester, UK)

In America, hydrophones for medical ultrasound metrology were made by Onda Corporation of Sunnyvale California, from the early 2000s onwards, including needle and membrane types.

IV. Finite amplitude measurements

The development and use of PVDF hydrophones with such broad and predictable frequency responses soon led to the observation of distortions in the pulsed waveforms from diagnostic imaging devices. Despite some initial criticism that the distortion in the voltage waveform still resulted from poor frequency response, it was soon demonstrated that this distortion was due to non-linear propagation of ultrasound in water at finite pressure amplitudes [33]. Distortion of pressure waves due to non-linear propagation was a well-known phenomenon in underwater acoustics but had not been considered in biomedical applications of ultrasound [34]. As the pulse propagates from the transducer, in a diffractive field, the compressional parts of the wave become enhanced and the rarefaction parts reduced (see Fig. 11). [35] Further propagation leads to the formation of a shock wave containing higher harmonics of the transmitted frequency, which then become attenuated more rapidly. Such distortions created new questions on how to make relevant acoustic pressure measurements in water [36]. Non-linear propagation effects are much weaker in tissue than they are in water, leading to difficulties in estimating in-situ exposure levels in tissue from measurements made in water. The presence of high frequency harmonics also placed additional demands on hydrophone performance in terms of wider frequency response and better spatial resolution to cope with their shorter wavelengths.

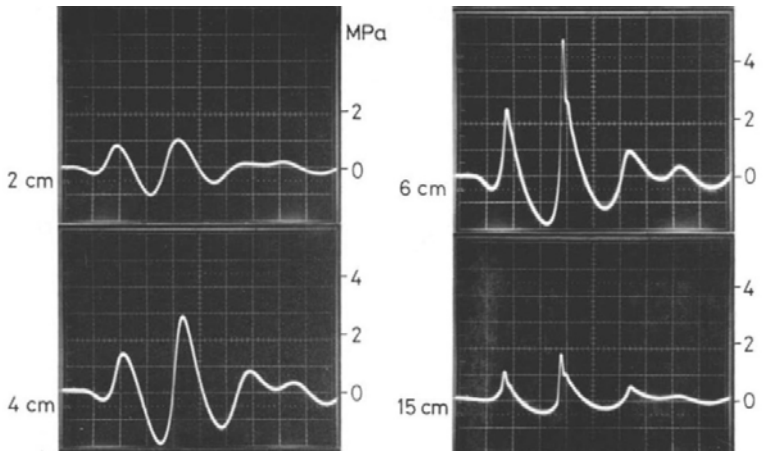

Fig. 11 The pressure pulse measured in the field from a focused 2 MHz transducer at 2 cm, 4 cm, 6 cm and 15 cm range. (From Duck FA, Starritt HC. Acoustic shock generation by ultrasonic imaging equipment. *Brit J Radiol* 1984; 57: 231-40.) [35]

V. Hydrophone calibration

Hydrophones are not absolute measurement devices and must be calibrated in terms of sensitivity and frequency response to allow measurement of absolute pressure. Most of the earliest methods used were difficult and time consuming. The reciprocity method described by Ludwig and Brendel involved calibrating an auxiliary transducer by self-reciprocity using reflection from a plane interface and then determining the sensitivity of the hydrophone by placing it in the known field of the transducer [37]. At megahertz frequencies, however, uncertainties arise in determining the acoustic beam profile and the electrical characteristics of the transducer. The uncertainties increase with frequency, limiting the use of the method to frequencies up to 15 MHz.

A widely used early method of calibration was the elastic sphere radiometer [38-40], in which the acoustic intensity at a point in an ultrasound field was determined from the radiation force acting on a small (a few wavelengths diameter) sphere suspended in the beam by fine nylon filaments (Fig. 12). The beam was directed horizontally in water at the sphere and the radiation force (*F)* was calculated from the measured horizontal deflection (*d*), from which the acoustic intensity could be derived. A value for the sensitivity of the hydrophone could then be determined by placing it at the same point in the field. A determination of hydrophone sensitivity in terms of volts per pascal could be made using the plane wave assumption, in which intensity is given by the square of the acoustic pressure divided by the acoustic impedance of the propagating medium (water). In theory, the intensity can be derived from first principles knowing the dimensions of the sphere, its mass and acoustic properties, and the

length of the filament. However, the measured force tends to be strongly frequency dependent and the deflection quite small for diagnostic intensity levels [41].

