MEDICAL PHYSICS
International

EDITORIAL

A HISTORY OF MEDICAL ULTRASOUND PHYSICS – PART I

INTRODUCTION

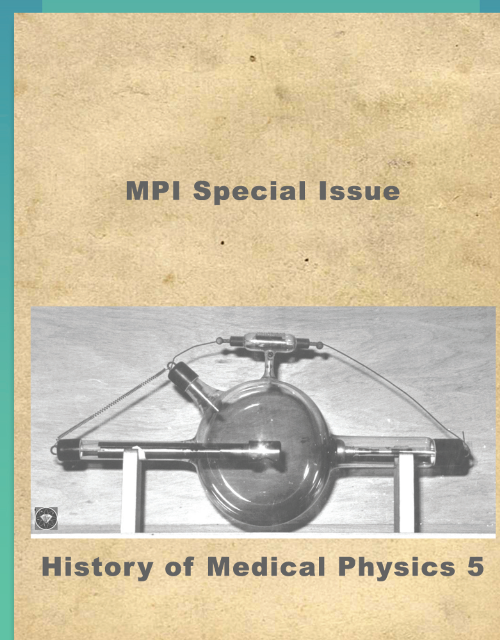
ULTRASOUND – THE FIRST FIFTY YEARS

ULTRASONICS METROLOGY:

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

II. THE HISTORY OF THE MEASUREMENT OF ACOUSTIC POWER AND INTENSITY USING RADIATION FORCE

III. THE DEVELOPMENT OF THERMAL METHODS FOR ULTRASOUND MEASUREMENTS



IOMP

The Journal of the International Organization for Medical Physics

Special Issue 5, January 2021

MPI

MEDICAL PHYSICS INTERNATIONAL

THE JOURNAL OF

THE INTERNATIONAL ORGANIZATION FOR MEDICAL PHYSICS

MEDICAL PHYSICS INTERNATIONAL Journal, Special Issue, History of Medical Physics 5, 2020

MEDICAL PHYSICS INTERNATIONAL
The Journal of the International Organization for Medical Physics

Aims and Coverage:
Medical Physics International (MPI) is the official IOMP journal. The journal provides a new platform for medical physicists to share their experience, ideas and new information generated from their work of educational, professional and scientific nature. The e- journal is available free of charge to IOMP members.

MPI web address: www.mpijournal.org

Published by: The International Organization for Medical Physics (IOMP), web address: www.iomp.org ; post address: IOMP c/o IPEM, 230 Tadcaster Road, York YO24 1ES, UK.

ISSN 2306 - 4609

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EDITORIAL

Slavik Tabakov, Perry Sprawls and Geoffrey Ibbott

MPI Special Issues Co-Editors

In this Special Issue dedicated to Medical Physics History includes four new articles from the IOMP History Project covering the history of Ultrasound, a very important field of our profession. Part I published here begins with an introduction of Ultrasound into our lives and in medicine followed by the evolution of Ultrasonic measurements which are now an essential part of Ultrasound Quality Control. Part II of this History of Ultrasound will be published early next year. The Ultrasound part of the IOMP History project has been developed by a team of British IPEM scientists, led by Dr. Francis Duck, MBE. Future articles are planned from a wider international authorship.

The History Project is growing in popularity with each of the past Special Issues having more than 10,000 downloads. This is a clear indication of the value that the profession sees in the project. All of the medical physics history articles can be accessed through: http://www.mpijournal.org/history.aspx

The topics extensively covered so far include:

Special issue 1- http://www.mpijournal.org/pdf/2018-SI-01/MPI-2018-SI-01.pdf
* X-ray Tubes Development - IOMP History of Medical Physics (R. Behling), p.8-56
* Film-Screen Radiography Receptor Development – A Historical Perspective (P Sprawls), p.56-82
* History of Medical Physics e-Learning Introduction and First Steps (S Tabakov), p.82- 109

Special issue 2 - http://www.mpijournal.org/pdf/2019-SI-02/MPI-2019-SI-02.pdf
*Fluoroscopic Technology from 1895 to 2019 Drivers: Physics and Physiology (S. Balter), p.111-141
*The Scientific and Technological Developments in Mammography (P. Sprawls), p. 141-167
*Review of the Physics of Mammography (C R Wilson), p.167-225

Special issue 3 - http://www.mpijournal.org/pdf/2020-SI-03/MPI-2020-SI-03.pdf
*History of Dental Radiography (P Rubens), p.235-278
*The History of Contrast Media Development in X-Ray Diagnostic Radiology (A Thomas), p.278-303
*Medical Physics Development in Africa (T Ige et al), p.303-317

Special issue 4 - http://www.mpijournal.org/pdf/2020-SI-04/MPI-2020-SI-04.pdf
*A Retrospective of Cobalt-60 Radiation Therapy (J van Dyk et al), p.327-351
*The Many Steps and Evolution in the Development of Computed Tomography (P Sprawls), p.351-387
*Medical Physics Development in South-East Asia (K Ng et al), p.387-399
*History of Medical Physics Education and Training in Central and Eastern Europe (S Tabakov), p.399-457

Additionally, we have published summative papers related to the development of medical physics in the Middle East (A Niroomand-Rad et al, MPI vol.5 No.2, 2017) and in Central America (W Chanta et al, MPI vol.7 No1, 2019). In the MPI Issue of June 2020 we included a paper related to the History of IUPESM. In the coming regular issue of the MPI Journal we include a paper describing the new activity of the AAPM History Committee – the Virtual Museum of Medical Physics, as well as a paper about the History of AFOMP.

The Content of the MPI Special History Issues supports the objective of the History project: to research, organize, preserve, and publish on the evolution and developments of medical physics and clinical applications that are the foundations of our profession.

We welcome contributions of colleagues from all societies, organizations and companies who would like to join the History project with articles on specific topics. We look forward to receiving your suggestions.


Prof. Slavik Tabakov


Prof. Perry Sprawls


Prof. Geoffrey Ibbott

A HISTORY OF MEDICAL ULTRASOUND PHYSICS – INTRODUCTION

Francis Duck
Formerly University of Bath and Royal United Hospital NHS Trust Bath UK

The four articles in this issue of Medical Physics International form the first of a series intended to document the contributions of physicists and engineers to the application of ultrasound to clinical medicine. They are part of the broader initiative of the International Organisation of Medical Physics to document the history of medical physics in all its aspects. I would like to thank Slavik Tabakov most sincerely for his original invitation to participate in this project and his quiet guidance and support in reaching this stage.

Ultrasound scanning now probably contributes at least 30% of all medical imaging worldwide. By 2014, the year that the UK NHS stopped gathering imaging statistics, the number of ultrasound scans in England was approaching ten million, of a total imaging of 43 million, well exceeding the combined totals of CT and MRI. It is a technology that is used far beyond the confines of departments of imaging and radiology. The technology is so ubiquitous that it has been suggested as a replacement for the stethoscope for every junior doctor.

How have we reached this astonishing position? First and foremost it is because ultrasound scanning is clinically useful. Many new medical technologies never emerge beyond the headline-grabbing launch phase, and others only find permanent homes in niche areas of medicine. Not only is ultrasound widely diagnostically valuable, it is cost-effective, safe, small-scale and, in particular, it is kind to the patient.

The articles selected for this issue do not describe the long, slow development of the techniques that are now part of modern clinical ultrasound, including Doppler imaging, elastography, harmonic imaging and all the rest. Articles covering some of these topics are planned or in preparation and will come later. Instead, first, we document the first fifty years of ultrasound up to 1950, during which a few pioneers explored its destructive power, and the only serious established medical application was at the end of this period, for therapy. It was a time that encompassed the two world wars, both driving developments in ultrasound that were necessary before medical uses could follow. There were contributions of physicists from many nations during this period: Wilhelm Altberg, Paul Langevin, Robert Boyle, Frank Lloyd Hopwood, André Dognon, Robert Wood, Reimar Pohlman, Floyd Firestone and numerous others.

The remaining three articles in this issue cover a central function of most medical physicists, the measurement of radiation. From the earliest years it was necessary to quantify the acoustic power, acoustic intensity and acoustic pressure in the beams being generated by the new ultrasonic transducers. The methods that evolved in the laboratory, using thermometry, radiation force and hydrophones, were given impetus once medical applications emerged. They were used for the measurement of the ultrasonic properties of tissue, for the development and testing therapeutic ultrasound systems, for quantifying high intensities for surgery and finally to ensure safe output from diagnostic ultrasound equipment.

These measurement techniques now underpin all medical uses of ultrasound. Manufacturers must ensure calibration and safety, set by international and national standards. National standards laboratories establish reference measurements, cross-calibration honing precision. Medical physicists make measurements to evaluate conformance and stability of output, and to educate clinical colleagues. Modern ultrasonic metrology is based on the slow evolution that is described in these articles.

Roland Blackwell (1943-2017) introduced me to ultrasound in 1966. We had both been appointed as junior medical physicists by John Clifton at the University College Hospital Medical Physics Department in London. My job was in nuclear medicine and his was in ultrasound, particularly to support the Diasonograph, the newly installed ultrasound scanner in the basement. Before solid-state electronics, I gave a hand when valves failed and needed replacing. We sneaked in after hours to scan my wife Di, expecting our first son Roger, in November 1967. I still have the Polaroid photograph, a starkly black-and-white image in which the head and thorax can just be made out. He arranged our visit to see Kit Hill at the Institute of Cancer Research, to use his balance to measure the power output for our MSc Doppler projects. Roland's central role as a leader and educator in the growth of ultrasound has not gained as much recognition as it should, and I am very pleased to dedicate these historical articles to his memory.

ULTRASOUND – THE FIRST FIFTY YEARS

Francis Duck

Formerly University of Bath and Royal United Hospital NHS Trust Bath UK

I. Introduction

The application of x-rays to medicine has a historical perspective that is quite unlike any other medical technology. The step change that arose from the discovery of X-rays by Röntgen, announced in January 1896, and in its immediate application by medical doctors for clinical diagnosis, was highly unusual. The time-line for ultrasound is closer to the more common, slower ingress of a technology into clinical medicine. A new phenomenon, infrared or ultraviolet radiation for example, is first discovered and its properties explored in academic laboratories of physics, where a greater understanding is developed. In due course, typically as a result of a medical doctor working in partnership with an engineer or scientist, clinical applications are explored. The first serious investigation of ultrasound occurred at the end of the extraordinary final decade of the nineteenth century, which also saw the discovery of x-rays, radio waves and radioactivity, and the first use of ultraviolet waves and of high frequency currents for medical applications [1]. It would be nearly half a century before ultrasound started to emerge clinically as a potential therapeutic agent, followed shortly by its development for diagnosis and imaging.

In this article I shall trace the most important milestones during these fifty years, as ultrasonic technologies were developed for other purposes before they started to be explored for medical applications, and as the effects on living organisms were investigated, so giving a basis for the development of ultrasonic therapies and also a framework for the safe exploitation of ultrasound for medical diagnosis.

II. Air-borne ultrasound

Physiological acoustics, the investigation of hearing, was an area of active investigation during the nineteenth century, a discipline within which ‘Sensations of Tone’ by the physiologist and physicist, Hermann Helmholtz, first published in 1862, was the dominant text. It was well appreciated that the ear was limited in the range of frequencies to which it could respond, and there were several early attempts to measure the upper limit, notably by physicist César Depretz in Paris and by the English-born psychologist William Preyer working in Germany [2,3]. There was little quantitative agreement on the upper limit to human hearing, however, the reported age-dependent threshold between audible and ultrasonic frequencies lying between about 16 kHz and 40 kHz [4].

Rudolph Koenig’s publication in 1899 marks the first serious exploration of ultrasonic waves in air [5]. Koenig demonstrated that ultrasonic waves up to 90 kHz could be generated using a series of small tuning forks and steel bars only a few millimeters long. However, they were not loud enough for audiological testing and Max Edelmann then used an improved Galton whistle to generate louder ultrasonic frequencies up to 110 kHz [6,7]. Eventually, using a series of careful measurements, Franz Schultze confirmed Preyer’s estimate of 20 kHz as a reasonable upper threshold for human hearing, so setting the lower limit of the ultrasonic spectrum that remains generally accepted today [8].

Wilhelm Altberg (1877-1942) was born in Belarus, completing his schooling in Smolsensk. After graduation from Moscow University he continued working there under the guidance of the physicist Pyotr Lebedev, carrying out two key experiments in the history of ultrasound. Lebedev had been the first to demonstrate, in 1899, the existence of radiation pressure in an electromagnetic wave, as predicted by Clerk Maxwell. In 1903, the year after Lord Rayleigh’s analysis of the relationship between energy density and radiation pressure in acoustic and electromagnetic waves [9], Altberg was the first to use radiation pressure to measure the intensity of a sound wave [10,11]. This work marks the first occasion in this story when a key experiment is directly relevant to later developments in the physics of medical ultrasound. Radiation force is now the primary means to measure acoustic power in beams used for therapeutic and diagnostic applications of ultrasound, and details are given elsewhere in this history. It also forms the basis of radiation force elastography [12].

In 1907, Altberg secured his place in the history of ultrasound by his discovery that the ‘singing arc', described by William Duddell [13], could also emit acoustic waves at ultrasonic frequencies [14]. In his delightful acoustic experiment, he investigated the spectrum of the pulsed acoustic waves generated by a spark discharge, using a diffraction grating made from glass rods. He measured the diffraction patterns using the radiation force exerted on a 4x12 mm mica vane suspended in a draft-proof box by a quartz thread, observing its displacement by the radiation force. From the angle of the main diffraction lobe he concluded that he was able to generate wavelengths as small as 1 mm, in other words, frequencies in air of a little over 300 kHz. One further observation completes this preliminary overview. In 1911, Lebedev and his student Neklepajev explained why Altberg had been unable to detect wavelengths shorter than 1 mm. It resulted from their absorption, for which the coefficient is dependent on the square of the frequency, so placing a limit on the useful penetration of higher ultrasonic frequencies through air [15,16]. Altberg eventually settled in Petrograd (later Leningrad) where he died during the siege in 1942 [17].

III. Ultrasound for submarine detection

The frequency-dependent absorption of ultrasonic waves in air served to limit the initial interest in the use of acoustic radiation above the audible limit, initially relegating the phenomenon to no more than a scientific curiosity. Then two men, one in Britain and the second a Russian émigré working in Austria, recognized that ultrasound had the potential to resolve a practical problem, that of underwater location.

The first documented proposal that ultrasound could be used for underwater detection was made in 1912 by Lewis Fry Richardson (1881-1953). He was born in Newcastle-upon-Tyne, England, into a prosperous Quaker family. A decade after graduation from Cambridge University, he was facing redundancy from the Sunbeam Lamp Company as it faced bankruptcy. It was at this moment that he made his single contribution to the history of ultrasound. On 15th April 1912, the Titanic collided with an iceberg south of Newfoundland and sank with a huge loss of life. He immediately submitted two patent applications in quick succession to the British Patent Office [18,19], the second proposing an underwater ultrasonic system instead of the airborne system of the first. The purpose of the apparatus was to warn a ship of its nearness to an object ahead. Whilst the provisional specification, filed on 10th May 1912, stated that the best frequency 'will probably be of the order of 100,000 complete vibrations per second', the complete specification, filed on 10 December, abandoned his idea of using ultrasound, setting the operating frequency to be '4780 or more complete vibrations per second'. Resetting the frequency into the audio range altered his approach to directionality, now achieved using a rectangular aperture, creating a fan beam to avoid reflections from the seabed. There is no evidence that any instrument was ever constructed. Indeed, this episode in Richardson's life is considered so unimportant in his career that the two patents are not even included among the 127 publications listed in his biography [20].

It was left to the impetus from submarine warfare to cause the development of the first operational ultrasound pulse-echo location system. Details of these developments have been drawn from several sources [21,22,23].

The sinking of three British cruisers in September 1914 by submarine-launched torpedoes served to emphasize the huge threat they posed to Allied shipping. A young Russian electrical engineer, Constantin Chilowsky, was convalescing from tuberculosis in a Swiss mountain hotel in 1915 when he developed his plan for an ultrasonic system for submarine detection. He had been politically active in Moscow and his family had helped him to escape to Switzerland. His proposal was to use a magnetically-driven diaphragm as a source of ultrasound instead of the underwater whistle suggested by Richardson.

His proposal passed over several academic and diplomatic desks before Paul Painlevé, Minister of Public Education and Inventions, passed it on to Paul Langevin (1872-1946) in February 1915. The renowned French physicist, pacifist by inclination, had been initially unwilling to be drawn into war activities, but he had been encouraged by his past student, Maurice de Broglie, who was working for the navy on radio communications, to find a way to detect enemy submarines.

During the next two years, Langevin's team slowly developed a working pulse-echo system in his laboratory in the *École municipal de physique et chemie industrielles*, and at the French naval base at Toulon. They replaced Chilowski's loudspeaker with a 'singing condenser' as transmitter. They drew on electronic techniques designed for wireless telegraphy for the French navy, and one of the designers, Capitaine de Frégate Colin, was loaned to the project. The condenser transmitter is shown in Figure 1. It consisted of a circular metal plate, insulated by a thick coating of resin, on which a sheet of mica was held under a vacuum using a Gaede rotary vacuum pump, and was free to vibrate. Sea-water formed the other electrode. By July 1915, the team was able to generate ultrasonic intensities around 100 mW cm^{-2} measured using radiation force. They were unable to make the transducer work as a receiver, and designed a special carbon granule hydrophone, based on well-established carbon microphone technology for audio frequencies. By April 1916, they filed a patent on this system [24] shortly before Chilowsky and Langevin parted company.

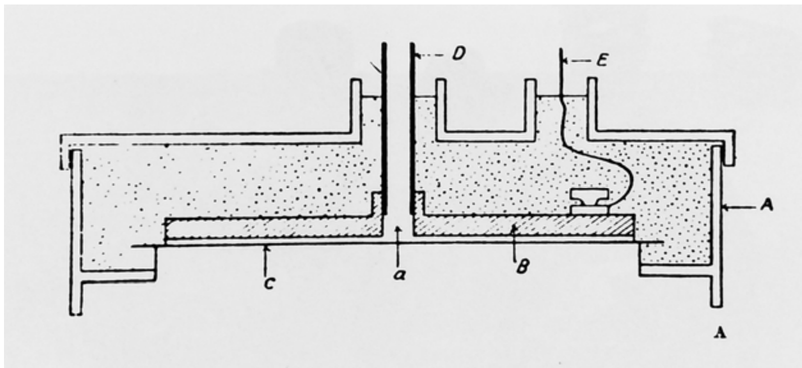

Fig 1. Langevin's singing condenser transducer c. 1916, consisting of a circular metal disc (B), mica sheet (C), vacuum tube (D) and box (A) filled with insulating compound.

On 17 Feb 1917, when Maurice de Broglie visited London to present Langevin's report to the British Board of Inventions and Research (BIR) he declared that the methods described

'represent the result of two years' continuous work, during which other methods have been tried over and over again without success. Though it is probable that they may be eventually improved, it appears certain that time will be saved by not departing (for the present) from the constructional details hereafter described. ' [25]

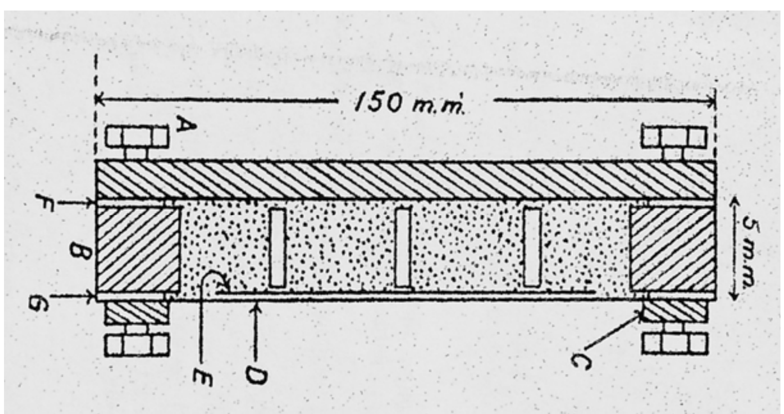

Fig 2. Experimental large area carbon hydrophone. Langevin 1917. The carbon granules are packed in water between a brass disc A and a sheet of mica D, clamped onto an ebony ring B. A steel plate E is cemented inside the mica. Glass panels prevent the carbon becoming too tightly packed.