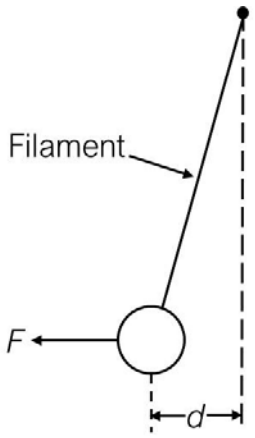

Fig. 12 The elastic sphere radiometer suspended in a sound field.

Planar scanning has also been widely used as a means of calibration. In this case, the hydrophone is scanned over a plane perpendicular to the beam axis and the square of the hydrophone voltage measured at each elemental area. Again, assuming plane waves, the square of the hydrophone voltage is proportional to the intensity at each point and the sum of the values over the beam cross section is proportional to the total power in the beam. The total power in the beam is then measured using an alternative absolute measurement device such as a radiation force balance or calorimeter. The hydrophone sensitivity may then be calculated from the ratio of the summed pressure squared values to the total power in the beam [42, 43]. The method can be quite time consuming and errors arise from instability in the transducer output and noise from the hydrophone.

From the early 1980s, the use of PVDF membrane hydrophones for the determination of the spatial and temporal characteristics of ultrasonic fields had become firmly embodied in national and international standards [44]. The calibration techniques just described were no longer regarded as being sufficiently accurate and reliable to meet the requirements of these standards. Calibration of hydrophones for use in acoustic field measurements should be traceable to a national primary standard. In the UK, the primary standard is held by the National Physical Laboratory and calibration is disseminated via secondary standard hydrophones to the user community [45]. At NPL, the primary standard method of calibration is based on optical interferometry. In this technique, acoustic displacement is detected at a point in the field using a thin plastic membrane (the pellicle) which is coated on one side with a reflecting layer of gold. The pellicle is thin enough (3.5 – 5 μm) to be able to follow the acoustic waveform. The displacement of the pellicle (of the order of tens of nanometres) is then measured using a Michelson optical interferometer, from which the acoustic pressure can be calculated. The hydrophone to be calibrated is then placed at the same point in the acoustic field and its output voltage measured. An advantage of the interferometry method is that it measures a primary property of the ultrasound field, i.e. displacement, offering direct traceability to primary standards of length [46].

The interferometer facility was developed at NPL in the 1980s in collaboration with AERE Harwell [46, 47]. The technology was originally developed at AERE Harwell in the early 1970s to measure the integrity of materials through the measurement of surface displacement of ultrasonic transducers, but was improved at NPL to meet the requirements of the hydrophone calibration method. The main improvements were the extension of the frequency response and the reduction of the noise level [47].

Calibration of secondary standard hydrophones was carried out by taking advantage of the non-linear distortion that occurs in a high amplitude pulse propagating through water. A relatively low frequency, high amplitude source (e.g. 1 MHz) was used to generate an acoustic waveform which becomes strongly distorted due to non-linear propagation during transmission to a specific point in the acoustic field. The distorted waveform contains many harmonics at multiples of the original transmit frequency to beyond 20 MHz [44, 48]. Comparison of the frequency content of the waveforms from the test and secondary standard hydrophones gave a rapid calibration over a wide range of frequencies.

The calibration information provided with a hydrophone typically consisted of a series of values for voltage sensitivity at discrete frequencies over the usable bandwidth of the device. For probe hydrophones, the frequency response typically showed variations in sensitivity at low frequencies which can lead to inaccurate representation of the true pressure waveform and errors in pressure measurements (Fig. 13) [49].

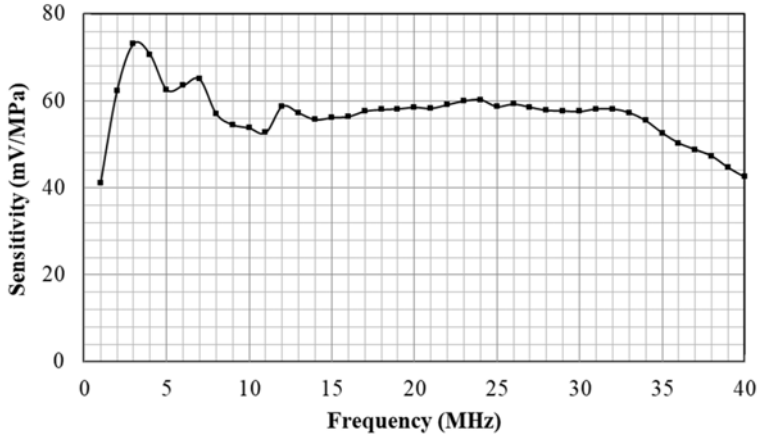

Fig. 13 Amplitude response of a 9 μm, 0.2 mm diameter PVDF needle hydrophone. (Courtesy of Dr A Hurrell, Precisions Acoustics, Dorchester, UK)

For a membrane hydrophone, a typical frequency response would show the sensitivity rising gradually towards the resonant frequency of the membrane (Fig. 14). The increased sensitivity at higher frequencies could result in overestimation of peak positive pressure values from distorted waveforms such as those shown in Fig. 11.

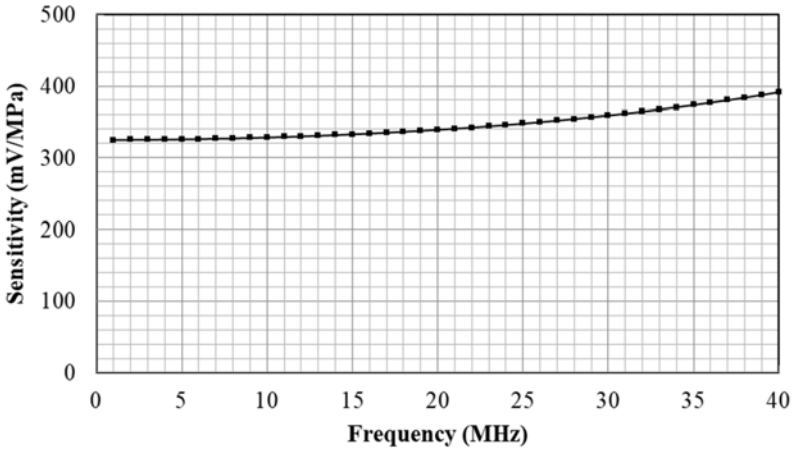

Fig. 14 Amplitude response of a 16 μm, 0.4 mm PVDF differential membrane hydrophone. (Courtesy of Dr A Hurrell, Precisions Acoustics, Dorchester, UK)

Measurement standards impose a flatness criterion on frequency response which restricts the acceptable change in sensitivity over the usable bandwidth, so that pressure measurements can be made using a single value of sensitivity at the acoustic working frequency of the source. To meet the flatness criterion, the hydrophone voltage sensitivity had to be within ± 3dB of the value at the working frequency of the source transducer over a defined frequency range [50]. To cope with the harmonic frequencies generated by non-linear propagation of ultrasound waveforms in water, this range extended from half the source frequency to eight times its value (or 40 MHz). For a membrane hydrophone, flatness could be improved by using a preamplifier whose response rolls off towards the hydrophone

resonance. However, this approach restricts the bandwidth of the hydrophone and does not compensate for variations in the low MHz region of probe hydrophones.

More effective compensation can be achieved by deconvolution of the voltage waveform with the whole amplitude frequency response of the hydrophone. This has been shown to be even more effective if amplitude and phase information in the hydrophone frequency response is used in the deconvolution. The deconvolution is carried out digitally on the voltage waveform and results in much more accurate measurements of acoustic pressure quantities and extends the frequency response to beyond the resonant frequency [51]. The method is effective for both membrane and probe hydrophones.

VI. Measurement of pressure and derived intensity from diagnostic ultrasound fields

One of the main uses of the hydrophones described above is for the characterisation of the acoustic output of diagnostic ultrasound systems. The typical output from a diagnostic ultrasound imaging system consists of short pulses of ultrasound, typically a few cycles long (see Fig. 11), transmitted at regular intervals of the order of milliseconds. Typical frequencies within the pulse are in the range 3 - 15 MHz, leading to pulse lengths of the order of microseconds. Characterisation of the acoustic output from a diagnostic ultrasound scanning system requires the measurement of parameters that describe the amplitude of transmitted pulses and their temporal and spatial characteristics, and the total time-averaged power. A wide range of parameters for making such measurements has been proposed, including peak positive and peak negative acoustic pressure, and various intensity parameters defined by their spatial and temporal characteristics [52-54]. Peak negative pressure is of interest as it is related to the risk of cavitation, whereas temporal average intensity and power are relevant to the potential for thermal effects in tissue. In addition, the safety indices Thermal Index and Mechanical index are derived from these measurements. While operating modes such as M-mode and Doppler emit a stationary ultrasound beam which can be assessed without too much difficulty, ultrasound imaging modes involve scanning of the beam through the imaging plane adding further spatial and temporal considerations [52].