The capacitor transmitter emitted 50 ms pulses of 100 MHz ultrasound, at a rate of 2 Hz. The carbon microphone was placed at the focus of a parabolic mirror, 120 cm diameter, to receive the echoes. The envelope of the received signal was demodulated and detected audibly, valve amplifiers being used at both the high frequency and audio stages to increase sensitivity. Nevertheless, they had been unable to build robust transducers with reproducible characteristics. Sparking could not be excluded and each new transmitter had 'but an extremely short life' [26]. The sensitivity of the carbon microphone was found to alter with depth and movement of the sea and a new large-area carbon hydrophone, 15 cm diameter, was under development that it was hoped would be more stable (Figure 2). So, while the potential of ultrasound as a means for undersea detection and communication was by then established, a reliable means to generate and receive ultrasound signals of sufficient strength still remained elusive.

By the time of de Broglie's visit, Britain had been operating its own research programme for about 6 months. Ernest Rutherford was a member of the BIR advisory panel for submarines and wireless telegraphy. At this stage, the preferred option for submarine detection was to use hydrophones to detect audible sounds emitted by the submarine and, indeed, this remained the preferred option throughout the war. Two other panel members, Richard Paget and William Duddell had visited France in June 1916 to learn more about the work of Langevin's team, returning to encourage Rutherford to investigate the use of ultrasound. As a result, in August, the Canadian physicist Robert Boyle (1883-1955) who was already working for the BIR in Manchester, was placed in charge of a new project to investigate underwater ultrasound, assigned the task to reproduce and test the Chilowsky/Langevin system.

A. Quartz ultrasound transducers

Once appointed to the ultrasound project, Boyle worked initially in Duddell's testing laboratory in South Acton, London, eventually moving to the Admiralty Research station at Parkeston Quay, Harwich. During the latter months of 1916, assisted by S G Brown FRS, Boyle did his best with limited resources to verify and develop some of the French results. They concentrated on the design of carbon hydrophones for reception and needed a calibrated source to test sensitivity. They had previously rejected quartz as a primary hydrophone material and also as a replacement for the French capacitative transmitter, but were considering it as a test source for hydrophone response.

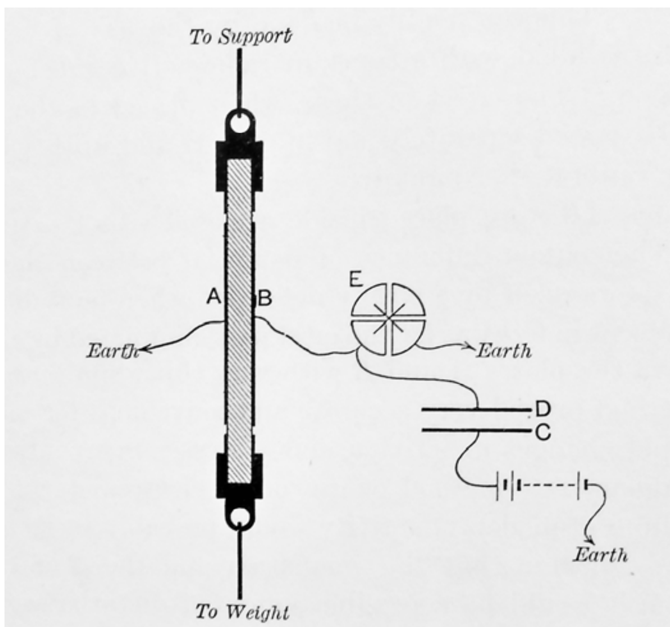

Fig 3. The quartz piézo-électrique for measurement of ionization. Piezoelectric charge across the quartz AB is balanced against the capacitor charge CD using the electrometer E, radioactivity measured by the discharge time for a fixed weight. Rutherford *Radioactivity* (1904).

On 12 December, Boyle wrote to Rutherford, 'We have done what we could to get some measurement of the minimum amplitude able to affect our high frequency detection hydrophones, using your quartz piézo-électrique to give us small measureable(?) amplitudes' [27]. The 'quartz piézo-électrique' was a slice of quartz cut perpendicularly to the electric axis, widely used in Rutherford's laboratory for the measurement of radioactivity by transducing charge from ionization into force, measured by weighing. (Figure 3). This method had been created by Jacques Curie, and used by Marie and Pierre Curie to quantify the radioactivity of radium some twenty years before.

Almost immediately after de Broglie had told the British team that the French design was complete, Langevin finally realized the key to exploiting quartz for transmission. Langevin had worked as a PhD student under Pierre Curie, and so understood the difference between the X-cut quartz, used by the

Curie brothers in their investigations of piezoelectricity in 1880, and the slice orientation used in the quartz piézo-électrique. In his own words 'I thought (February 1917) of utilizing the piezoelectric properties of quartz, first of all for receiving.' adding, significantly 'Instead of the deposition of it made by Pierre Curie in his apparatus for electrostatic measurements …. I thought it preferable to make use of the compression of the quartz in the same direction as its electrical axis, instead of by traction in a direction perpendicular to the electrical and optical axes. ' [21]

The first step was to replace the large-area carbon microphone with a quartz plate. Langevin obtained a quartz crystal of sufficient purity and volume from Ivan Werlein, an established Paris supplier of optical instruments. He persuaded Werlein to cut slices, 10 cm x 10 cm, from a 25 cm quartz crystal showpiece in his shop window. His first quartz receivers were 'composed of a thin sheet (of) quartz [or several plates in juxtaposition], placed perpendicularly to an electrical axis, and in contact with the water on one side (either directly, or by the medium of a thin protective sheet attached, mica for example); and on the other side with an insulated metallic plate forming the internal armature of a condenser in which the quartz formed the dielectric, and the salt-water the external armature, this condenser forming part of an attuned oscillating circuit.'

The received 100 kHz signal was connected to a tuned multi-stage triode amplifier which had just become available as part of a separate French research programme by the *Radiotelegraphie militaire*. The first results immediately showed promise and, at the end of March, Langevin visited London to discuss further collaboration between the two groups. The subsequent close working relationship was undoubtedly facilitated by Langevin's fluency in English, and Boyle's competence in French, having spent many years in French-speaking Montreal. At the beginning of May a joint Anglo-French mission set off to the USA to discuss co-operative working to support the Allied war efforts. This initiated several programmes of new work in the USA, notably at the New London Naval Experimental Station, at Columbia University, at the General Electric Company and at Weslyan University, the home base of Walter G Cady [23].

Fig 4. Members of the French and British teams for ultrasonic submarine detection. Summer 1917, Toulon dockyard. L to R. Capitaine de Frégate Colin, Lt R Saville, Robert Boyle, Paul Langevin, Marcel Tournier.

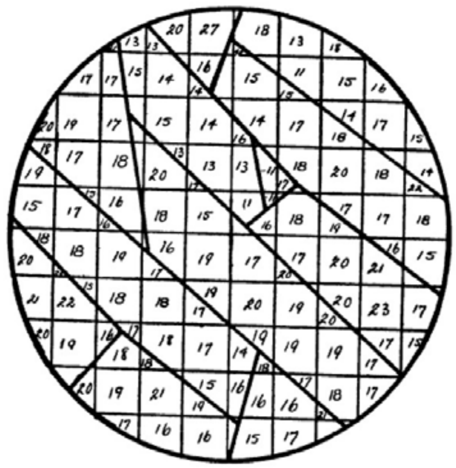

Fig 5. A quartz mosaic marked with the relative sensitivities of each element. (Boyle, [31])

When Boyle arrived in Paris on 23 May, accompanied by Lieut R Saville RNVR, he found Langevin 'about to try a quartz, piezo-electric transmitter' [28]. They visited Langevin's laboratories and the workshop where the quartz plates were being cut, tested and mounted. They then continued to the Arsenal at Toulon where they took part in the experiments there (Figure 4). By then, the French team had entirely replaced the carbon receivers with quartz, but were still working with both condenser and quartz transmitters at frequencies up to 100 kHz. The quartz was being driven at resonance by two 50 W valves in parallel, pulsed to evaluate pulse-echo detection, and continuously for underwater communications. By the time Boyle returned to England at the end of July he could report very strong one-way transmission over 500 m, with a practical limit of sensitivity of about 1000 m, attributed to refraction rather than absorption.

The limited supply of large quartz crystals led to the use of a mosaic of smaller thinner crystals. This brought its own challenges, as the cut slices had to be selected and polished to ensure uniformity of response across the whole transducer (Figure 5). By bonding the quartz to a steel substrate, ultrasonic resonance could be achieved without resorting to thick quartz plates and, for a given dimensional change, a lower driving voltage could be used.

Both the French and British teams would use these principles as their designs developed. Beyond these, their designs diverged. The French solution was Langevin's quartz sandwich transducer [29] (Figure 6). A 20 cm square mosaic of quartz plates about 4 mm thick was bonded using insulting cement between two plates of steel, each 3 cm thick, the whole sandwich air-backed. The freedom to build a transducer with a wider aperture was exploited, allowing the operating frequency to be lowered to 40 kHz.

Boyle recognized the need to move fast, even if this meant compromising design. He therefore decided against Langevin's air-backed resonant sandwich design. Instead, the British group built a simpler and more robust transducer in which the quartz mosaic was assembled between a steel plate and a mica sheet using no special bonding apart from Vaseline to exclude bubbles of air. Then the assembly was encapsulated with a waterproof and insulating mixture such as that used in cable construction (Figure 7). The amplifiers were initially obtained from France.

The system was operated initially at 75 kHz, but as Boyle pointed out in his report to the BIR, 'experiments have shown that thin sheets of quartz work well at all frequencies up to as high as have yet been tried, say, 120,000. They work "off resonance". i.e., by a "forced vibration", yet will work at any one frequency about as well as another.' [30]

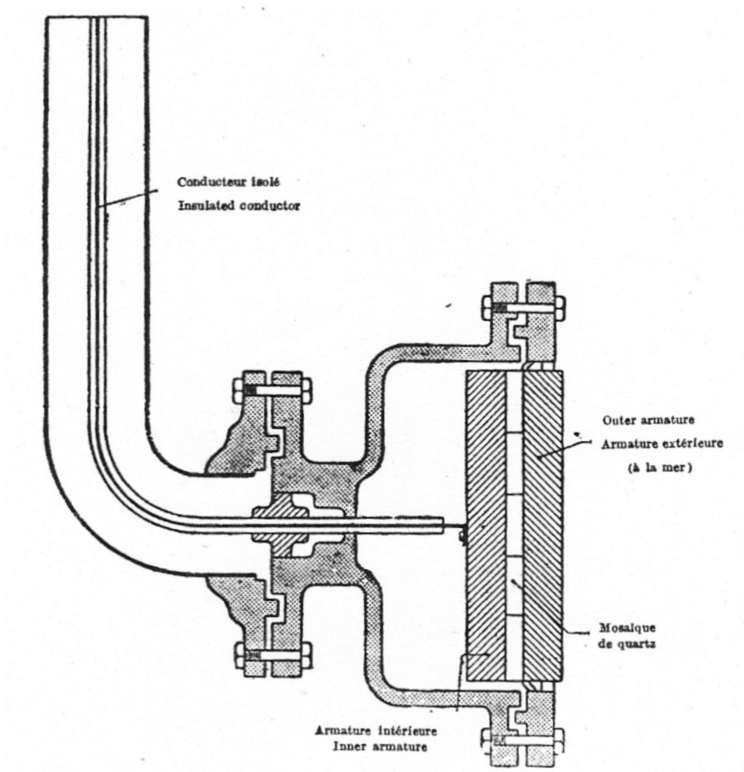

Fig 6. The Langevin-Florrison resonant quartz sandwich transducer.

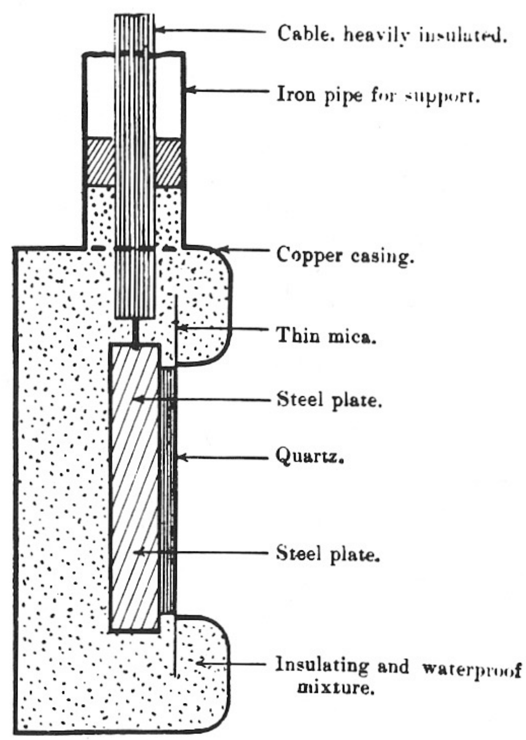

Fig 7. Boyle's broad-band encapsulated asdic transducer.

Langevin's theoretical five-fold gain from resonance would not be essential, provided sufficient attention was paid to coupling and amplification. The risk of water ingress during submersion would be eliminated, and it would be straightforward to alter the frequency of operation.

The initial arrangements were complete by October and by the end of 1917 Boyle's team had successfully transmitted between the ship *Heidra* and M.L.14, using their non-resonant transmitter to explore range at 45 kHz, 80 kHz and 127 kHz. Even at the highest frequency the range was nearly a mile, about 1.5 miles otherwise, including tests with a repaired mosaic transducer on loan from Langevin operating at 58 kHz.

It took until the February 1918 before the first signaling trials were commenced in France, when a signaling distance of 8 km was achieved. Operating in pulse-echo mode, clear echoes were obtained for the first time from a submarine. Boyle was not far behind. By March 1918, Boyle had successfully detected echoes from a submarine at a distance of 500 yards using the same transducer for transmission and for reception. By mid 1918, the French system had demonstrated submarine detection with a range extending occasionally to 1500 m.

The British stopped using the words quartz and supersonics. The term 'asdics' was introduced, for secrecy, an acronym apparently derived from 'Anti-Submarine-Detection-ics'. Quartz was known as 'asdevite'. The 'asdephone' was developed using quartz as a low-frequency listening device.

The war was almost over when both teams attended the Conference on Detection of Submarines by the Method of Supersonics in Paris in October 1918, to share their progress with each other and the four-man contingents from the USA and Italy. Although sea trials were underway, peace was declared before any navy used ultrasonic detection of submarines in action.

IV. ULTRASONICS AND SUPERSONICS 1920-1940

It was not until the 1939-45 war that Sonar re-emerged as an effective anti-submarine technology. The inter-war decades were characterized by progress in three areas of ultrasonics: the academic study of the ultrasonic acoustics, the exploration of possible applications of high-intensity ultrasound in engineering, chemistry and biology, and the development of underwater sensing for commercial and naval purposes. Each created knowledge and techniques that would later underpin the medical use of ultrasound, and the following sections outline some of the main developments

Robert Boyle declined an American offer to work there on ultrasound after the war, and failed to find an academic post in Britain. He returned to the University of Alberta, Canada, where he spent the next ten years working on the fundamental physics of ultrasound [31].

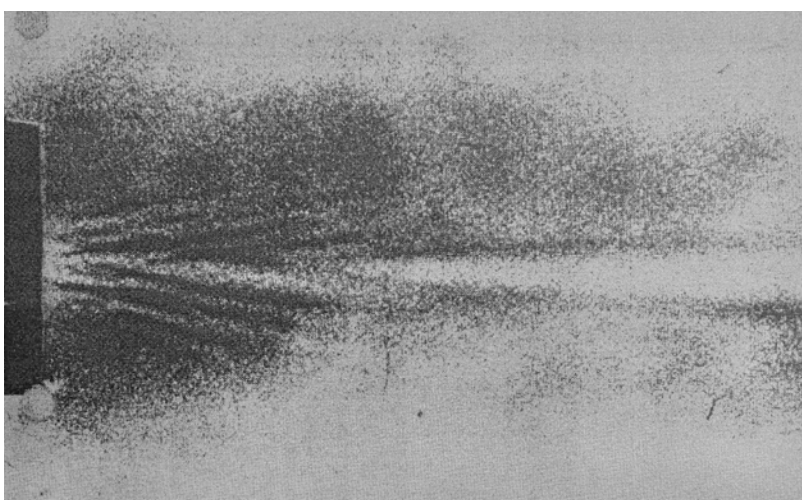

Fig 8. Boyle's dust images of the field of an ultrasound beam from a 178 kHz, 15.3 cm diameter circular source. [31] 1928.

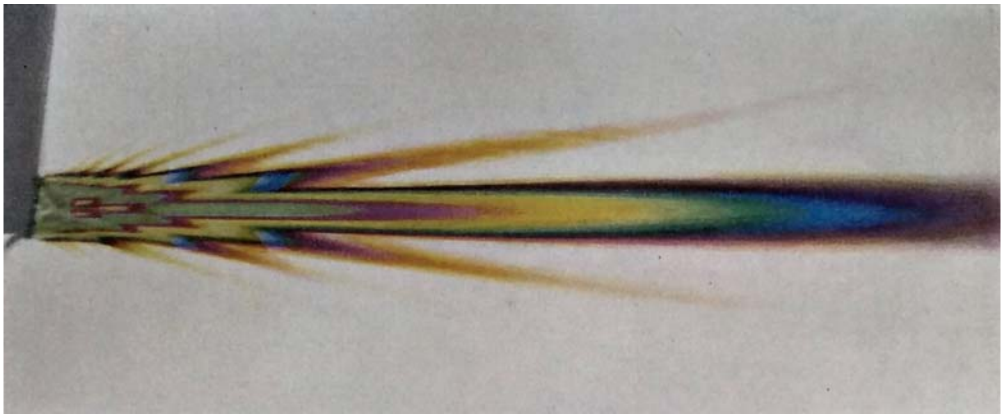

Fig 9. Beam from a source, 11 λ diameter, in C_6H_6. Hiedemann & Osterhammel 1938 [33].

His transducers were quartz mosaic slices bonded to metal plates and fully encapsulated with resin. Some were of the Langevin sandwich design, and some were piles of interleaved quartz and metal plates. In each case the resonance was determined by the overall thickness being λ/2. A simple detector hydrophone was built from a small capsule with mica sides connected to a stethoscope, which generated an audible tone at the pulse repetition frequency. Using fine dust powder, 'jostled from the regions of maximum to the regions of minimum energy density' whilst sinking onto a tray below the beam, he formed images of stationary waves, near-field and side-lobe diffraction patterns, two-beam interference and beam reflection (Figure 8) [32]. Within a decade, such crude approaches to beam visualization had been superseded, notably by the use of schlieren methods in the hands of Hiedemann and his co-workers in Cologne (Figure 9).

During the summer of 1928, Boyle visited several laboratories in Europe and recruited Donald Sproule from Hughes. Boyle and Sproule used radiation pressure to measure the relative axial and radial intensity profiles, using various designs of torsion pendulum. Using these as receivers, they measured transmission through thin layers, confirming that when the plate thickness was an integral number of half wavelengths nearly all the incident energy was transmitted [34]. Boyle also investigated velocity dispersion in liquids, concluding that any alterations of speed over the frequency range 30 kHz to 600 kHz were beyond the accuracy of his experimental measurement methods to observe.

In all, Boyle published twenty-five papers on the basic physics of ultrasound in the decade from 1922 to 1932. The first of these was on acoustic cavitation, and was probably the first academic paper to address this topic [35]. His interest arose from the association between bubble formation and the loss of sensitivity that he and Langevin had observed when trying to operate ultrasonic underwater systems at high power. Boyle correctly explained acoustic cavitation as the rupture of liquid during the rarefaction half-cycle, and also understood that this was less likely at higher frequencies. In a series of fundamental studies over the next few years he showed that there was a threshold for acoustic cavitation that increased with ambient pressure, and that it depended on the liquid, being lower in volatile hydrocarbons than in water. In studies into standing waves, he observed that bubbles migrated to nodes in stationary waves.

His studies on volatile liquids demonstrated that the released gas was largely air, and that liquids that had been so degassed showed a higher threshold for cavitation. He concluded that the liquid was not being ruptured and that cavitation was initiated as bubbles formed around dust particles. His specific conclusion was that cavitation 'tends to restrict the maximum intensity of energy transmissible'. Others investigated the increase in attenuation caused by cavitation [36]. Sonoluminescence was observed [37]. More relevant to the present topic is the association between cavitation and biological effects, and this will be addressed below.