Parameters that are defined at their spatial peak are measured with a hydrophone. Whilst pressure values are calculated from the amplitude of the hydrophone voltage signal using the hydrophone calibration factor, values of intensity must be derived from the pressure measurement assuming the plane wave approximation. For derated intensity parameters, as required by the FDA, the maximum derated values must be found. Measured values of pressure and derived intensity are affected by the operating mode of the ultrasound system and the multitude of possible control settings, making for a potentially time consuming process to meet the requirements of regulatory authorities. When the beam is stationary, such as in M-mode, the spatial peak, temporal average intensity (I_{SPTA}) could be derived from the intensity during a single pulse by simply multiplying by the ratio of the pulse duration to the time interval between pulses. For real time scanning modes, the measurement of I_{SPTA} was more difficult due the fact that the beam was scanned past the hydrophone and was detected in several successive positions on each sweep.

To measure the acoustic output parameters required by international standards by methods traceable to national primary standards, a digital system known as the Ultrasound Beam Calibrator (UBC) was developed in the 1980s at NPL in the UK (Fig. 15) [55]. This system made use of a 2 x 25 μm bilaminar PVDF hydrophone which contained a linear array of 21 elements, each 0.5 mm in diameter and spaced at 1 mm intervals. The pre-amplified signal from each was digitised and stored to enable calculation of the various pressure and intensity parameters, but also displayed in the form of a real-time beam profile. As the hydrophone array was moved through the ultrasound beam, the peak values of parameters could be automatically updated and stored to find the spatial maximum values. For real time scanning modes, the hydrophone signal from each beam contributing to the time averaged intensity over the duration of the scan was captured. This required a reliable and stable triggering system to synchronise the capture to the transmission of pulses from the scanning system. An electromagnetic pickup coil position close to the transducer was used for this purpose.

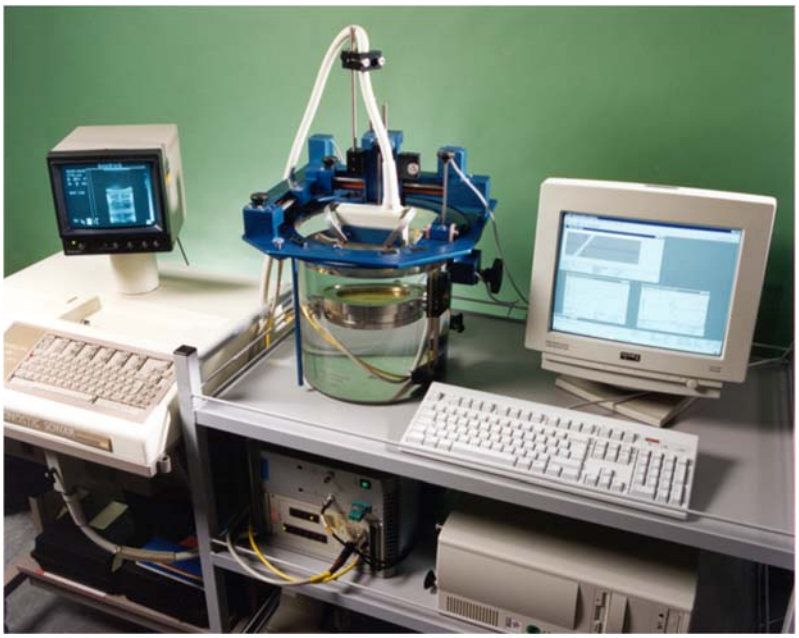

Fig. 15 The NPL Ultrasound Beam Calibrator. (Image courtesy of Dr B Zeqiri, NPL, Teddington, UK)

An alternative method for the measurement of temporal average intensity was described by Martin [56, 57], which made use of an RF power meter to integrate the contributions from all signals received by the hydrophone. The hydrophone signal was amplified and connected to the thermocouple sensor of the power meter. This measured the time averaged electrical power in the hydrophone signal by measuring its heating effect within the sensor. The real-time read out of electrical power was proportional to the time averaged intensity in the beam and could be used to locate the position of the spatial maximum and calculate its value. This analogue approach avoided the extensive calculations required in the digital approach to the measurement of time averaged intensity and obviated the need for a trigger signal.