Quartz transducers were by far the most common sources of ultrasound during the interwar years. Not only that, it seemed clear that the technology was not yet fully exploited. As upper frequencies increased towards 1 MHz and beyond, a potential upper limit was even being considered, estimated to be about 50 MHz for which a quartz slice only 0.054 mm would be required. Furthermore, it was shown that quartz could be shaped into a concave bowl, creating a focused ultrasound beam [38]. Both these ideas anticipated later developments in diagnostic imaging and focused surgery.

A. Non-linear acoustics

The narrative so far has largely concerned transducer engineering and the applications of ultrasound. The underlying acoustic principles, set out by Lord Rayleigh in 1877 in his *Theory of Sound*, had been transferred without modification to encompass the low ultrasonic spectrum, including beam structure, diffraction, refraction, scattering, and calculations of energy density and acoustic intensity. There was no need for further theory. One aspect of theoretical acoustics remained obscure, however. Acoustic wave transmission is an inherently non-linear process, both from local and cumulative effects. The resulting waveform distortion and associated harmonic generation was known, and had been analysed for a loss-less medium, but no progress had been made to include dissipation. In the 1930s a few steps were taken to resolve this difficult challenge. These formed the foundation on which, several decades later, an understanding of the relevance of non-linearity to medical ultrasound led very quickly to the successful commercial development of harmonic imaging.

In 1931, Fay published his analysis of the non-linear propagation of a periodic wave through a viscous gas [39]. His solution balanced the generation of harmonics as a result of non-linear propagation with their loss as a result of viscous absorption. The harmonic amplitudes for what is now called the Fay solution are $B_n = 2/n(1+\sigma)$ where $\sigma = x/x_s$ the distance with respect to the shock formation distance x_s [40]. Once an acoustic shock forms 'the wave forgets its origin' and the fundamental component approaches a value independent of the drive. This so-called acoustic saturation is of central importance in modern ultrasonic metrology and its use in safety standards. The explicit solution published by Fubini in 1935, is perhaps of less relevance because it is valid only for regions of weak non-linearity when $\sigma \leq 1.0$, that is before an acoustic shock has formed.

In Paris, Langevin's student Pierre Biquard (1901-1992), investigating the attenuation of ultrasound in liquids using a radiation force detector, observed an anomalous increase in transmission loss that was especially pronounced at high frequencies. He attributed this additional loss to the progressive formation of an acoustic shock, including a theoretical evaluation in his 1935 PhD thesis and in his extensive discussion of Langevin's 1923 lectures at the *College de France* [41] He gave an expression for the coefficient of non-linearity β that was very similar to the one used today, $\beta = 1 + B/2A$, in which A and B are the coefficients of a series expansion of the isentropic pressure-density relation.

B. Physical and chemical effects at very high intensities (supersonics)

A number of laboratory experimentalists started to explore the physical, chemical and biological effects of exposure to high intensity ultrasonic waves during the interwar decades. In the next sections, some of these effects will be described, with an emphasis on the biological effects which would underpin future considerations for the safety of ultrasound diagnostic devices, and its use for therapy and surgery.

When Langevin reported in Paris in October 1918, he had told the delegates that 'fish placed in the beam in the neighbourhood of the beam were killed immediately, and certain observers experienced a painful sensation on plunging the hand in this region'. Ten years later, the American physicist Robert Wood, recollected that 'when the frequency was adjusted for resonance the narrow beam of supersonic waves was shot across the tank causing the formation of millions of minute air bubbles and killing small fish that occasionally swam into the beam. If the hand was held in the water an almost insupportable pain was felt, which gave the impression that the bones were being heated.' [42]

Robert Williams Wood (1868-1955) [43] had spent two years in Berlin before settling in 1901 as professor of experimental physics at Johns Hopkins University. Well respected for his work in ultraviolet and infrared photography, his physics interests were wide ranging. By 1900 he had demonstrated how Toepler's 'schlieren' imaging could be used to visualise the propagation and reflection of the acoustic pulse caused by a spark, the precursor for modern ultrasonic pulsed schlieren imaging [44]. His physics career was built on excellent experimental science aided by a large dose of showmanship.

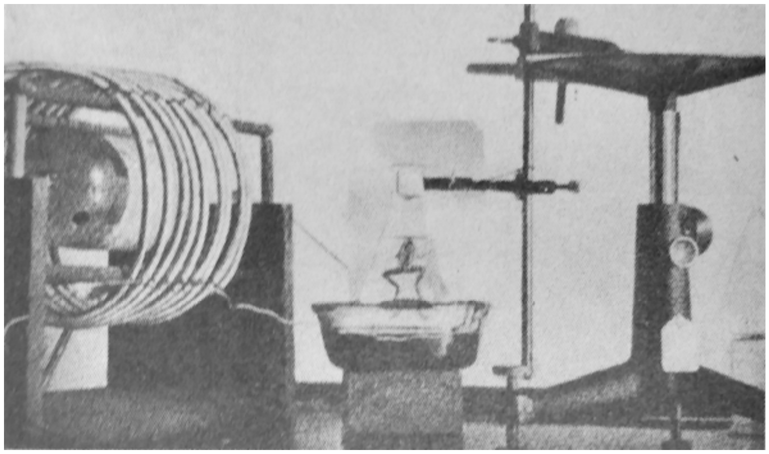

Fig 10. The coil and reaction chamber used by Wood and Loomis in their studies into the physical and biological effects of high-intensity ultrasound. [42]

Wood was known and respected in Europe. When he recalled that ‘it was my good fortune during the war to be associated for a brief time with Prof. Langevin’ …. on the submarine detection problem’, he was probably there as an individual scientist, and not part of any official visit to Toulon during the war. He made no recorded contribution to the US submarine detection programme. After the war, Wood returned to his studies in optical spectroscopy, publishing a raft of papers and improving his own 40-foot diffraction grating, probably the largest in the world at that time, which was housed in his personal laboratory. It was here that he was visited by his old friend, the financier Alfred Lee Loomis, a Harvard law graduate who had become very rich by investing in utilities and was seeking a new interest. Loomis asked Wood if he could help him by financing a study ‘which required more money than the budget of the Physics Department could supply’, no doubt assuming this would probably be an even larger optical instrument of some sort [45].

Instead, Wood suggested they should explore ‘supersonics’. In 1926, with Loomis’ money, they bought a 2 kW oscillator, originally designed for an induction furnace, added capacitative tuning to operate between 100 kHz and 500 kHz and an output transformer, to give a maximum potential across the quartz plate of 50 kV (Figure 10). No subtlety was applied to transducer design: ‘The quartz plates were circular discs, and when in operation, one of them rested on a disc of sheet lead at the bottom of a dish of transformer oil. The other electrode consisted of a disc of very thin sheet brass resting on the upper surface of the quartz. ‘ [46]

The results were spectacular. Their paper, Loomis’ first in any scientific journal, was deliberately eye-catching [47]. The purpose was to give maximum publicity to the newly established ‘Alfred Lee Loomis Laboratory, Tuxedo, N.Y.’, from which the paper heading announced that it was ‘Communication No 1’. Publication, in September 1927, was not in America, but in the prestigious *London, Edinburgh, and Dublin Philosophical Magazine.* With few exceptions the paper was descriptive, and low on explanatory science. Much was made of the high radiation forces generated at the oil/air interface, and its chaotic disruption. Variation in output due to acoustic loading was noted, but without serious analysis. Extreme temperature increases were deliberately established. Glass was melted. A fine glass needle caused a severe burn that took a long time to heal. Aggregation and emulsification was observed. Mists were generated. Stationary waves were generated in tubes. The only graph showed the change in measured velocity with frequency in glass rods and disks.

This was Wood’s only contribution to ultrasonics, apart from his later monograph ‘Supersonics’ (1939). (The term supersonics was introduced to mean high-intensity ultrasound). For Loomis, things were different. This paper launched his new career as a scientist. His strategy was to invite established scientists to work with him in the private laboratory that he built in his new home in the fashionable New York district of Tuxedo Park. Funding their equipment and attaching himself to their work, he became co-author on their papers. Simultaneous publication on both sides of the Atlantic extended the publication list and the publicity. Having equipped Tuxedo Park to carry out ultrasonic experiments, several scientists took Wood’s place to work there. Loomis published on the chemical effects of ultrasound with William Richards from Princetown. With the physicist J C Hubbard, a colleague of Woods at Johns Hopkins, he developed an interferometer for the measurement of sound speed in liquids, solutions and liquid mixtures. Hubbard published four papers with Loomis on its design and use and subsequent papers during the 1930s [48,49]. In particular, they published accurate values for the temperature and salinity-dependence of sound speed, which were important when tissue-equivalent phantoms were first being designed for the calibration of medical ultrasound scanners.

C. Ultrasonic biological effects

The first paper from Tuxedo Park included 'biological effects' in its title. It established an association between ultrasound and lethality that coloured later attitudes to its safe use. Wood called ultrasound 'death rays'. 'We worked together, killing fish and mice, and trying to find out how and why they were killed, that is whether the waves destroyed tissue or acted on the nerves or what'. In order to help them understand what was going on, they recruited Edmund Newton Harvey, professor of zoology at Princeton, sometimes working with his wife Ethel [50]. Emphasis was on the role of cavitation in cell lysis, at first using Wood's high power system and then a 75 W system more suited to microscopic studies, operating at 1.25 MHz and 2.25 MHz [51]. The results from Tuxedo Park sparked interest around the world in the effects of ultrasound, with attention focused on the mechanisms causing cell destruction [52].

Frank Lloyd Hopwood (1884-1954) was another physicist who had been introduced to ultrasound during the war, and who was studying the biological effects at the same time as Loomis set up his laboratory. Professor of physics at St Bartholomew's Hospital Medical School (Bart's) in London, Hopwood was one of the earliest physicists in Britain to devote his career to medicine, appointed as demonstrator in physics there in 1906. During the war he had worked on directional hydrophones to listen for submarine noise, introducing an air film to make them uni-directional [53,54]. The hydrophones were being tested at Parkeston Quay, Harwich, during the time that Boyle was developing asdics there. Hopwood heard about the biological effects observed in Toulon and 'determined that when circumstances permitted he would investigate the matter'.

Hopwood carried out similar studies as those by Harvey, giving an overview in 1931 in which he referred to work carried out 'during the past few years' [55,56]. In common with all early experimentalists, he used an X-cut quartz plate. This was clamped to a lead substrate with a light copper foil as the second electrode, immersed in oil. He used a 3 kW oscillator. Intensity was estimated using radiation pressure. His biological results were similar to those reported by Harvey, lysis of cells, plant cell destruction and alterations in nerve and cardiac muscle action. (Figure 11). Given the purpose of the present article, the important distinction between Harvey's and Hopwood's work was in the placing of the laboratory: in the one case in a personal laboratory designed to publicize private investment in science, and the other, a physics laboratory that was embedded in the work of a medical school. Hopwood was president of the British Institute of Radiology in 1932-33. In his presidential address he included ultrasound in a review of new medical technologies, alongside 100 MHz diathermy, infrared photography, soft x-ray microscopy and MV x-ray therapy. Nevertheless, his ultrasound studies dwelt on its destructive power, and he failed to interest his clinical colleagues in ultrasound as a therapeutic agent [57].

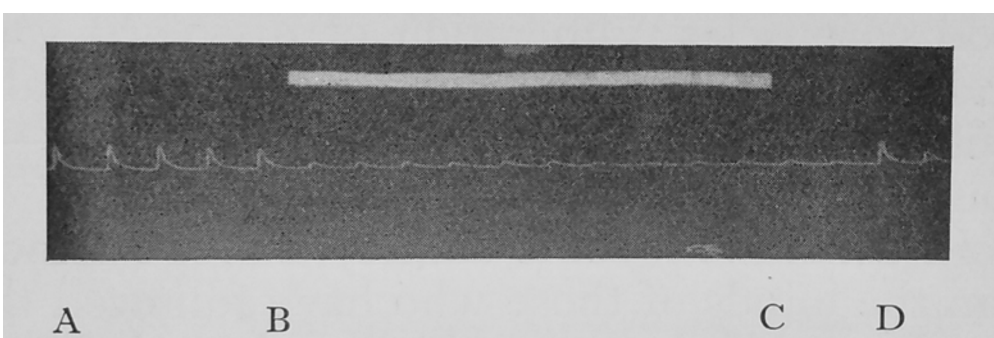

Fig 11. Inhibition of nerve conduction due to ultrasound exposure from B to C. (Hopwood, 1931). [55]

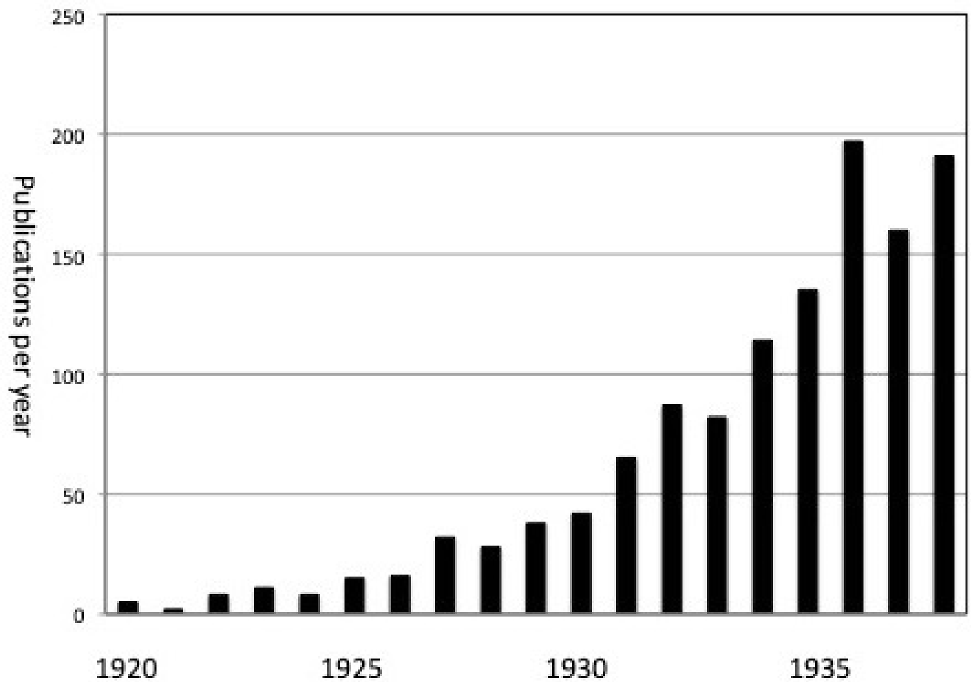

Fig 12. The growth in papers on ultrasound and associated topics from 1920 to 1938. Hiedemann 1939. [58]

Biological studies formed only one component of the growth in ultrasonic acoustics during the 1930s. (Figure 12). These years also saw rapid developments in acoustic metrology, including new devices using radiation pressure, thermal sensors and optical methods for beam visualization. There were several remarkable studies of optical diffraction and secondary interference, gaining insights into the structure of anisotropic solids. Ultrasound spectroscopy gave insights into molecular structure through studies of absorption and velocity dispersion in liquids and gasses. Physico-chemical techniques, for dispersion, coagulation, emulsion formation and degassing, were widely reported. It is not appropriate to review most of these developments, except where they form the basis for future medical applications.

V. Ultrasonic therapy

Serious studies into the bioeffects of ultrasound emerged, notably in France. The Romanian physicist Néda Marinesco had moved to Paris, where he completed his doctorate on molecular biophysics under Jean Perrin in 1927. After working on the biological effects of electric fields, he set up an ultrasound laboratory at the *Intitut de Biologie Physico-chemique* [59]. His main results were published in two parts in 1937 [60]. He constructed his two systems from second-hand and scrap equipment, one at 2 kW and the other operating at 250 W. His quartz transducers were from 2 mm to 6.5 mm in thickness, operated at their fundamental resonance of between 428 kHz and 1.43 MHz. He reported results on the destruction of micro-organisms, preparation of colloids, photochemical reactions and, most impressively, on explosive reactions. He concluded that water from the Seine could be made safe to drink after 60 m insonation at 1 kW, but it would not be practicable 'because a litre of Seine water made drinkable in this way would cost 0.25 Fr, obviously overpriced '.

Another physicist, working like Hopwood in a medical faculty, started to investigate the therapeutic potential offered by ultrasound. André Dognon (1900-1970) deserves more recognition for his role in the development of medical physics and medical ultrasound. His *Précis de physico-chimie biologique et médicale* (1929) established a reputation early in his career, when he was associate (agrégé) professor in physics at the Faculty of Medicine in Paris under the lead of André Strohl. Dognon would eventually be appointed as professor of medical physics on Strohl's retirement.

Writing in 1953, he recalled 'When, around 1930, I discussed with E. and H. Biancani the study of ultrasound as regards its biological actions, it was hardly known except by the few specialists who dealt with underwater sounding, or acoustics physicists, to whom it offered new possibilities for theoretical developments or experimental studies [61].

Dognon was offered a laboratory at the hospital *Hôtel-Dieu*, and it was here, with Elio and Hugo Biancani, that he explored the bio-medical effects of ultrasound. The Biancani brothers were already using several physical agents in their medical practice, including light, infrared and ultraviolet therapies, and were interested to explore with Dognon what ultrasound had to offer. Their work appeared in three main publications between 1935 and 1938 [62,63,64]. Unlike the other early investigators who largely built their own equipment, Dognon sensibly used the expertise that had been developed in France for underwater echo-sounding in the company SCAM, using Florisson's advice, and a generator that the company made available to them.

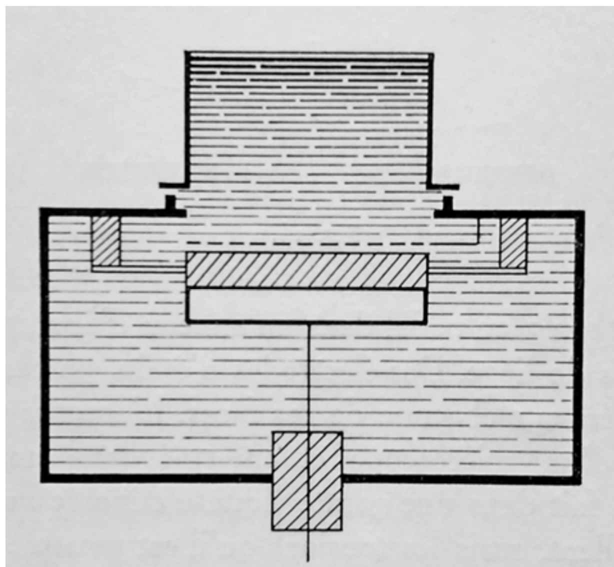

Fig 13. The French SCAM physiotherapy transducer (c.1933).

All previous investigators had immersed the quartz transducer in the insonated liquid. Instead, Dognon used Florisson's patented design, a stand-alone 'hand-held' transducer [65] (Figure 13). 'The original idea was to be able to project an ultrasonic beam in any position onto a given surface or into any container.' The quartz plate, 6 cm diameter, resonance 250 kHz, was held by three wires, with a hollow metal case applied to the back face, which served as one electrode. The upper, ground, electrode was a thin perforated sheet of aluminium. The surrounding metal box was fitted with extension tubes of various lengths, closed by a metallic or organic membrane, and the whole container filled with oil or petroleum. The maximum power was about 450 W. It was this transducer that they used for most of their experiments.

The second change was a practical approach to metrology. They argued that cavitation caused radiation pressure measurements to be unstable so, for the first time, investigated calorimetry. In a rather crude thermal sensor, they filled a glass tube with paraffin oil, and the temperature increase generated in the oil during exposure was measured with a small mercury thermometer. Whilst such an arrangement is of no value for absolute intensity, he was able to identify resonant conditions for maximum output, and was even able to plot an approximate beam profile.

Using this and other exposure arrangements they reproduced many of the phenomena observed by Harvey, Marinesco and Hopwood, although working with a somewhat lower power. Like them, Dognon observed that cavitation and mechanical agitation were the most obvious effects. But, whilst the others were aware of heating, Dognon investigated the temperature rise more thoroughly. He compared the temperature rise after a standard exposure time (10 s or 30 s) in small volumes of various liquids and soft tissues.

'Examination of these figures leads us to important remarks as to the possible mechanism of the biological action of ultrasound. The basic facts are as follows. There are substances whose heating always remains insignificant: water, gelatine gels or proteins, whatever their viscosity or cohesion. On the other hand, a whole series of other bodies, of a fatty or lipoid nature, are heated, sometimes strongly, the magnitude being linked in part to their low conductivity and specific heat, but also to their high absorption. ….. It is natural to think that the considerable heating of fatty or lipid tissues must play an important role in the genesis of physiological disturbances caused by ultrasound of great intensity. On the other hand, it is interesting to see, contrary to what one might have expected, from the theoretical evaluation of the absorption coefficient, the small influence of viscosity.'