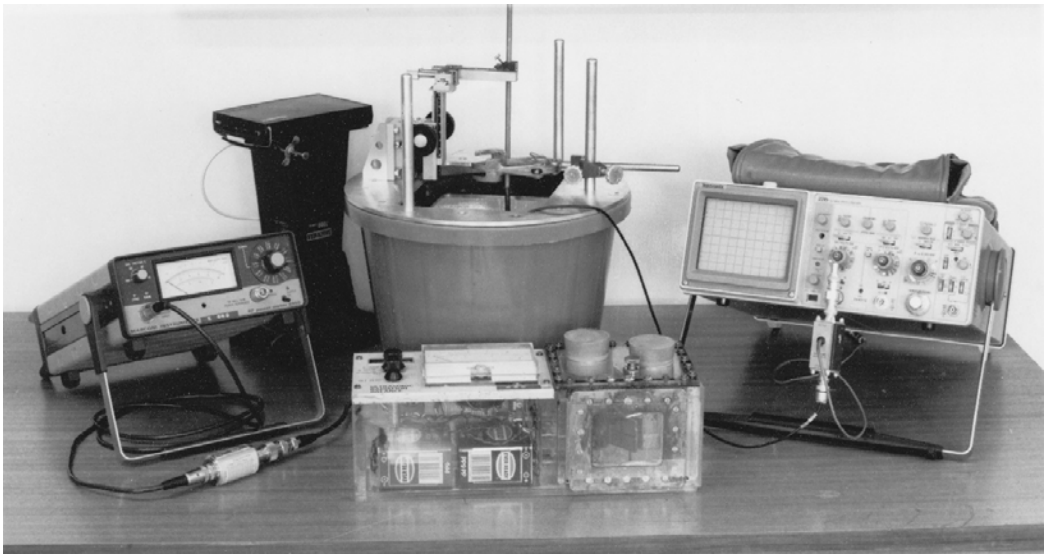

Fig. 16 Portable measurement system used to make acoustic output measurements on clinical ultrasound machines in the Northern Region, including the RF power meter (left) and Farmery and Whittingham radiation force balance (centre front). The membrane hydrophone is mounted inside the water bath (centre rear).

Identifying the location and operating conditions which gave the maximum values of the required pressure and intensity parameters became a significant challenge as ultrasound imaging systems became ever more sophisticated and included more modes of operation. Such measurements were often carried out by NHS Medical Physics Departments in the UK as part of their monitoring programmes of the safety and effectiveness of ultrasound systems in clinical use. They were particularly challenging when they had to be performed in a scanning clinic environment under significant time pressure between scanning sessions rather than in a laboratory. The measurement system needed to be relatively portable and compact (Fig. 16). To assist in the search process, Henderson et al. developed protocols to speed up the search process, based on some simplifying assumptions about the behaviour of the imaging system [58]. These were shown to benefit the process in terms of consistency and time savings. Later, Whittingham et al. developed a portable system for checking the accuracy of displayed thermal and mechanical indices [59], as recommended in the Safety Guidelines of the British Medical Ultrasound Society [60]. This involved finding the maximum values of de-rated I_{SPTA}, for assumed attenuation coefficients of both 0.3 and 0.6 dB cm^{-1} MHz^{-1}, with the machine controls set in a repeatable way as opposed to the way that gave the maximum possible value of I_{SPTA}.

VII. Surveys of acoustic outputs of diagnostic ultrasound equipment

A number of surveys of maximum acoustic output parameters from diagnostic ultrasound systems were carried out in the UK in the 1980s and 1990s. An early survey, reported in 1985, of a small number of static B-scanners, real-time linear array and mechanical sector scanners in clinical use in the South West of England [36], showed little apparent change in output levels from previous surveys [61]. Later surveys, reported in 1991 and 1993, included measurements of key parameters on 108 different pulse-echo transducers from all types of real-time scanners, including linear arrays, phased arrays and mechanical scanners operating in various modes [62,63]. The surveys were primarily of equipment in clinical use in the NHS Wessex Region and in NHS hospitals within the NHS Northern Regional Health Authority. These surveys combined measurements made using the NPL UBC system and the portable system described above. They showed that there had been a steady increase in acoustic pressures since the earliest surveys. Time averaged intensity (I_{SPTA}) was shown to have increased between two and three times over a period of 10 years. The greatest time averaged intensities were found for non-scanned modes, especially pulsed Doppler systems, whose average value was about two orders of magnitude greater than that for real-time imaging mode. Median values of total acoustic power had approximately doubled and were higher in pulsed Doppler than in imaging mode.

Surveys reported in the mid-1990s showed some further interesting trends in the acoustic output characteristics of diagnostic systems [64, 65]. These surveys made comparisons with earlier surveys based on peak negative pressure, I_{SPTA} and total power. The 1997 survey reported on measurements of over 350 different probes [65]. These were all worst case values that could be measured in water for each scanner and probe combination in each available operating mode. It showed that peak negative pressures generally had increased only slightly since 1991. However, there had been some dramatic increases in measured values of I_{SPTA}. While the mean values in pulsed Doppler mode had increased by about 20%, the mean and maximum values in B-mode had increased to match those of pulsed Doppler. A similar picture was seen with total power measurements: there was an increase in mean and maximum values in pulsed Doppler mode but a very large increase in total power values for imaging mode, resulting in little difference with pulsed Doppler mode in terms of mean and maximum values.

Independent surveys of acoustic output parameters from diagnostic ultrasound equipment carried out by Medical Physics Departments in the UK have provided useful evidence of changes that occur as ultrasound imaging technology is improved. Such measurements are important in ensuring that clinical users are able to make informed judgements on their use of equipment and avoid potentially hazardous exposures of sensitive targets. They have also served as a check on acoustic output information provided by manufacturers, in some cases identifying serious discrepancies in exposure values provided in equipment manuals [66].

VIII. Metrological challenges at high intensities and pressures

Acoustic characterisation of therapeutic systems has been shown to challenge the robustness of PVDF hydrophones [67, 68]. High intensity focused ultrasound (HIFU) was developed to treat cancers and conditions such as benign prostate hyperplasia by thermal ablation. In HIFU systems, a high power

(>100 W) ultrasound source is brought to a focus within tissue with sufficient intensity (>1000 W cm^{-2}) to raise the local temperature above 55 °C. The focal region is typically 1 -3 mm wide and 10 – 30 mm long. Characterisation of the treatment beam is important to ensure that the intended ablation temperature is reached in the treatment zone, while sparing tissues that lie between the zone and the transducer [69]. In HIFU systems, PVDF probe and membrane hydrophones may be exposed to high peak negative pressures resulting in damage due to cavitation at the hydrophone surface [68]. Nucleation of cavities is more likely to occur with a probe hydrophone due to the small dimensions at the tip. In addition, warming of the hydrophone may affect its calibration, and heating to beyond the Curie temperature is likely to lead to loss of polarisation. Hydrophone heating may be minimised by operating the system with a low duty cycle tone burst rather than continuous wave. Cavitation can be reduced by operating at reduced pressure levels and extrapolating to higher values [70], but this approach excludes the effects of non-linear propagation.

There have been some developments aimed at protecting probe and membrane hydrophones from damage in HIFU fields. In 2005, Zanelli and Howard of Onda Corporation, constructed a robust probe hydrophone for HIFU measurements. The probe contained a piezo-ceramic element encased in a metallic coating 20-70 μm thick. The coating provided a smooth outer surface to minimise nucleation sites for cavitation and protect the piezoelectric element. The frequency response was reasonably constant between 3 and 10 MHz. The hydrophone showed no sign of degraded performance after 30 minutes exposure to cavitation [71]. Wilkens et al. constructed spot-poled PVDF membrane hydrophones with additional protective layers to avoid cavitation damage. They used thin stainless steel foil to protect the front face of the hydrophone. This provided robust protection for the front electrode as well as an increased cavitation threshold due to the flatness of the surface. At the highest pressures, cavitation occurred at the rear surface of the membrane. This was reduced by adding a polyurethane backing, and measurements of peak rarefactional and compressional pressures up to 15 and 75 MPa respectively were performed [68].

Lithotripsy, or extracorporeal shock wave therapy, was developed in the late 1970s as a method for the disintegration of kidney stones, as an alternative to surgical removal [72]. The technique was developed in collaboration between the University of Munich and the German company Dornier GmbH. In 1980, the first cases of treatment in humans were reported [73]. Lithotripsy systems generate shock waves at the focus of a large aperture transducer with pressure amplitudes up to 10 MPa peak rarefaction pressure and 114 MPa peak compressional pressure [67], and have been shown to generate cavitation damage on the surfaces of metal sheets placed in the treatment zone [74]. Lithotripsy pressures have been measured using PVDF membrane hydrophones. However, pitting of the front surface of the membrane due to cavitation was observed after prolonged exposure in the lithotripsy field [67].