These remarks, with their focus on absorption and heating, concern the biological action in whole tissues. Cavitation, central to the interpretation of in-vitro results, is not mentioned. Dognon considered that ultrasonic heating was the therapeutic agent [66]. 'The ultrasonic projector that we have described was precisely intended to transmit energy to the surface of the skin, and we know that this energy is transmitted very well inside the tissues.'

In fact, the idea of using ultrasound for therapy pre-dated Dognon, who noted that 'the original patents of MM Langevin and Chilowsky on piezoelectric transmitters already envisaged a possible application of vibrating baths to therapy.' Langevin had said the same to his audience during his ultrasound lectures at the Collége du France in the 1920's. The physicist Frithiof (Fred) Wolfers (1890-1971), recalled this in his review '*Les ultra-sons et la biology*' written in 1930 [67]. Much in his fifty-page treatise is drawn from notes for Langevin's course, and Wolfers noted his thanks for entrusting him with them. In a brief section on the possibility of therapy he wrote: '… it does not seem impossible, *a priori*, to obtain interesting therapeutic effects. With relatively low frequencies and low or moderate energies, and by applying the source directly to the surface of the body, it would be possible to adjust the wavelength and pressure amplitude, and therefore the pressure gradient at each point of the body. We would create in this way a kind of "internal massage" …. the creation of a real "phonotherapy" '

In a footnote, Wolfers added that this was not his idea, noting that it had been already suggested by Langevin and his student Léon Brillouin. This reference may well relate to Brillouin's 1925 patent for a method of 'indirect vibratory massage' [68].

Unsurprisingly, a few laboratories around the world started to investigate whether ultrasound had the potential to treat cancer. Studies were reported from Europe, Japan and the USA, but the results were equivocal, whether used alone or in conjunction with radiotherapy [69].

It was left to the German physicist Reimar Pohlman (1907-1978) to establish the use of ultrasound for therapy. Pohlman was a physics and chemistry graduate from Heidelberg and Berlin, and gained his PhD in 1932 [70]. In 1935, now assistant at the Physikalisch-Chemischen Institut of the University of Berlin, he had commenced research on air-borne ultrasound. A year later he published his first patent for a 'Device for the detection of defects in solid and liquid bodies by means of sound, in particular ultrasound waves' [71]. In 1939 he proposed a means for imaging the ultrasound wave that became known as the Pohlman cell [72].

He later described how, in 1938, he commenced work in ultrasound therapy [73]. He specifically rejected the high intensity approach to cancer therapy being explored at the time. Instead, he proposed that ultrasound at lower intensities could be used to stimulate healing through a combination of thermal and mechanical processes [74]. His first experiments were designed to select the most appropriate frequency to use. 'In connection with these investigations, the question of the absorption of ultrasound in human tissues, its frequency dependence, and the depth of penetration of the radiation was of major significance'. From measurements at 800 kHz and at 2.4 MHz, he concluded that he would gain greater tissue penetration at the lower of these frequencies, which he selected for the clinical evaluation. Attenuation at 800 kHz, expressed as half-value thickness for intensity, was 6.8 cm for adipose tissue, and 3.6 cm for muscle. For mixed muscle and fat, the half-value thickness at 2.4 MHz was 1.5 cm, compared with 4.9 cm at 800 kHz. He found no difference between the attenuation of tissue of adults and children, within an error limit of about 10%. He confirmed Dognon's observations that the frequency dependence of the absorption coefficient did not follow the square-law dependence expected from classical theory of viscous loss, concluding incorrectly that the lower-than-expected absorption at the higher frequency resulted from the absence of cavitation [75].

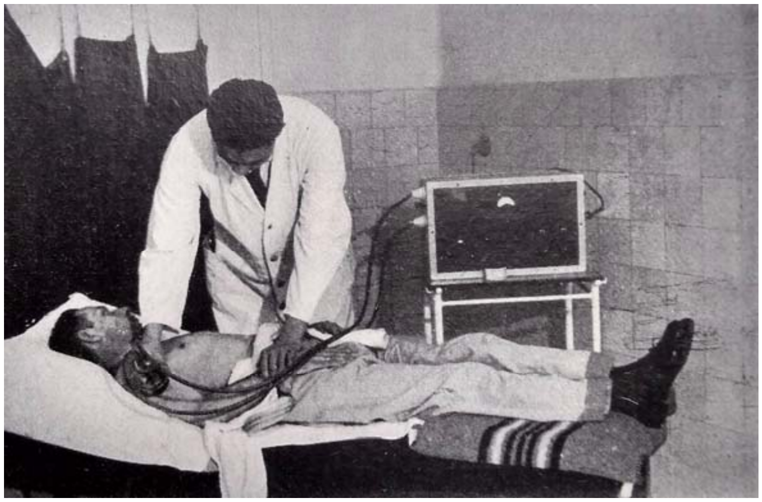

Fig 14. Clinical ultrasound therapy. Martin Luther Hospital Berlin. 1938. [73]

The first clinical trials of his ultrasound therapy were carried out before the end of 1938 in the Martin Luther Hospital Berlin-Grunewald. (Figure 14). His clinical colleagues were E Parow-Souchon and R Richter. Favourable results were reported for some neurological and neuro-muscular disorders [76].

Pohlmann described his first crude prototype transducer, and their initial approach to dosimetry:

'The radiation head was made of glass and had a thin, ultrasound-permeable membrane. This caused a certain burning feeling, and at that time the dosage was carried out in the primitive way that when the treatment head was put on locally, the intensity was regulated so that the resulting warm pain was just at the limit of tolerance. Measurements showed that this corresponds to an intensity of around 4-5 W cm^{-2}. If one stroked the body part to be treated with the massage head adjusted in this way, applying light pressure, the patient had a subjective feeling of pleasant warming, and a slight hyperaemia was noticed, which resulted partly from the warming and partly from the intense vibration.'

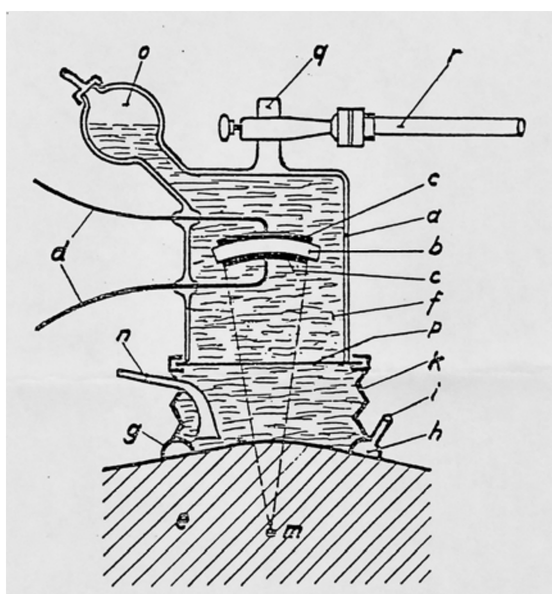

Fig 15. Siemens patented focused bowl transducer for treatment. 1935. [77].

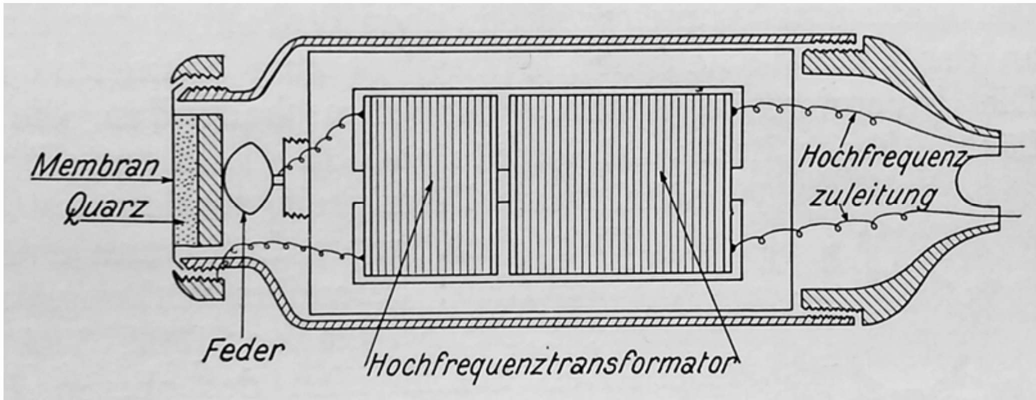

Fig 16. Ultrasound transducer for therapy. 800 kHz. Pohlmann 1939.

At the same time that Pohlmann was starting to investigate ultrasound therapy there was also interest in ultrasound treatment at Siemens-Reiniger-Werke in Berlin. In December 1935, Siemens had filed a patent for a water-coupled focused transducer for treating patients [77] (Figure 15). At this date, the bowl transducer would have had to be made from several angled and shaped crystal pieces [78]. The department responsible for this work was led by the physicist and Nobel Laureate Gustav Hertz. He had been taken under the protection of Siemens in 1934 by being appointed as overall director of Research Laboratory II, where he could continue his work in isotope separation. At the end of the war, Hertz joined a pact with other German scientists and moved to the USSR.

In 1939, following their successful clinical demonstration, Pohlmann was recruited by Siemens to take responsibility for the ultrasound laboratory in Hertz' department. Access to improved engineering support helped him to create a better transducer (Figure 16). A 4 cm^2 quartz plate was mounted in oil between a spring and an outer membrane, and a step-up transformer was built inside the transducer housing. He slightly reduced the overall maximum power to 15 W. In use, the head was massaged into the skin, using paraffin oil as a coupling material, or an anaesthetic treatment cream containing bee venom or forapin. During the war years, Siemens started to supply equipment to other clinics in Germany, who reported their early results. Pohlmann himself investigated enhanced transcutaneous transport of topical medication, with Florstedt at the Charité, Berlin [79]. Other work was reported from Berlin, Wolfsburg and Erlangen.

Siemens-Reiniger-Werke partially relocated to the comparative safety of Erlangen in 1943. Pohlman was joined by Theodor Hueter, and together they extended the measurements of the frequency-dependence of tissue attenuation up to 4.5 MHz, concluding that the absorption coefficient showed a broadly linear frequency dependence [80]. Pohlman explored alternative means to measure transmission loss, including optical [81] and thermoelectric methods [82]. He left Siemens in 1948 to move to Switzerland, and Hueter moved to the USA in 1950, to take up a position at the Massachusetts Institute of Technology.

Civilian economic reconstruction after the war favoured the growth of ultrasound therapy. Pohlmann could report 75 publications during 1949 on ultrasound therapy, and listed seventeen companies manufacturing therapeutic ultrasound equipment, twelve in Germany, four Austrian and one in France [73]. In the same year, André Denier identified three more manufacturers in France, one in Belgium and four manufacturers in the USA. The operating frequencies lay between 800 kHz and 3 MHz, and manufacturers limited the maximum intensity to between 2 W cm^{-2} and 10 W $^{-2}$. One manufacturer, Dr. Born of Frankfurt, offered pulsed operation at ratios of 1:5, 1:10 and 1:20. Ultrasound therapy had become fully established.

Fig 17. Echo-sounding in the popular press 1923.

VI. Underwater sensing and sonar

The underwater use of ultrasound, successfully demonstrated during the first war, continued to develop once hostilities ended in 1918. Langevin worked with the electrical engineer Charles-Louis Florisson to develop directional ultrasonic depth-sounding equipment, operating at 40 kHz [83] (Figure 17). The first sounding took place off Nice in October 1920. This approach had distinct advantages over other non-directional echo-sounding systems that were being concurrently developed that operated in the audible range of frequencies. Langevin licensed his patent portfolio to *Société de condensation et d'applications méchaniques* (SCAM), and ultrasonic echo-sounding became a commercial success. He shared some of the income associated with his patents with Marie Curie's daughters, Irene and Eve, and with Jacques Curie, whose PhD had originally described the quartz piézo-électrique [84].

The new challenges were comparable to those facing the later designers of medical ultrasonic equipment. There was interference between the transmit pulse and echoes from close targets. Using a bistable preamplifier to switch from transmission to reception, Langevin reported that he could distinguish an echo from a target as close as 1 m [26]. A means of continuous recording was needed. Langevin's collaboration with P.A.D. Marti of the French Hydrographic service gave rise to a continuous depth recorder, plotting the change in echo with time as the boat moved. The '*Sondeur Ultra-sonore Langevin-Florrison-Marti'* was widely installed on merchant and passenger ships by SCAM. In Britain, Henry Hughes and Sons produced their first recording echo-sounder for the Admiralty in 1923, with a recorder designed by their chief engineer Donald Sproule [85]. Characterization of the sea-bottom from the nature of the echo, whether sand, mud or clay, was also investigated at this time, a technique revisited in the 1970s for tissue characterization [86].

Naval applications generated interest in transducers other than quartz. There were two early piezoelectric alternatives. 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 [87]. The second alternative was X-cut Rochelle salt in which the piezoelectric effect is considerable greater than either quartz or tourmaline [88]. Rochelle salt was used as the receiver for the prototype US echolocation system before the end of the war, but it was a decade before a reliable method of manufacture was established. By 1940, Rochelle salt had become the preferred material used by the US Navy for its echo-ranging and submarine detection equipment. Moreover, off-resonance response meant that it could operate over a range of frequencies [89]. Later in the war, experimental assemblies of ammonium dihydrogen phosphate (ADP) crystals were tested, some capable of operating up to 100 kHz. By contrast, the British navy retained the pre-war resonant quartz transducers, largely operating at frequencies below 20 kHz.

A more robust alternative arose from the development of magnetostrictive transducers [90]. Magnetostrictive generation of sound results from periodic extension and contraction of a rod or tube of ferromagnetic material, typically a nickel-iron alloy, when placed in an oscillating magnetic field parallel to its length. The main advantages over quartz were cheapness and sturdiness. However, even at the lower frequencies where they were mostly used, magnetostrictive receiver sensitivity was about 40 dB below quartz, and, in transmission, they had to be driven at higher electrical powers to achieve the same acoustic output. In addition, eddy currents restricted the practical upper frequency limit to about 300 kHz. Whilst there were early ultrasound therapy units that used magnetostrictive transducers, such as the 175 kHz Atlas-Werke 'Supersonic', they were unusual [91]. No medical diagnostic systems used magnetostrictive transducers.

In all operational sonar and asdics systems, scanning was achieved by mounting the transducer on a rotating arm, and repetitive scanning carried out every 5˚, sweeping 90˚ each side of the direction of travel. Such a procedure required about 2.5 minutes for a complete sweep. By the end of the war, experimental switched scanning sonars had been developed. The Harvard Underwater Sound Laboratory system used a cylindrical array, flooding the whole 360° field during a 30 ms transmission (Figure 18). On reception, a small number of elements were selected, and rapidly switched in sequence to achieve a scan repetition rate of 30 Hz, 'introducing suitable phase displacement for making these elements into a directional hydrophone' [92]. The echoes were presented on a cathode ray screen.

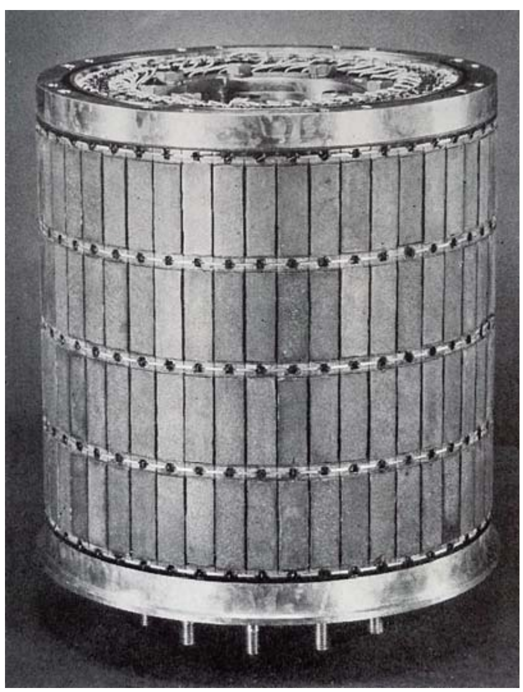

Fig 18. US Navy Scanning Sonar XQHA transducer with rubber cover removed. 1945. [92]

The frequency shift in the echo, caused by relative movement between the ship and the target, the Doppler shift, was also used by Sonar operators, who were trained to listen for any change in pitch of the 'pings' and interpret this in terms of speed and direction of the target. Doppler-shift would become an integral part of medical ultrasound. Thus, by the end of the war, underwater pulse-echo techniques were not only well-established in practice but also the transducer, driver and detection electronics and display systems underpinning this use were well advanced. This included array beam forming on reception, and echo display using cathode ray oscilloscopes.

VII. THE FIRST FOUR BOOKS ON ULTRASOUND

The end of the first wave of expansion in ultrasound was marked by the publication of four specialist books on this new part of acoustics within the space of four years, from 1937 to 1941. Two were in Germany one in France and the fourth in the USA.

Ludwig Bergman's *Ultraschall* was published by VDI-Verlag in Berlin in July 1937 [90]. Bergmann (1898-1959) had obtained a PhD from Geissen University in 1921 and then worked for Telefunken and at the University of Marburg. He joined Clemens Schaefer at the University of Warsaw in 1927, with whom he compiled the standard physics textbook, *Bergmann-Schaefer Lehrbuch der Experimentalphysik.* His first publications in ultrasound were in 1932, working on the piezoelectric properties of quartz and opto-acoustic effects. In 1939, he was appointed director of the physics department of the Technical University, and honorary professor at the University of Warsaw. After the war, he worked for Leitz and returned to Hesse as honorary professor at Giessen.

Bergmann's book emphasized the practical side of ultrasound, with many details of experimental studies and industrial applications. Indeed, the author gives credit to several German manufacturers for assistance with illustrations and other details. Following its publication, there was immediate interest in the USA and, the following year, Bergmann's book appeared in English translation by H Stafford Hatfield, published by John Wiley, with an additional section on ultrasonics in television. A second edition was published in 1946.

The second German book was *Grundlagen und Ergebnisse der Ultraschallforschung* (Basics and results of ultrasound research) by Egon Hiedemann (1900-1969) [58]. It included a greater emphasis on the fundamental physics and laboratory experiments, and included a remarkable 1346 cited references. Hiedemann had gained his PhD from Göttingen, before returning to Cologne where he was elected associate professor in 1931. Like Bergmann, he applied optical techniques to investigate ultrasonic fields, and was one of the earliest investigators into non-linear propagation effects. Hiedemann moved to the USA after the war, eventually becoming head of physics at Michigan State University.

Robert Wood's 150-page book '*Supersonics, the science of inaudible sounds',* was based on three lectures given in the Colver Lecture programme at Brown University in 1937 [42]. It covered broadly the same material as Bergmann but in perhaps less detail, with greater emphasis on the work from the Loomis laboratory.

The last of the early books on ultrasound was *Les ultrasons*, by Pierre Biquard [93]. It was very different from the others both in scope and intended audience. First published on 11 August 1941 in occupied Paris, it was volume 21 in the series of short paperbacks called *'Que sais-je'*. The series aim was to present an accessible introduction to a subject for a lay reader, written by an expert, within a small format 128-page book. The opening sentence, "14 avril 1912. – A bord du grand paquebot de la "White Star Line" tout le monde est heureux', (On board the great liner of the "White Star Line" everyone is happy), introducing the sinking of the Titanic as the trigger for the development of ultrasound, sets the tone of this well-illustrated, concise scientific and technical description of ultrasound. It undoubtedly reached a far wider audience than the other three books, the 9th edition still in print in 1983.

The circumstances of these two men, Biquard and Hiedemann, in 1941 could not have been in greater contrast. The German had just been appointed as professor and head of applied physics at the new Reichsuniversität at Strasbourg, an appointment that was made even though he was not a party member, a common requirement for new academic appointments at that time.

Meanwhile, Biquard was a fugitive in Lyons. A confirmed pacifist, Biquard had been secretary of the Paris section of the *Comité de vigilance des intellectuels antifascistes,* formed under Langevin's lead in 1935. By this stage in his life, Langevin had given priority to politics above science. As the left-wing British scientist JD Bernal wrote in his preface to Biquard's biography of Langevin, 'I have always remembered his dictum …. that if we did not do our scientific work someone else would, but if we neglected the political work, there would soon be no science' [94].