In 1993, Staudenraus and Eisenmenger of the University of Stuttgart, described a fibre optic probe hydrophone suitable for shock wave measurements in water [75, 76]. This consisted of a 100/140 μm step index silica fibre (core/cladding diameter with a step change in refractive index between the two) which was cleaved so that the end face was perpendicular to the fibre axis. Laser light transmitted along the fibre was reflected at the end face and detected on its return by a silicon p-i-n photodiode via a coupling port. The light reflection coefficient at the end of the fibre is determined by the refractive indices of the silica and the water medium. When exposed to an acoustic wave, the temporal pressure changes give rise to corresponding changes in reflectance at the end face of the fibre. This is due to changes in the densities of the water and silica, leading to changes in their refractive indices. As silica is much stiffer than water, the refractive index changes in the water dominate and are mainly responsible for the reflectance changes. Such fibre optic sensors are intrinsically less prone to cavitation as adhesion between water and the glass fibre exceeds the cohesion of water, so nucleation of cavities is less likely, even at high negative pressures. If damaged by cavitation during repeated high pressure exposures, a new tip can be formed quickly by cleaving a new end. The small diameter of optical fibre enables good directional characteristics.

Commercial versions of this type of hydrophone (FOPH 2000, RP Acoustics, Leutenbach, Germany) claim to be able to measure pressures in the range -60 to +400 MPa. They have been used to measure pressures in HIFU fields [77], and are the recommended device for the measurement of lithotripsy shock waves [78]. The fibre optic hydrophone can have a very wide bandwidth, limited only by that of the signal detection system, and high immunity to electromagnetic noise induced by the firing of the lithotripsy transducer. However, the high noise equivalent pressure (NEP) of approximately 0.5 MPa limits its usefulness in characterising the lower pressure regions of the field.

In 1997, Beard and Mills of University College London, described an alternative form of fibre optic hydrophone that achieves sensitivity comparable to that of a PVDF hydrophone by making use of a

Fabry-Pérot interferometer (FPI) mounted on the fibre tip. The FPI consists of an optical cavity with two reflecting surfaces. Light from a laser source is incident on the FPI via the fibre and is multiply reflected by both mirrors, interfering at the inner surface of the cavity. Cancellation occurs when the phases of the reflected and incident light are opposite, resulting in minima in reflectance (Fig. 17). The Beard and Mills hydrophone consisted of a thin polymer film (~50 μm) mounted at the tip of a 50 μm optical fibre [79]. The inner surface of the film was coated with a 40% reflective aluminium coating and the outer surface with a 100 % reflective coating. An incident pressure wave produces a linear change in the optical thickness of the polymer film. The resulting optical phase shift $d\varphi$ is converted to a reflected optical power dP_r via the intensity-phase transfer function (ITF) of the interferometer (Fig. 18). The device can be adjusted to work on the slope of the ITF (the optimum phase bias point) by tuning the wavelength of the laser light source and the reflected optical power used to obtain a measurement of pressure in the ultrasound wave [80].

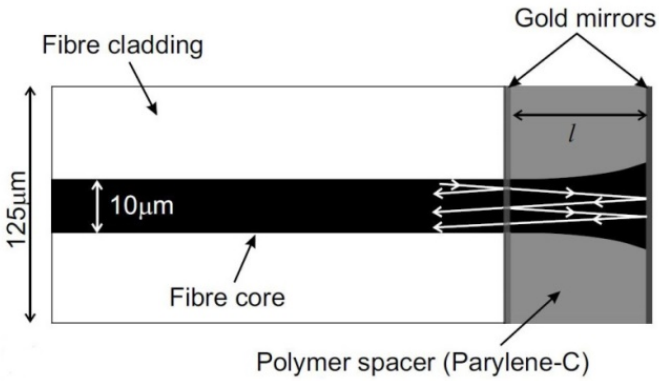

Fig. 17 The sensor head of a Fabry-Perot miniature optical fibre hydrophone. (Reprinted with permission from Morris P, Hurrell A, Shaw A, Zhang E, Beard P. A Fabry–Pérot fiber-optic ultrasonic hydrophone for the simultaneous measurement of temperature and acoustic pressure. *J Acoust Soc Am* 2009; 125: 3611–3622. Copyright 2009, Acoustic Society of America.)