Biquard took part in the agitation caused by the arrest and imprisonment of Langevin after the capitulation of France. He was dismissed from his job as a scientist in the French Navy in December 1940, under the racial law. Under threat of arrest, Biquard went underground in Lyons where he spent rest of the occupation, participating in the French resistance movement. Which is why, when the wartime editions of his book were published, it was under the name of Francis Draveil, one of several cover names that Biquard used during the war.

VIII. ULTRASONIC TESTING BY THROUGH-TRANSMISSION

During the pre-war growth of ultrasonics, interest developed in its use for testing materials. Whilst there were occasional proposals that reflected waves could be used, these innovations largely centred on methods using transmitted ultrasound.

When Mühlhäuser was granted his 1931 German patent for ultrasonic methods for testing materials he included proposals for testing by transmission or by reflection [95] (Figure 19). In both cases, two transducers were used, and inhomogeneity within a sample was inferred by reduced transmission between them. The Russian physicist Sergei Sokolov was working at the Lenin Electrotechnical Institute in Leningrad in 1928 when he proposed his through-transmission technique for flaw detection in metals. At first he measured the transmitted intensity using light diffraction. By 1935 he had proposed the first image projection method using optical investigation of the surface displacement of oil [96]. A prototype was manufactured by VEB Jenoptik, Jena, for testing large sheet panels. By 1937 he had showed that the charge distribution induced on a quartz receiver could be scanned using the electron beam in an evacuated cathode ray tube, the so-called Sokolov Tube. Instruments working at 4 MHz based on this principle were later constructed by Jenoptik and proposed for medical applications [97].

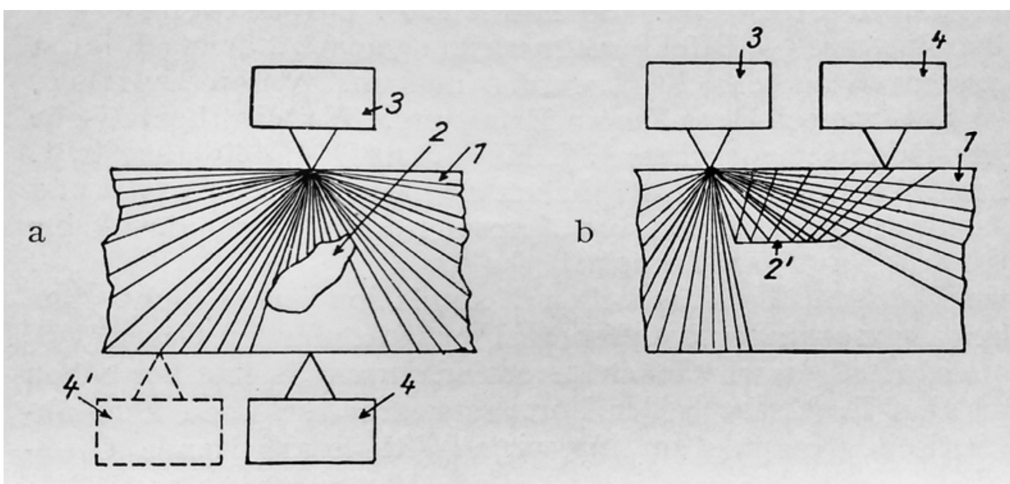

Fig 19. Ultrasonic materials testing by transmission (a) and reflection (b). 3: transmitter: 4,4`: receiver. Mühlhäuser 1931. [94]

A. Medical applications of transmission techniques

Drawing on these industrial developments, there were a number of attempts to exploit through-transmission of ultrasound for medical diagnosis. But, as Dognon said in his 1953 monograph on ultrasound ‘Obviously, it is urgent to apply methods already rich in results in the industrial field to the study of the human body. Unfortunately, the situation is much less favourable, because the body, the organs, and even most of the tissues are roughly heterogeneous in terms of their acoustic properties. The air, often present, constitutes, in the form of bubbles or thin sheets, interfaces of total reflection. Other more or less marked reflections occur at each zone of contact between tissues of different density or compressibility’ [98].

André Dénier’s simple homemade ‘ultra-sonoscope’, announced in 1946, was one example, a device that recorded the insertion loss through a tissue sample or organ [99]. In one application, the transmitter was placed on the trochanter and a vibration was picked up at the heel, perhaps as a means to diagnose fracture [100]. Another novel study by Wolf-Dieter Keidel, working in the Medical Physics laboratory at the University of Erlangen, used 60 kHz trans-thoracic transmission to investigate the heart [101]. The transmitted amplitude tracked the cyclic change in heart volume generating an ‘ultrasonocardiogram’. (Figure 20). But, as Dognon pointed out ‘truth be told, we have simpler and more precise methods for this’.

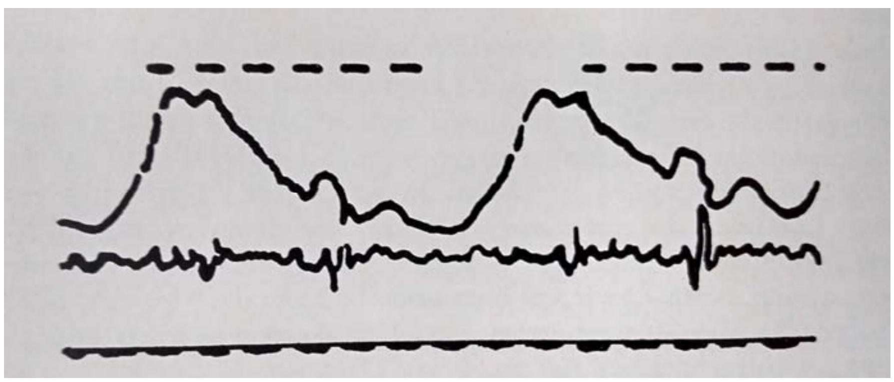

Fig 20. Keidel’s ultrasound cardiogram by transmission (upper trace). The lower trace shows the heart sounds. (1950). [101]

The most well-known failure from this period of transmission studies was due to the Austrian physician Karl Dussik, amplified by the Massachusetts Institute of Technology (MIT). Dussik’s tentative first paper on ultrasound was published in 1942 [102]. His first paper concerned transmission imaging of the heart [103] and the second on the brain [104]. The skull was immersed in a box containing water, and opposing quartz transducers were moved in a raster to scan the whole organ. The intensity of the reception modulated the brightness of a lamp and recorded photographically. Dussik’s work came to the attention of the American physicist Richard Bolt, who gained funding to construct a similar instrument in Bolt’s newly opened Ultrasonics Laboratory at MIT using improved electronics and replacing quartz transducers with barium titanate. Theodor Hueter was recruited from Siemens. Before leaving Erlangen in 1950, Hueter had the opportunity to try out pulse-echo methods with animal tissues but found the resulting echo signals difficult to interpret. He was easily persuaded to take part in a transmission imaging project instead, accepting the parallel with the x-radiography.

In reality, the images were swamped by diffraction artefacts, and much identified as pathology was noise. Refraction and diffraction limit the information in the transmitted beam, and attenuation limits the frequency that can be used and hence the resolution that can be achieved. Critical independent comment soon followed [105]. While the project was high in visual impact because images were produced it was never destined to open significant new avenues for clinical diagnosis. The episode had one positive outcome, however. In 1955, Hueter and Bolt published ‘*Sonics. Techniques for the use of sound and ultrasound in engineering and science’* in which their knowledge and skills from both sides of the Atlantic combined into a single seminal text-book [106]. Dussik’s heroic failure did not prevent him from being identified by some as ’The Father of Ultrasonic Diagnosis’ [107].

IX. Ferro-electric ceramic transducers

During the first thirty years, X-cut quartz, X-cut Rochelle salt and ADP had been the only piezoelectric sources of ultrasound. The invention of ferromagnetic ceramic piezoelectric materials created a revolution in ultrasound transducer technology. In 1947 this innovation was announced by three separate groups, notably by Warren Mason at Bell Telephone Laboratories [108,109]. All crystals have a second-order electrostrictive effect in which distortion occurs which is proportional to the square of the electric displacement. This ferroelectric effect is large in Rochelle salt, in potassium dihydrogen phosphate and in barium titanate, over a particular temperature range. The Curie temperature for barium titanate is 120 °C, above which ferroelectric properties are lost. In order to exploit this ferro-electric effect in a polycrystalline ceramic it is necessary to align the domains using a DC polarising voltage at a temperature above its Curie temperature [110]. It was soon also discovered that the introduction of small amounts of lead titanate into the ceramic caused a considerable increase in the thickness expansion, exceeding that found in magnetostrictive materials. It also prevented the decay of the ferroelectricity after the poling voltage had been removed. Further work on alternative ceramic mixtures led eventually to the dominance of the various forms of lead zirconate titanate ferroelectric ceramics that were used exclusively in the first development of medical ultrasonics. As Mason observed, a definite advantage of the ceramic transducer is that it can be moulded, leading to the possibility of producing focusing radiators of any size or shape. Unlike the situation with shaped transducers cut from pieces of crystal, this type of transducer can be made equally efficient for all points on the surface. Some of the weakly focussed ceramic transducers that were central to the achievement of good lateral resolution in medical imaging were shaped in this way, whilst focussed bowls have been essential to the development of ultrasound surgery.

X. Ultrasonic echo testing

On 27 May 1940, Floyd Firestone (1898-1986) filed for his first USA ultrasound patent ‘Flaw detecting device and measuring instrument’ [111]. From the Departments of Physics and Engineering Research at the University of Michigan, he stated that his invention ‘pertains to a device for detecting the presence of inhomogeneities of density or elasticity in materials’ and that it ‘may also be used for the measurement of dimensions of objects, and is particularly useful where one of the faces to which the measurement extends is inaccessible.‘

Firestone was well established as an acoustician, and had been editor-in-chief of the Journal of the Acoustic Society of America since 1939. He would become president of the ASA in 1943. Nevertheless, his previous publications had been limited to the audible spectrum, concentrating particularly on electro-mechanical analogies [112]. It was rare for papers in JASA, or at the ASA meetings, to include anything on ultrasound. The paper given by Hanfried Ludloff from Cornell University at the 23rd meeting of the ASA, 29-30 April 1940 in Washington, ‘Ultrasound and Elasticity’, may well have sparked Firestone’s interest. Ludloff was a recent German émigré who had worked on acoustic surface waves in Bergmann’s ultrasound laboratory in Warsaw before the war [113]. His paper at the 1940 meeting was unusual, not only because it was about ultrasound, but also because it concerned its transmission though solids in order to test their material properties [114]. His method, using optical interference caused by ultrasonic wave gratings to uniquely determine the elasticity and symmetry of an object, could not be applied easily to real-world testing. But it could well have been the trigger that led Firestone to invent a more practical ultrasound testing device.

Firestone was conscious that if any new testing regime was to be universally applicable, it needed to detect flaws very close to the surface, as well as those at depth. This led him initially to reject the pulse-echo method used for underwater detection. Instead, his method was a pulsed version of the method proposed by Mühlhäuser. It used two transducers, which are shown, in the patent, one on the end of the sample under test and the other on its side. He proposed using ‘perhaps 1,000,000 cycles per second’ although he identifies a lower bound of 15 kHz and notes that thinner crystals might generate ‘10 or even 50 megacycles’. The pulse repetition frequency would be about 100 Hz and the pulse length one cycle or less, a specification at odds with the resonant crystal transducers then available. The receiver amplifier could be tuned or broadband ‘from 1000 to 10,000,000 cycles per second’. Two transducers, piezoelectric or magnetostrictive, would be used, one for transmission and the other for reception, allowing them to be placed at any relative positions. The patent does not address the difficult question of coupling, a concern addressed by earlier workers on through-transmission testing.

When the patent was granted on April 21 1942, the USA was at war. It was the first of five patents filed by Firestone, as his understanding of the practical challenges of ultrasonic testing developed. The second, filed two years after the first on 29 June 1942, was the first to specify a single transducer acting as both transmitter and receiver [115] (Figure 21). Firestone had by then negotiated with industry, by then being 'assignor to United Aircraft Corporation East Hartford Conn'. This patent includes a control on the number of cycles in the pulse, and a receiver amplifier that 'shall within approximately a millionth of a second recover its sensitivity so that it may be sensitive to the feeble voltage trains generated by the crystal when feeble trains are reflected back to it from flaws distant only a fraction of an inch from the sending point'. To approach his aspiration to reduce the pulse to 'as short as half a cycle', he damped the quartz resonance with a backing block of Bakelite. In due course he patented a stand-off to resolve the problem of near-field detection [116].

Development of the 'Mark 1 Supersonic Reflectocope' continued in Michigan during the war, the construction and testing carried out by Julian Frederick. By 1945, Firestone had transferred his industrial partnership to Sperry Products Inc of Hoboken, New Jersey. Commercial development of the Sperry Reflectoscope Mark 2 used a single 5 MHz quartz transducer in pulse-echo mode, the transducer driven by 1 μs pulses, and a repetition frequency of 60 Hz, the values suggested in the second patent. Transducer resonance caused the acoustic pulse to lengthen to about 5 μs. Oil was used for coupling [117,118]. The first academic paper includes 'reflectograms', oscilloscope traces both as RF and demodulated (A mode) pulse, typically including five sequential sweeps on one screen. Results showing the evaluation of laminations, welds, grain size and wall thickness were given.

The electronic design of the Mark II Reflectoscope was primarily due to the engineer Benson Carlin (1915-1996). Carlin had taught radio engineering for the Signal Corps at Monmouth until July 1943 and, after a brief period in the Radiation Laboratory at MIT, joined Sperry Products Corporation as a research and product engineer. Later, he played an important role in the introduction of medical ultrasound, recalling that Firestone once pointed the transducer onto his leg and remarked, 'See, you might be able to find things in the body' [119].

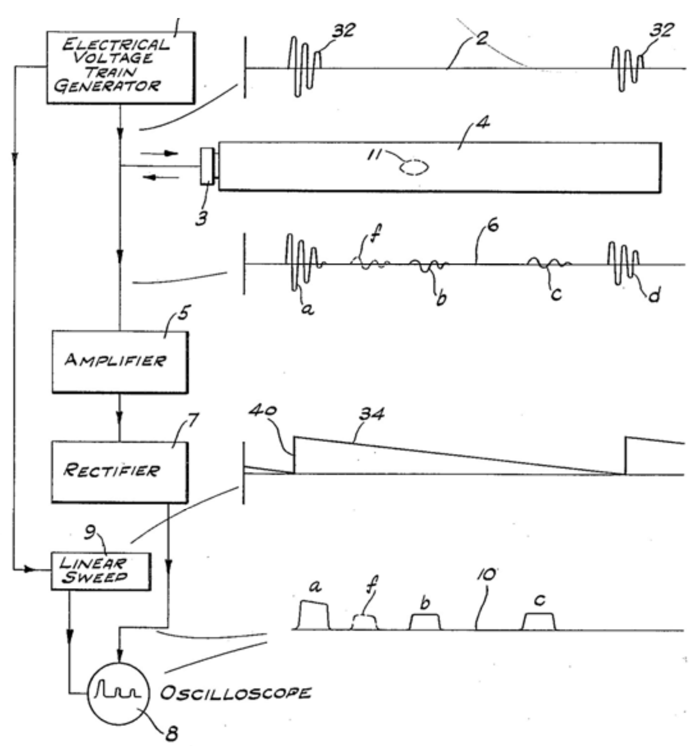

Fig 21. Block diagram from Firestone's pulse-echo patent. 1942

It is not surprising to find other initiatives at this time. In the United Kingdom, a Hairline Cracks Sub-Committee was set up, reporting to the Alloy Steels Research Committee of the Iron and Steel Research Council. The chairman was the eminent metallurgist Cecil H Desch FRS, recently retired from academic life. In November 1939 it was decided to investigate the possibility of employing "supersonic waves" for industrial purposes. Desch knew of the pre-war development of asdic systems for the Royal Navy and invited the company responsible for this development, Henry Hughes, to assist in his investigations. As a result, much of the experimental work for the sub-committee was carried out in the Hughes Research Division, under the direction of Donald Sproule, now returned from Canada. The initial experiments used well-established transmission techniques, but were found to be very limited in sensitivity. It was only in 1942 that Sproule began to investigate pulse-echo techniques, using, as he put it, 'echo-sounder principles to sounding in a solid sea'. By July 1943 Sproule and his colleagues were able to demonstrate a working prototype, using separate transmit and receive transducers with overlapping beams. (Figure 22) They used two heavily damped quartz transducers about 1 mm thick and 2 cm diameter, angled downwards into the sample to be tested. Heavy damping allowed the transducer, with a nominal resonant frequency about 2.5 MHz, to be operated at subharmonic frequencies down to 625 kHz. The 300 V pulse was generated by the rapid discharge of a capacitor through a thyratron. Once the war ended, details were released and Hughes produced the Mk I ultrasonic metal flaw detector [120]. The use of two transducers allowed flaws to be detected as close as 1 cm from the surface.

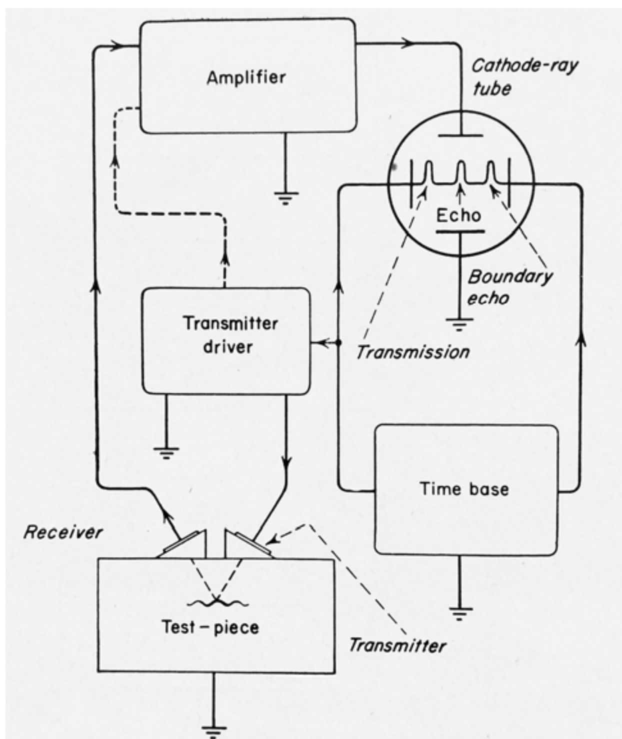

Fig 22. Block diagram from the Hughes pulse-echo system. 1943.

Carlin was well placed to record the technical details of developments at this time. In his book 'Ultrasonics", published in 1949, he identified the enabling technologies that underpinned these new developments in ultrasound [121]. Carlin recognised the similarity between short-range materials testing using ultrasonic pulses and the techniques used in sonar and in radar, noting that both work best if the transmitter and receiver are identical. This challenges the ability to detect objects close to the transmitter. For radar, pulse lengths were of the order of 1 μs, in contrast to pulse lengths up to 100 ms for sonar. Pulse circuits, capable of electronically shaping very short MHz pulses, had been developed for radar. RF pulses had been generated in a number of ways, using gas valves or thyratron switches, or switched circuits with high-vacuum valves. Cathode-ray oscilloscopes had been available before the war, but it was not until 1946 that the triggered-sweep oscilloscope, developed for radar, became commercially available from Tektronix. Mapping of echo location had been previously developed for both sonar and radar although at this stage materials testing with ultrasound was restricted to A-mode. Marker circuits enabled time reference points to be displayed, and hence a distance calibration to be set up. Broadband receivers were required to achieve good resolution. Carlin was very clear about the debt paid to radar technologies in the early development of ultrasonic testing.

Other engineers with radar and sonar experience were encouraged by their companies to explore new applications, including ultrasound applied to medicine. RP McLoughlin was working as an engineer in the Argentinian subsidiary of RCA Victor when he started to explore the possibility of foreign body location in the body using pulse-echo methods, together with the physician GN Guastivino, They published their prototype results at 1.5 MHz using the acronym LUPAM (Localizador ultrasonoscopico para aplicaciones medicas) in 1949. The paper included an A-mode oscilloscope trace showing reflections from a bone and from a stone embedded in an excised kidney. Notably, it also suggests that echo position could use the PPI (plan position indicator) presentation developed for radar, with a rough sketch showing how a medical image might appear. [122]

In France, Florisson continued his interest in medical applications, contributing to the first issue of *UIltraschall in der medizin* in 1949 [123]. Josef Krautkrämer in Cologne and Karl Deutsch in Wuppertal both started investigating pulse-echo materials testing in 1949, offering commercial pulsed test equipment a year later. Siemens soon followed with the Ultraschall-Impulsgerät, released in 1952. It was a development of this device that was borrowed, in 1953, by Gustav Hertz' son Hellmuth, working with Inge Edler in Malmö, to initiate ultrasound echo technology for cardiology.