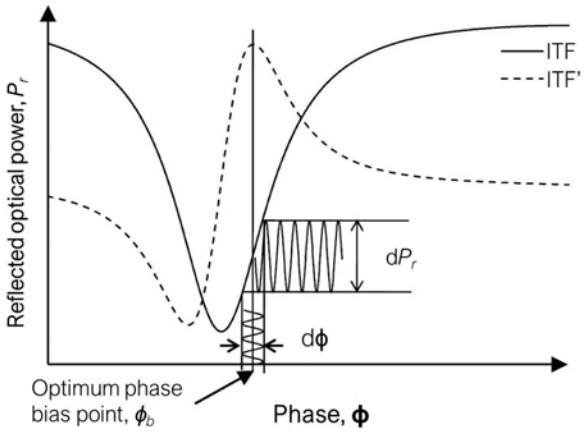

Fig. 18 The intensity-phase transfer functions (ITF) for a Fabry-Perot interferometer. (Reprinted with permission from Morris P, Hurrell A, Shaw A, Zhang E, Beard P. A Fabry–Pérot fiber-optic ultrasonic hydrophone for the simultaneous measurement of temperature and acoustic pressure. *J Acoust Soc Am* 2009; 125: 3611–3622. Copyright 2009, Acoustic Society of America.)

The Fabry-Perot device achieves much enhanced sensitivity over the simple fibre optic hydrophone as well as having small element size and wide bandwidth. A device with a Noise Equivalent Pressure (NEP) of 15 kPa, an acoustic bandwidth of 50 MHz and an element size of 10 μm was described by Morris et al. [81] The small diameter and wide bandwidth of the FP hydrophone make it potentially more suitable than PVDF membrane hydrophones for characterising high frequency, focused ultrasound fields. The fact that the sensor is at the end of a narrow fibre, make it suitable for invasive measurements. Coleman et al. used a FP fibre optic hydrophone to measure the acoustic pressure

within the ureter in 4 patients undergoing clinical extracorporeal shock-wave lithotripsy [82]. However, FP hydrophones of this construction are not considered to be sufficiently robust to the high pressures generated by HIFU and lithotripsy devices in water.

IX. Summary

The development of the piezoelectric hydrophone to detect sound waves in water began in underwater acoustics during the 1914-18 war, driven by the needs of submarine warfare, and by the end of the 1939-45 war, a range of hydrophones had been designed for naval use. In the 1950s, hydrophones small enough for use in biomedical research and medical applications of ultrasound were developed. At this time, the main medical applications were in physiotherapy and surgery, and hydrophones were used to check the timing regimes of treatments, while exposure levels were assessed via acoustic power measurement. It was not until the 1970s that hydrophones were considered for measurement of acoustic field quantities such as pressure or intensity. The main history of the development of the science and technology of hydrophones as pressure measurement devices in medical ultrasound fields began in the late 1970s with piezo-ceramic devices. However, the shortcomings of such materials in this application were soon obvious and they were quickly superseded in the early 1980s by hydrophones based on the piezoelectric polymer material PVDF. Since that time, PVDF membrane and needle hydrophones have become established as industry standard devices for medical ultrasound field characterisation, and their performance and reliability has steadily been improved. Such hydrophones have been widely used by industry and health services to characterise the acoustic emissions from medical ultrasound equipment and ensure its safety.

Advances in the performance of medical ultrasound technology over the last few decades were achieved partly by the use of higher pressure amplitudes, resulting in strongly non-linear propagation in water, the normal measurement medium. The resultant distortion of the pressure waveform and generation of high frequency harmonics presented new challenges for hydrophone measurements. These have been met by improvements in PVDF hydrophone frequency response and smaller sensing elements, and by the use of deconvolution methods to extract the pressure waveform from the measured hydrophone voltage waveform. Traceable and repeatable measurements of acoustic pressure in medical ultrasound fields are now possible using PVDF membrane and needle hydrophones.

In the last two decades, new hydrophone technologies have emerged. Hydrophones based on the use of optical fibres offer the possibility of measurement devices with smaller sensing elements and extended frequency response. The Fabry-Perot fibre optic hydrophone can be made with a 10 μm sensing element with sensitivity comparable to that of a PVDF hydrophone. The simple, bare fibre optic hydrophone, although having lower sensitivity offers a much more robust device for measurements in HIFU and lithotripsy fields and is less prone to cavitation, avoiding the risk of serious damage to much more expensive PVDF hydrophones.

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The biographical profile of Francis Duck is given elsewhere in this issue: Ultrasound - the first fifty years.