But, when Pohlman reviewed the status of ultrasound in 1950, he could report only that 'The echo method is a very promising approach for diagnostics, but it is also not yet fully developed. So far nothing has become known about practical applications'. Practical diagnostic ultrasound was still in the future.

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AUTHOR BIOGRAPHY

Francis Duck BSc PhD DSc MBE is from Bath UK. He is a retired medical physicist who spent most of his career developing applications of ultrasound to medicine. His earliest association with ultrasound was using a Diasonograph in the basement of University College Hospital London, to scan his wife Di for their first son Roger, in November 1967. Early work, first with intravascular Doppler systems and then experimental ultrasonic imaging, took him to Newfoundland and then the Mayo Clinic, MN, USA. Returning to the UK, he settled in Bath, forming a link between the Royal United Hospital and the physics ultrasonics group at Bath University. Much of the focus of his later work was on ultrasonic metrology and ultrasound safety. Since retirement he published widely on the history of medical physics. bathduckf@gmail.com

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].

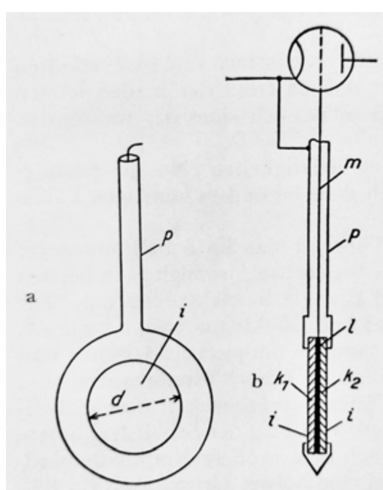

Fig. 1 Langevin's quartz crystal probe (1926). a) Front view. b) cross section. k_1 k_2 quartz plates; l metal plate; p metallic housing; m insulated cable; i electrodes. (From Langevin 1926) [1]

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.

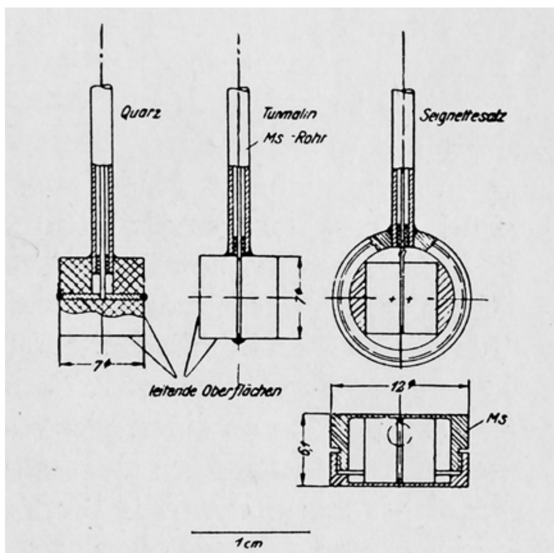

Fig. 2 Three alternative hydrophones, using quartz, tourmaline and Rochelle (Seignette) Salt. (From Meyer and Tamm 1939) [5]

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].

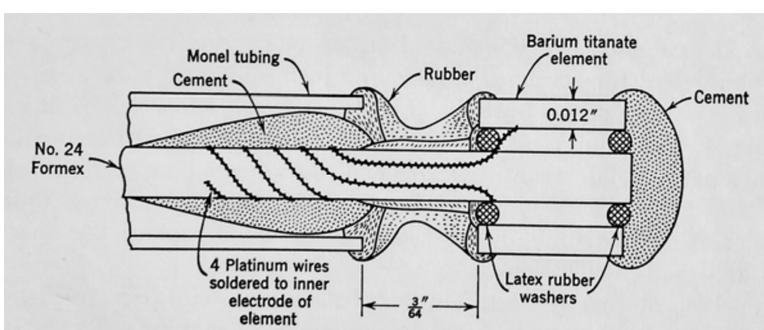

Fig. 3 A miniature piezoelectric ceramic hydrophone based on a small barium titanate cylinder element. (From Hueter and Bolt 1955) [10]

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.

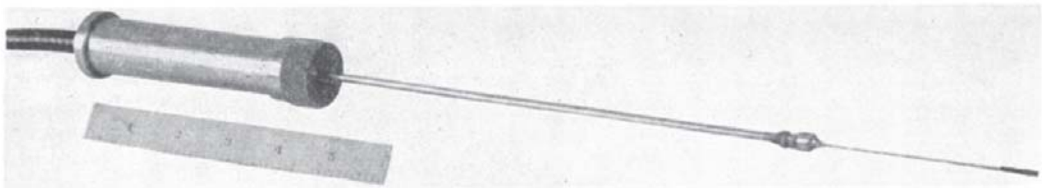

Fig. 4 The probe hydrophone of Mellen (1956) [11]. (Reprinted with permission from Mellen RH. An experimental study of the collapse of a spherical cavity in water. *J Acoust Soc Am* 1956; 28: 447. Copyright 1956, Acoustic Society of America.)

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.

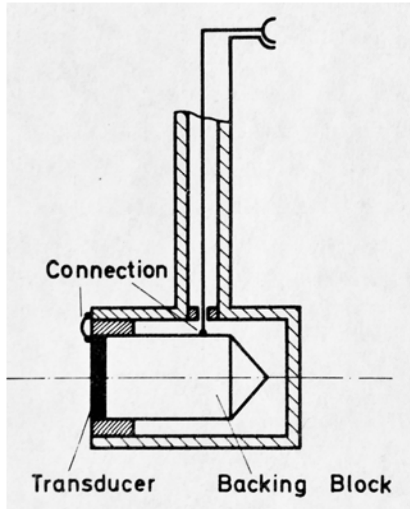

Fig. 5 A general hydrophone design for pulsed diagnostic beams. (From Brendel 1972). [15]

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.

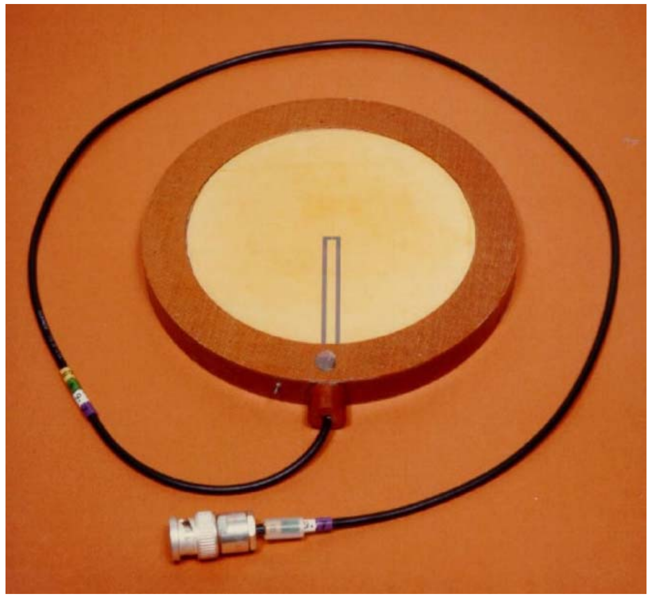

Fig. 6 An early NPL/Marconi bilaminar membrane hydrophone. (Image courtesy of Dr B Zeqiri, NPL, Teddington, UK)

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].

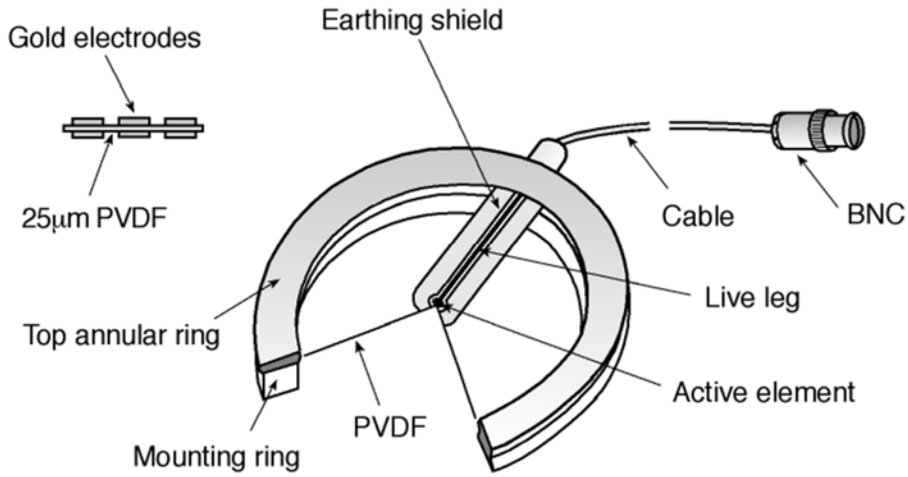

Fig. 7 The electrode configuration of the coplanar shielded membrane hydrophone. (Image courtesy of Dr B Zeqiri, NPL, Teddington, UK)

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.

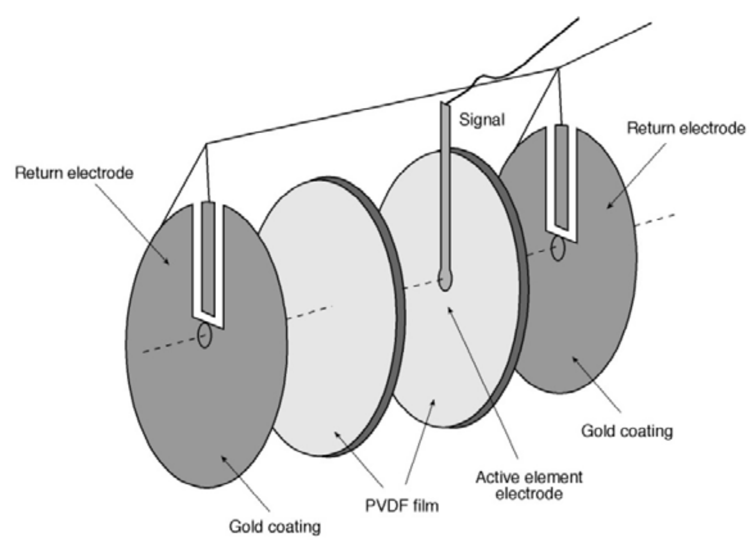

Fig. 8 Construction of the bilaminar shielded membrane hydrophone. (Image courtesy of Dr B Zeqiri, NPL, Teddington, UK)

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.

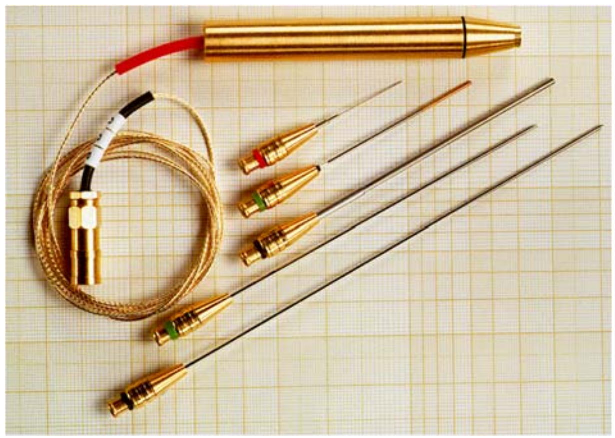

Fig. 9 A set of early Precision Acoustics needle hydrophones. The hydrophones could be connected interchangeably into the submersible preamplifier shown at the top of the picture. (Image courtesy of Dr A Hurrell, Precisions Acoustics, Dorchester, UK)

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.

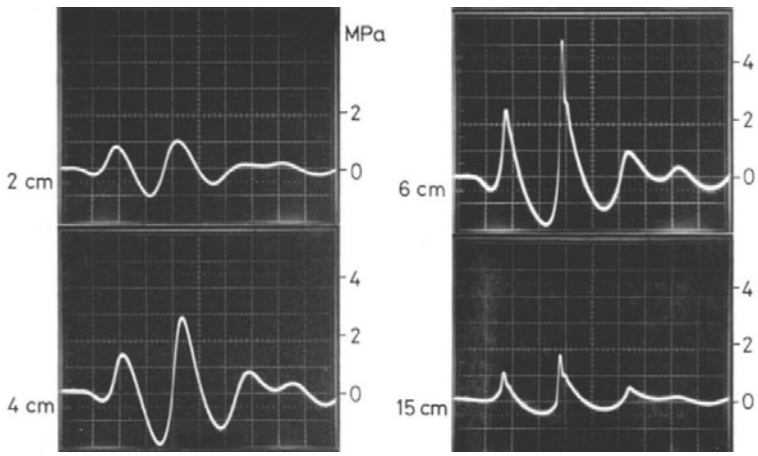

Fig. 11 The pressure pulse measured in the field from a focused 2 MHz transducer at 2 cm, 4 cm, 6 cm and 15 cm range. (From Duck FA, Starritt HC. Acoustic shock generation by ultrasonic imaging equipment. *Brit J Radiol* 1984; 57: 231-40.) [35]

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].

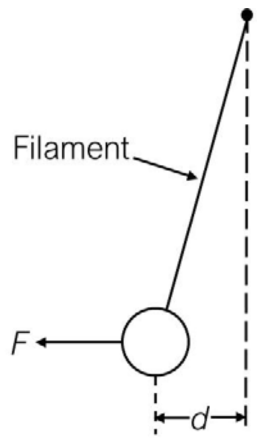

Fig. 12 The elastic sphere radiometer suspended in a sound field.

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].

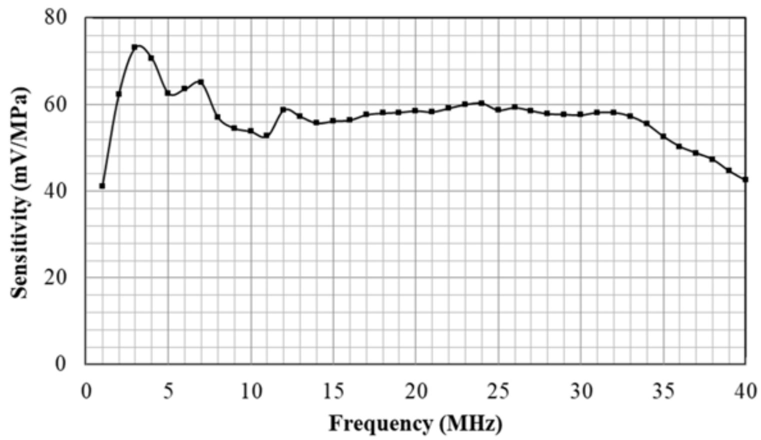

Fig. 13 Amplitude response of a 9 µm, 0.2 mm diameter PVDF needle hydrophone. (Courtesy of Dr A Hurrell, Precisions Acoustics, Dorchester, UK)

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.

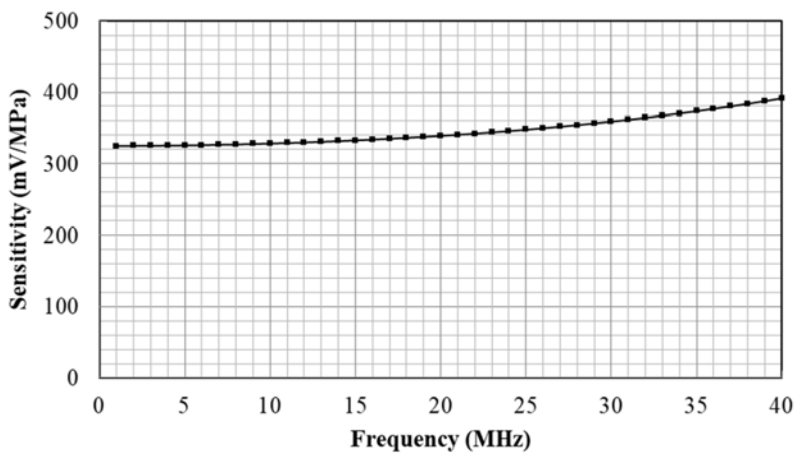

Fig. 14 Amplitude response of a 16 µm, 0.4 mm PVDF differential membrane hydrophone. (Courtesy of Dr A Hurrell, Precisions Acoustics, Dorchester, UK)

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.

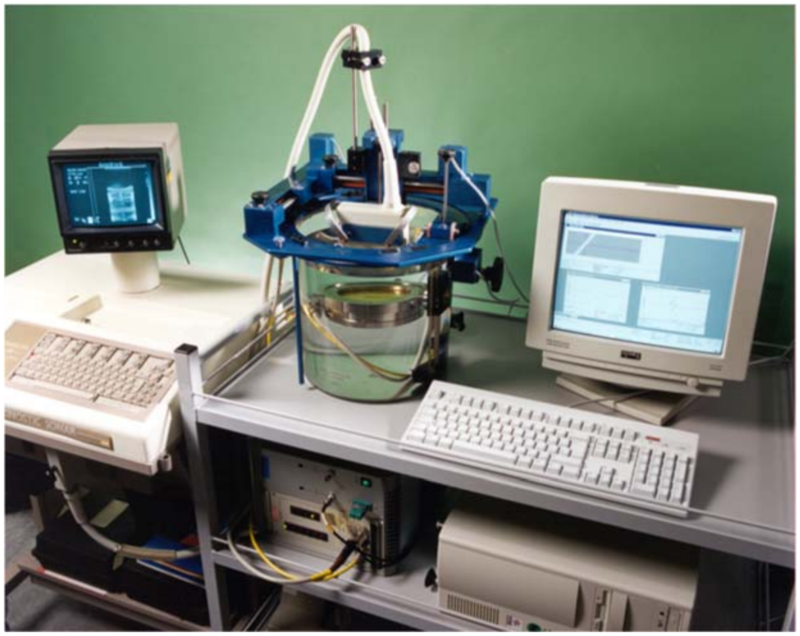

Fig. 15 The NPL Ultrasound Beam Calibrator. (Image courtesy of Dr B Zeqiri, NPL, Teddington, UK)

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].

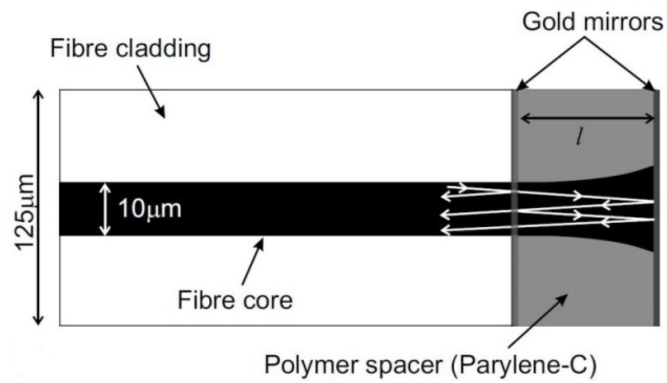

Fig. 17 The sensor head of a Fabry-Perot miniature optical fibre hydrophone. (Reprinted with permission from Morris P, Hurrell A, Shaw A, Zhang E, Beard P. A Fabry–Pérot fiber-optic ultrasonic hydrophone for the simultaneous measurement of temperature and acoustic pressure. *J Acoust Soc Am* 2009; 125: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

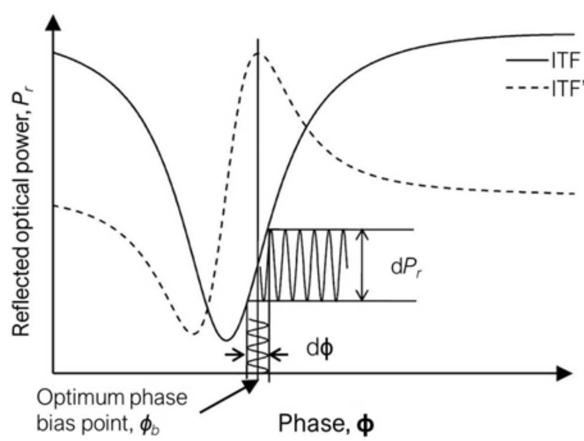

Fig. 18 The intensity-phase transfer functions (ITF) for a Fabry-Perot interferometer. (Reprinted with permission from Morris P, Hurrell A, Shaw A, Zhang E, Beard P. A Fabry–Pérot fiber-optic ultrasonic hydrophone for the simultaneous measurement of temperature and acoustic pressure. *J Acoust Soc Am*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

ULTRASONIC METROLOGY II – THE HISTORY OF THE MEASUREMENT OF ACOUSTIC POWER AND INTENSITY USING RADIATION FORCE

Francis Duck

Formerly, University of Bath and the Royal United Hospital Bath NHS Trust

I. Introduction

When a solid object is placed in a sound beam, it experiences a force perpendicular to the interface with the surrounding medium. This force is proportional to local energy density.

This phenomenon has been used in ultrasonic metrology in two ways. If a small object is placed within an acoustic beam, the force on that object can be used to measure the acoustic intensity. If the object is large enough to cover the whole beam, then the radiation force is proportional to the total acoustic power W. In this case the force F =kW/c, where c is the speed of sound in the medium and k is a constant depending on the nature and shape of the object, and the form of the acoustic wave. For an object with a plane surface perpendicular to the direction of propagation, k=2 if it is fully reflecting, and k=1 if it is fully absorbing.

It will be seen in this article how this very simple principle has been used in the design and use of ultrasound power balances of a wide variety of designs and sensitivities.

II. The torsion pendulum

The measurement of acoustic power using radiation force in an ultrasound beam is as old as the use of ultrasound for pulse-echo detection. Paul Langevin had started to investigate the use of ultrasound for submarine detection very early in the First World War. By July 1915 he reported an intensity of about 100 mW cm^{-2}, measured using a radiation force method.

'During the course of operations it was necessary to be guided constantly by the intensity of the ultra-sonic emissions produced, and for this purpose a convenient and valuable process based on the existence of *"pressure of radiation"* was used. …. When this radiation is directed onto a palette suspended by a torsion wire, the palette, if its thickness be appropriate, is pushed by the ultrasound and the torsion of the wire permits the measurement of the power emitted. ' [1]

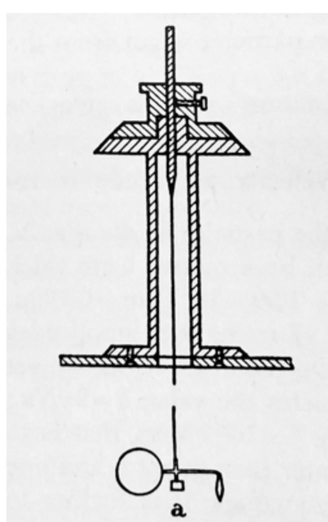

Fig 1. A torsion pendulum for the measurement of acoustic intensity.

Langevin himself gave no further details of his measurement method. However, in his 1941 book about ultrasound, Pierre Biquard, a student and subsequently close colleague of Langevin, confirmed that he had verified his first attempts at producing ultrasound with a torsion pendulum, describing the instrument as follows (Figure 1)

'A circular disc, the diameter of which must be large compared to the wavelength of the ultrasound, is fixed to a horizontal beam attached to a rigid rod T suspended from a fine wire f. (Quartz and phosphor bronze are particularly suitable, since they do not retain any residual twist.) A counterweight m balances the disc. A mirror M, attached to the rod T, allows optical observation of the rotation of the pendulum' [2].

The angle of rotation is proportional to the radiation pressure, and hence the intensity averaged over the area of the disc target. Counter-rotation of the twisted wire gives greater precision than the optical measurement of the angle of twist. The addition of a damping weight aided stability. Biquard added that the method had been since widely used for research in laboratory ultrasound studies, which included his own investigations into the loss of energy in finite-amplitude waves.

Langevin also gave values for the total power emitted by his transducers. It seems doubtful that his targets were large enough to encompass the whole beam, and so he must have estimated the power from the ratio of areas of the quartz transmitter to the target.

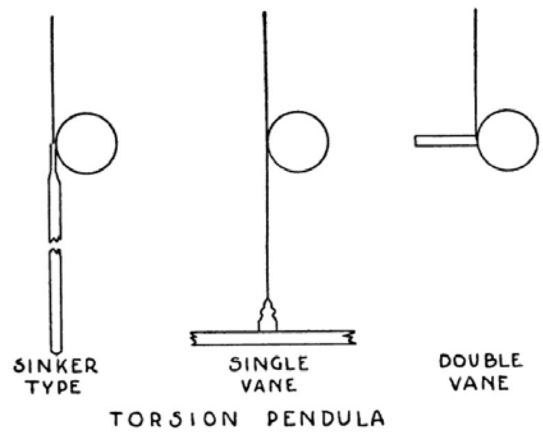

Fig 2. Various designs of torsion pendulum targets. [4]

When Langevin's wartime colleague, Robert Boyle, returned in 1919 from Britain to Alberta University in Canada, he established an ultrasonics laboratory in which he carried out many studies in ultrasonic acoustics. Amongst these, he used a torsion pendulum to investigate relative axial and radial intensity profiles of his ultrasound beams [3]. Figure 2 shows his various designs. By traversing the vane through the beam, Boyle was able to compare observed with predicted intensity profiles [4] (Figure 3).

Alternative targets to reflecting plates were investigated. The first power estimates in the USA were made using an absorbing target less than one wavelength across. In a report dated 1 April 1919, from the Throop College of Technology (later Caltech) Anderson et al reported that they had explored reflecting and absorbing targets, plates, cylinders and bulbs, their selected target being a small, absorbing cylinder made of blotting paper. They estimated the intensity by measuring the force it experienced close to the face of the transducer. Intensities up to 11.0 W cm^{-2} were reported [5].

Interest in the use of very small targets to measure local intensity continued, particularly in the USA, particularly once it was appreciated that the finite size of the disc target introduced diffraction errors for which corrections were required [6]. As an alternative, Elias Klein proposed measuring the radiation force on a suspended spherical target by observing its displacement [7]. Such direct methods were never widely adopted for intensity mapping, however, especially once high-fidelity calibrated hydrophones became available from which intensity could be calculated.

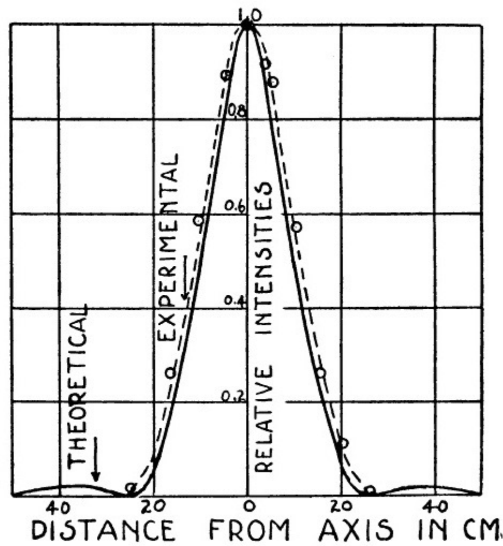

Fig 3. Measured and calculated intensity profiles using the torsion balance. Boyle 1928. [4]

III. Acoustic power in a standing wave

At the same time as some were mapping ultrasound beams using radiation force, others wanted to measure the total ultrasonic power used in laboratory experiments into the chemical, physical and biological effects of ultrasound. The physicist Robert Wood and the ex-banker Alfred Lee Loomis were notable among the early ultrasound investigators in the 1920's [8]. They soon discovered that it was challenging to measure the total emitted power by determining the force on a large plane reflector placed above a quartz plate. Their sketch (Figure 4) showed the problem. Reflected sound between the target and the quartz transducer set up standing waves in the exposed water bath. The energy density varied with position and the force on the reflector varied with the separation. Only with careful adjustment of the separation could a full standing wave be created, and only then could the force be related to the total acoustic power. This applied equally to Elias Klein's novel spring balance design in which the incident beam of ultrasound lifted a spring-loaded disc against gravitation forces, the spring being so designed that a very small contraction introduces a large axial rotation [9]. The standing-wave power balance principle continued to retain interest until well into the 1950s. For example, Walter Cady developed an interesting design at Caltech, which required a standing wave to be established between the plane face of the transducer and a liquid/air interface above it. In this case the transducer and not the target formed one arm of a balance, and its apparent change of weight was measured when the beam was switched on [10].

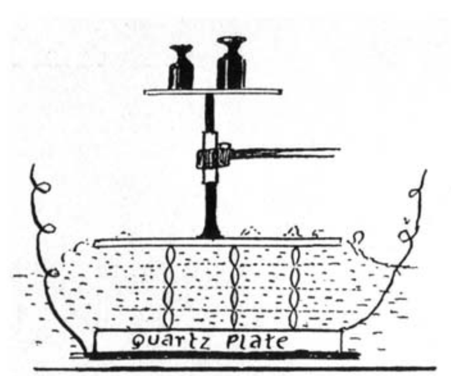

Fig 4. Sketch diagram showing the principle of weighing the variation of radiation force in a standing wave. Wood, 1939. [11]

IV. ACOUSTIC POWER IN A TRAVELLING WAVE: EARLY DEVELOPMENTS

The standing-wave approach was only appropriate for specific applications. The face of the transducer had to be plane. The acoustic load on the transducer depended on the experimental design. It could not be used for measuring insertion loss when investigating the frequency-dependent attenuation of liquids. Other designs emerged, arranged to inhibit or eliminate acoustic reflections and standing waves.

Sörensen's balance used a beam travelling in the vertical direction in oil, exerting force from beneath a target whose position was restored by the addition of weights, the use of oil inhibiting the creation of standing waves [12]. In 1937, in Paris, Ernest Baumgardt devised a balance using a conical target, absorbing the reflected waves in an acoustic lining (Figure 5) [13]. He used this balance to explore the linearity and resonant behaviour of quartz transducers. Ultrasonic laboratories in Japan [14] and in the USSR [15] similarly developed ultrasonic measurement devices based on radiation force.

It was not only laboratory scientists who needed to measure acoustic power. This pre-war period was also one during which underwater applications of sound continued, accelerating as war became increasingly inevitable. The large power balance shown in Figure 6, now in the PTB collection, was constructed at Atlas-Werke, Bremen during the 1930s. Its cone-shaped target is large enough to span the full beam width of a marine transducer, including perhaps one operating in the audio spectrum.

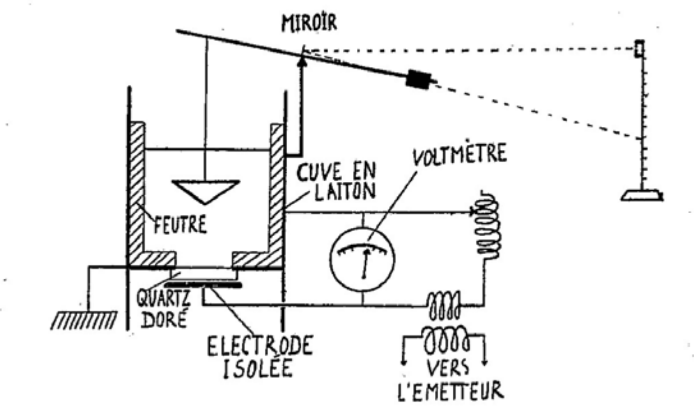

Figure 5. Ernest Baumgardt's balance with vertical beam and conical target. 1937. [13]

Fig 6. 1930s Radiation-force balance of Atlas-Werke, Bremen. Physikalische-Technischen Bundesanstalt, Brauschweig.

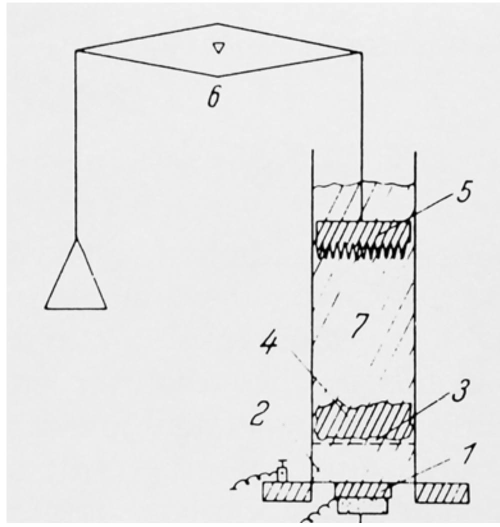

Fig 7. Pohlman's use of radiation force to measure ultrasound transmission loss in samples of tissue. 1 & 2 quartz fixed to tin foil: 3 & 4 tissue sample laid on a paper sheet: 5 scattering target: 6 arm of a sensitive balance: 7 sodium chloride solution. Pohlman, 1939. [17]

V. Commercial portable power meters

The rapid growth in the use of ultrasound for therapy in the late 1940s generated a significant interest in the measurement acoustic power. This resulted in commercial power balances becoming available by 1950 from two of the larger German companies selling ultrasound therapeutic equipment [16].

Reimar Pohlman(n) had joined Siemens in Berlin after completing his PhD in physics in 1932. As head of the ultrasound laboratory, he started to explore the therapeutic use of ultrasound, investigating the frequency-dependent attenuation of various body tissues, comparing adipose, muscle and mixed fat and muscle tissue samples [17]. Transmission loss was measured using radiation force (Figure 7).

Fig 8. The Siemens Sonotest c 1950. Physikalische-Technischen Bundesanstalt, Brauschweig.

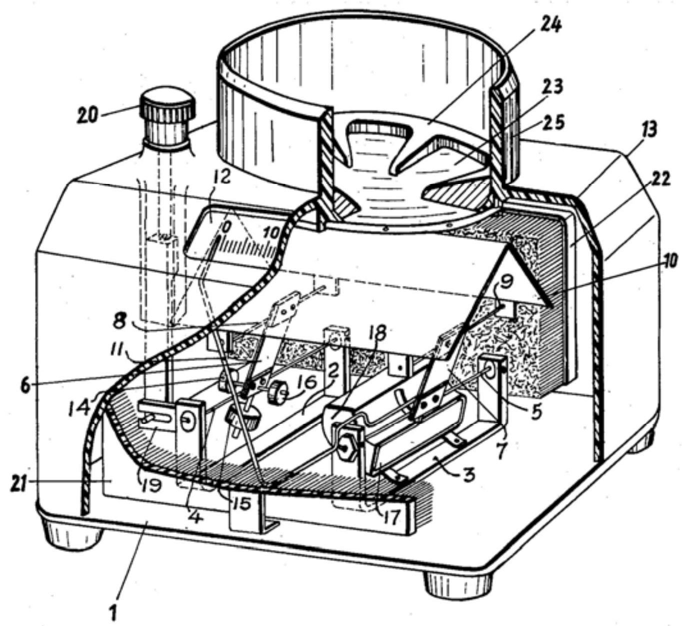

Fig 9. Cutaway plan of the Fiedler portable power balance as shown in the patent. Hueter & Bolt [18].

The beam was oriented in a vertical direction so that the force on the target could be ‘weighed’ directly. The target was large enough to encompass the whole beam so that the total acoustic power might be estimated. Standing wave formation was inhibited by deep indentations in the target creating a scattering surface. Pohlman had no need to measure absolute power: he was only concerned with the loss caused by the introduction of the tissue sample. Among the possible sources of error, Pohlman recognised that secondary phenomena such as streaming and acoustic cavitation could distort his results, and he made measurements with the quartz transducer driven at several voltages to check for any deviation from linearity.

Pohlman’s initiatives for the therapeutic use of ultrasound flowered with the renewal of European economies after the war. Under Pohlman’s guidance at the Seimens-Reiniger-Werke in Erlangen, Siemens launched their commercial therapy unit, the Sonostat, in 1947. It was designed to operate quartz transducers at two frequencies, 800 kHz and 2.4 MHz, with two transducers sizes, 10 cm^2 and 40 cm^2. The average intensity at the transducer was limited to 3 W cm^2, based on Pohlman’s pre-war safety assessments.

A simple means was required for calibration and maintenance. At Siemens, Georg Fiedler, working with Pohlman, developed and patented a rugged compact power balance with a roof-shaped reflecting target, which was sold by Siemens as the Sonotest (Figures 8,9) [18,19]. In Pohlman’s words:

‘The device consists of an open coupling chamber and a closed measuring chamber, which are separated from each other by a thin membrane. The coupling chamber accommodates the transducer to be measured either on a star-shaped guide grate or in simple retaining rings. To couple the transducer to the device, the coupling chamber is filled with degassed water. The measuring chamber contains the measuring system. It consists of a roof-shaped reflector that measures the radiation force, which is connected to a torsion spring parallelogram and a pointer. The pointer moves over a scale, calibrated in watts. Laterally attached absorbers prevent standing waves from occurring. A damping device shortens the settling times of the power meter. The measuring chamber is completely filled with distilled and degassed water. There is a rotary knob next to the coupling chamber, which allows the pointer to be readjusted to the zero point on the scale. The equilibrium of the system is so compensated that it is possible to measure in the main working position (irradiation from above) and in the horizontal position (instrument lying down). This means that horizontal sound fields can also be recorded by immersing the entire instrument in the water using a handle that can be attached for this purpose, as can be desired with permanently installed sound sources (marine transmitters) or water-bath treatments.

Fiedler had adapted a balance used by his colleague Theodor Hueter, who was investigating ultrasound transmission through solid rods in Erlangen at the time. In Hueter's design, the apex of the rooftop target was oriented towards a horizontal beam, and it was suspended from a vertical, counter-balanced arm [20].

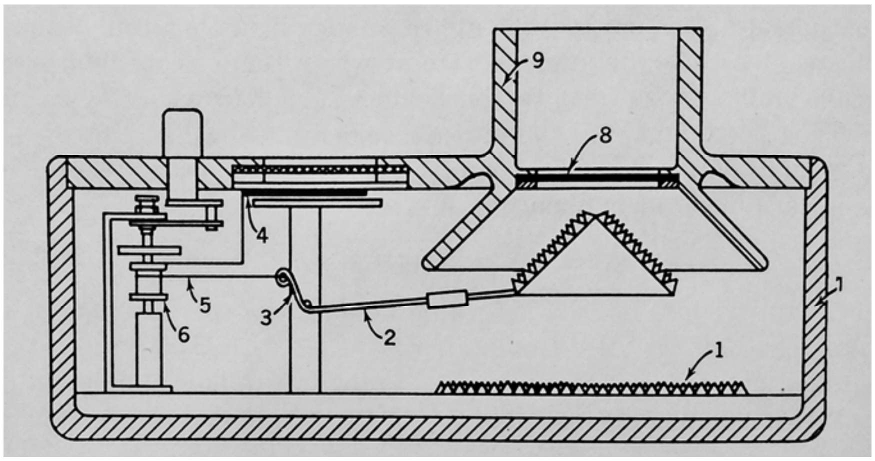

Fig 10. The Atlas-Werke portable power balance c 1950. Hueter & Bolt [21].

The chamber in the Sonotest was rendered anechoic using two panels of a brush-type lining material, one on each side of the target. The operating range was 0 to 50 W, and the maximum transducer outside diameter that could be accommodated was 80 mm. The angled target not only prevented standing waves, it was also less prone to error from poor alignment, and Pohlman claimed that misalignment of up to 8° could be tolerated. The radiation force acted against springs, rather than gravitational force, so the balance could be turned on its side so as to measure power with a horizontal beam in a water bath.

The reflector could be made from two metal plates separated by a sheet of dry paper, to approximate more closely to a complete reflector [21]. However, Fiedler's patent left open the possibility of a 'reflecting and/or absorbing target' and anticipated future designs that might use reflecting targets consisting of sheet metal enclosing an air space.

A different portable power balance was developed at the same time by Krupp Atlas-Werke in Bremen. It was designed to be used with their therapy unit, the 'Supersonic', a magnetostricitive device operating at 175 kHz. The balance is shown in Figure 10. The target was surfaced with small absorbing rubber cones, a material also used as the anechoic lining material. The target formed one arm of a balance; the other arm (3) actuated a pointer (4) by means of a string (5) wound on a drum (6) attached to the pointer axle.

VI. National laboratory calibrations for medical ultrasound

The growth in therapeutic ultrasound during the second half of the 1940s, particularly in Germany, created the need for a centralised ultrasound calibration and type-testing service [22]. This need was explored further by the specialist group for electro-medicine in the Central Association of the Electrotechnical Industry (Zentralverband der electrotechnischen Industrie, ZVEI). In 1948, an approach was made to Dr W Engbert, then working for Atlas-Werke, for advice. It was decided that type-testing should be carried out in a neutral laboratory rather than contracting a single manufacturer, so Engbert's patented design of power balance was taken up by the national standards laboratory, Physikalische-Technischen Bundesanstalt (PTB) in Braunschweig, under the lead of Paul Rieckmann. The requirements and test procedures that permitted manufacturers to label equipment as having achieved a standard of performance included one for acoustic power [23]. The selected design was for a free-floating target of neutral buoyancy (Figure 11). The beam was directed vertically downwards in a water bath onto a hollow-cone-shaped reflector. The shallow opening angle, 25°, ensured that reflected sound was deflected away from the transducer to be absorbed in the lining of the chamber. The floating target, or 'swimmer', was air-filled with a thin-walled cover surface, assumed to be a perfect reflector. Below the target there was a hollow stem with a scale, which was lowered into a volume of carbon tetrachloride giving the whole assembly neutral buoyancy. When exposed, the radiation force caused the target to sink by an amount proportional to the force, and hence the total power.

One novel aspect of the design was the concave hollow cone. This resulted in a self-centring property, any tendency for lateral displacement resulting in an asymmetric restoring force. Calibration of the displacement was achieved using small aluminium weights in the range 1 g to 7 g. For a reflecting target, the force in the direction of the beam depends on $\cos^2\alpha$, where α is the angle of incidence.

Several experimental details helped to improve the fidelity of the measurements. Cavitation was prevented by degassing the water under vacuum for 24 hours, which was then pumped through a closed system to the measurement chamber. The temperature was controlled at 30°C, so managing the neutral buoyancy and the sound velocity. An aluminium foil was placed obliquely over the target to protect it from streaming, still perceived as a 'quartz wind' generated by the transducer.

The balance was used for the certification of medical ultrasound therapy equipment by PTB, starting in about 1952. It would appear that Pohlman had a close informal relationship with PTB at this time because he mentions in his 1950 book that he had obtained a PTB performance certificate, which showed 'a display error of ± 5%' certificate. In their 1952 account, Oberst and Rieckmann reported that measurements using the PTB balance were broadly equivalent to those made using the Sonotest, and that both PTB and Pohlman had also calibrated their power balances at higher powers against a calorimetric method.

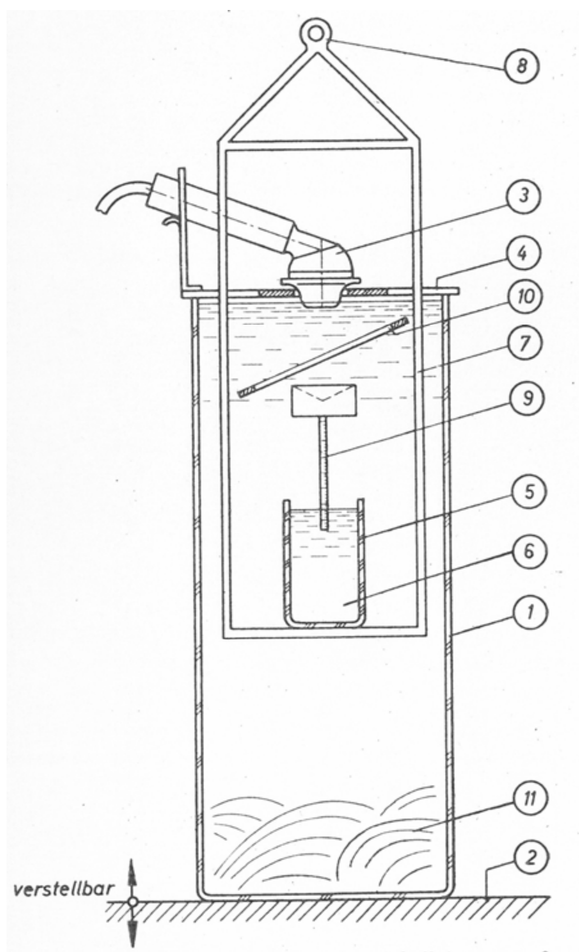

Fig 11. The PTB floating target power balance. 1952. The transducer 3, is centred over the streaming screen 10 above the buoyant target 9. The lower calibrated stem is lowered into carbon tetrachloride 6. Oberst & Rieckmann [22].

A. Development of the PTB buoyancy method

During the 1950s, others used the buoyancy technique in their own laboratories. George Henry was working for the General Electric Company in Schenectady, New York, when he published his review of power measurement in 1957. He selected radiation force as his preferred method for measuring 'gross acoustic power transfer' to a liquid load. He set out the criticisms of the two alternative methods, to use a hydrophone to plot of acoustic pressure from which power could be calculated from integrating the calculated intensity, and calorimetric methods calculating power from the rate at which sound energy is degraded into heat. In order to apply radiation force measurement for his applications, which included 'degassing, impregnation, emulsification and certain chemical processes', he used a neutral buoyancy, absorbing float target (Fig 12) [24].

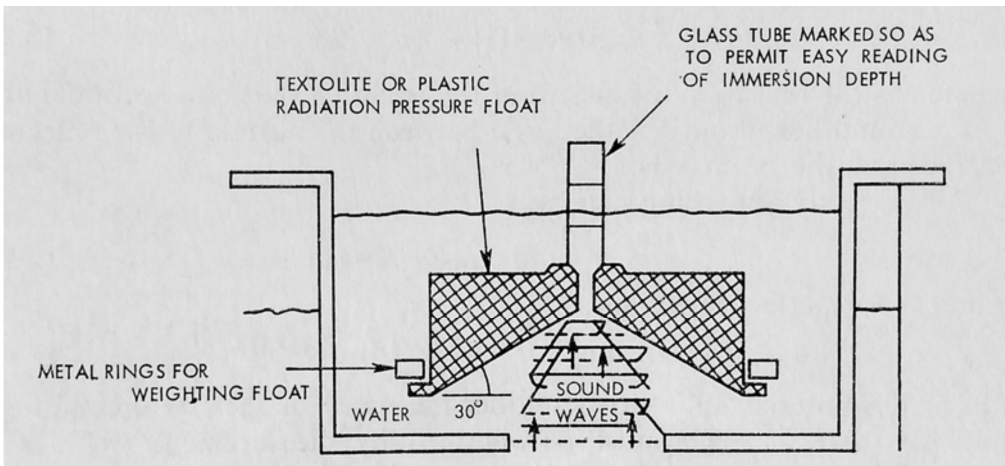

Fig 12. Henry's design for a power balance using a neutral buoyancy absorbing float target. 1957. [24] (©Iliffe Books Ltd)

The solid target was made from Textolite, a machinable laminated electro-insulation material made from resin-embedded fabric, probably chosen for its availability more than for its acoustic properties. The density is less that 1.0, so metal rings were added to give neutral buoyancy. A marked glass tube allowed changes in immersion depth to be read. The weighted floating target was freely suspended in water above the quartz transducer, and the displacement measured using a marked glass tube.

The absorbing target mitigated the some uncertainties resulting from reverberations but introduced new challenges of calibration. The Textolite was not a perfect absorber of sound, so Henry still had to use a re-entrant cone. There remained the difficulty of the calibration factor appropriate to a partially reflecting target. Henry proposed two methods for correction, neither entirely satisfactory. The first was to compare the force using two cones, one angled at 30° and a second at 15°. The second method involved evaluating the changed force as the target was progressively misaligned.

Mostly, others could not improve on Engbert's design as implemented at PTB. In 1962, George Kossoff, in Sydney Australia, described a variation of the float target in which a self-centring 30° cone was suspended under the transducer, with the beam oriented downwards. The target was made from a reflecting metal shell suspended in carbon tetrachloride [25]. The US National Bureau of Standards in Washington copied the PTB design for their own power balance [26]. The National Physical Laboratory (NPL) in the UK used the same principle in its later 'tethered float radiometer', with a downward vertical beam impinging on a floating, reflective target. In this case, however, the target was stabilised by attachment to three chains hanging partially below the float. As with other balances using a downward-directed beam, calibration was achieved using weights placed on the target. The stated range was 200 mW to 9W [27].

VII. MILLIWATT BALANCES FOR DIAGNOSTIC ULTRASOUND

A. Gavitational balances

The best sensitivity of the balances so far described was limited to about 100 mW. Newell's balance, in 1963, in which power was derived from the displacement of a fully immersed target, was still limited to a sensitivity of 60 mW [28]. This was entirely satisfactory for therapeutic ultrasound equipment but, by this time, new considerations about exposure and the safety of ultrasound for diagnostic purposes, and especially for obstetric applications, drove a need for the design of power balances with improved sensitivity. By the end of the decade, several balance designs had been reported for which sensitivity of about 1 mW was claimed. Peter Wells and Maurice Bullen, in Bristol in the UK, used a plane aluminium box target suspended by two fine wires about 1 m long, and set at 45° to the downward beam (Figure 13). A horizontal beam caused displacement of the target, which was measured using a travelling microscope. By using a horizontal beam, Wells and Bullen avoided the uncertainties associated with surface tension tethering of the suspending threads. For a reflecting target at 45°, $\cos^2 \alpha = 1$, so the force is the same as that would have been experienced by a fully absorbing target. ±3% accuracy was claimed in a 2 mW beam [29].

At the same time, George Kossoff pointed out that most laboratories were equipped with chemical microbalances that could be adapted to measure ultrasonic power [30]. Such balances were well able to resolve to 0.01 mg, so making measurements in the 1 mW range possible. One example was described in 1970 by Kit Hill, from the Institute of Cancer Research in the UK. (Figure 14) [31]. In his simple design, an aluminium reflector was hung using three very fine nylon filaments from one balance arm, minimising surface tension effects. The target was set at 45° to the downward beam. Reflected sound was trapped by a scattering and absorbing composition of glass wool and rubber at the end of the water-bath. Small amounts of paraffin wax were added to compensate for temperature variations in the weight of the plate. The balance was initially brought to equilibrium and then rebalanced when the downward radiation force was exerted on the target. A calibration of 67 mg W^{-1} was assumed. In Kossoff's review in 1972 he compared the sensitivity of microbalance systems with an absorbing target, an inverted cone reflecting target and an oblique target, concluding that Hill's gave the greatest sensitivity [32].

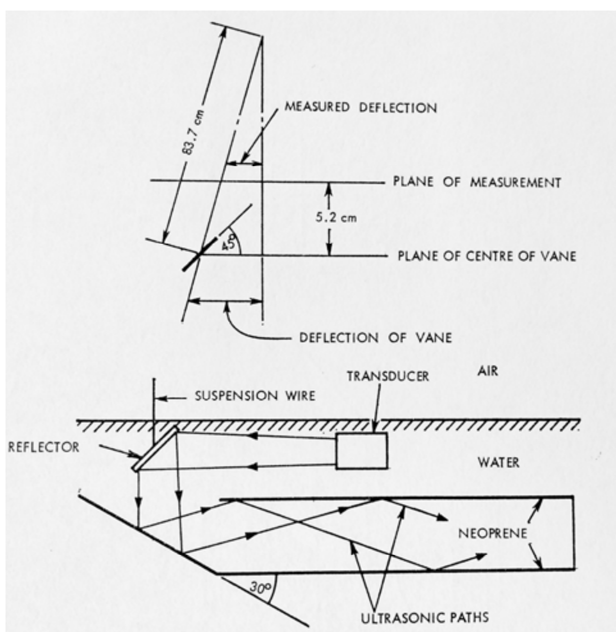

Fig 13. The Wells and Bullen balance. 1964. [31] (©Iliffe Books Ltd)

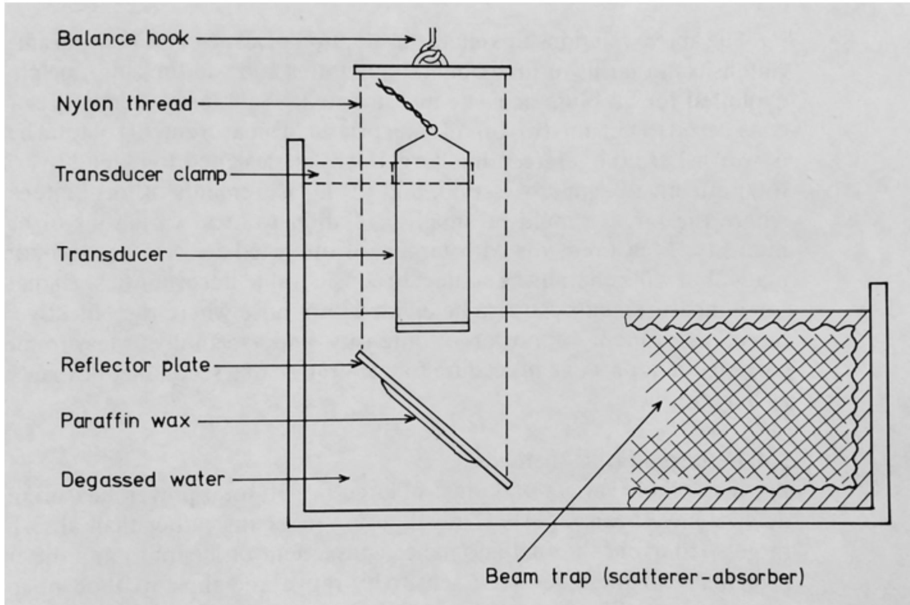

Fig 14. Hill's sensitive power balance based on a chemical microbalance, 1970. [32]

During the 1970s, numerous other laboratories reported their own experiences of approaching the design of sensitive power balances by weighing, with various degrees of success [33,34,35,36,37,38]. In due course, the general availability of force transducers has given rise to simpler, electronic measurement of weight, leading to portable commercial power balances of simpler designs. Nevertheless, sufficient sensitivity for diagnostic measurement remains challenging, and the inherent issues of acoustic design and calibration remain important.

B. Servo-controlled balances

Quantification of the radiation force in the systems so far reported used one of two approaches. In some cases the target was allowed to move under the effect of the incident ultrasonic beam, and the radiation force calculated by computing the mechanical forces associated the new position of equilibrium. Alternatively a mechanical force was applied, typically by the use of weights, to hold the target in a null position.

Force feedback methods developed on these approaches in two ways. First, a sensor was added to detect movement: and an electromagnetic means was introduced to restore the target to its null position.

André Dognon, professor of medical physics at the Faculty of Medicine in Paris, first introduced the concept of force feedback in his laboratory measurements in 1953 (Figure 15), probably as part of his work with Yvonne Simonot on the chemical and biological effects of acoustic cavitation [39]. The target was suspended by a fine wire attached to the coil of a milliameter, causing it to rotate when the target experienced a force. An adjustable current compensated the radiation force by the electromagnetic coupling. Whilst the experimental arrangement, using a small target and subject to standing waves, was never intended to measure total acoustic power, it demonstrated the first attempt to use a null method to measure radiation force by applying an equal an opposite electromagnetic one [40]. Incidentally, Dognon mentioned the problem of measuring power under conditions in which cavitation was an intended consequence, using degassed water and lower powers to calibrate his equipment.

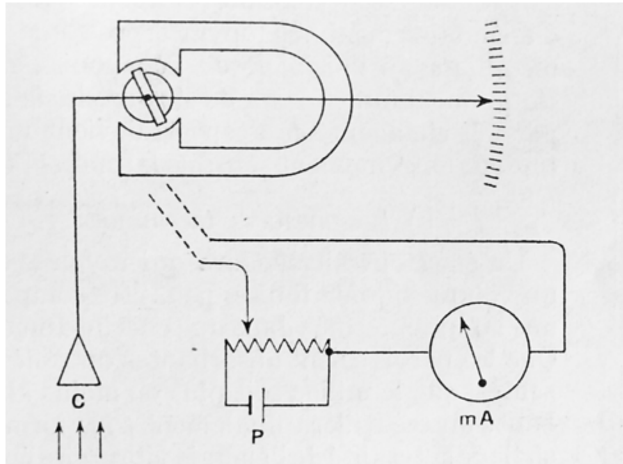

Fig 15. Dognon's electromagnetic feedback technique to measure radiation force. 1953. [39].

A fully servo-controlled balance for ultrasound power was described by Wemlen in 1968 [41]. A portable balance, capable of measuring down to 3 mW, by Lee Dunbar of Grumman Health Systems, New York, was described in 1976 [42]. His design reflected a vertical beam through 45º onto an angled reflecting target, made from a plastic washer with aluminium foil stuck to both sides, adjusted for neutral buoyancy. The target was mounted onto the mechanism of a taut-band meter and was held in a null position using amplified feedback from an occluding photocell, instead of Wemlen's capacitative detector. An alternative servo design was developed in 1982 for a small portable commercial force balance, primarily intended for the measurement of power up to 400 mW emitted from small Doppler transducers. This instrument, available from the UK company Doptek, used a plane absorbing target of Sorbothane® placed in a downward oriented beam [43]. The floating target had a stem holding a bar magnet, the whole tightly contained in a water-filled vessel. An optical position sensor generated a feedback voltage to an external coil, creating a force on the magnet equal and opposite to the downward radiation force. This design overcame the difficulty of placing the feedback coil in the water in the measurement chamber. However, independent calibration was essential since the target could not be assumed to be fully absorbing.

By the early 1970s several large medical physics departments had been established in the National Health Service in the UK, often on a networked regional basis. In some of these, physicists specialised in the technical support of the expanding clinical ultrasound services. They responded to the growing need for a robust and transportable instrument with which to carry out surveys of ultrasonic exposure in clinical use.

Working in the Regional Medical Physics Department in Newcastle, England, Tony Whittingham and Mick Farmery developed a servo-controlled balance to operate in the power range 1-100 mW. Whittingham presented a description of his first servomechanism balance in 1974. In this design, the beam was directed downward into a water bath. A horizontal balance arm carried a 45° degree reflecting target at one end and an opaque rectangular black plastic 'flag' (about 10 mm square) together with a small permanent bar magnet with its axis vertical at the other. Two light beams, partially blocked by the upper and lower edges of the flag respectively, were detected by photo-diodes. These responded to any movement of the flag, generating an out-of-balance signal which, when amplified, caused current to flow in a coil around the magnet, maintaining the balance in a null position. The restoring current was measured by an ammeter, calibrated in mW. This design allowed the use of water as a medium, did not require a plastic window and allowed the electronics to be completely out of the water [44].

Whittingham and Farmery subsequently patented a more rugged design in which the beam was aimed horizontally and, at Farmery's suggestion, the magnet and coil were replaced with the mechanism from a moving coil galvanometer. The reflecting target consisted of a Perspex box with a thin foil window attached to the pointer of the galvanometer and set at 45° to the incoming horizontal beam. The absorbing lining was made of carpet, as effective as the brush absorber in Fiedler's design. Analysis gave a calibration factor of 1.30±0.05 μA mW^{-1} [45,46] (Figures 16,17).

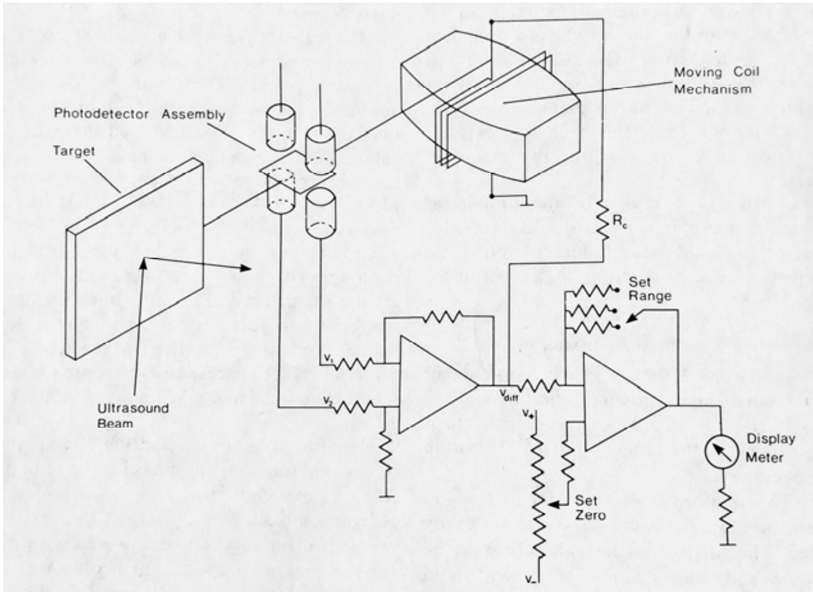

Fig 16. The servo-controlled circuit of Farmery and Whittingham. 1976 [45]. Permission to reproduce: Elsevier Inc.

The horizontal beam orientation introduced one very important attribute. Any sources of thermal instability, for example changed target buoyancy or convection currents, act in a direction orthogonal to the radiation force. By decoupling these forces from those from the ultrasound beam, greater stability could be achieved for measurements of powers down to 1 mW.

There were still some disadvantages with this design. A membrane was needed in order to couple the transducer to the measurement chamber, introducing uncertainties not only from coupling but also in positioning and angulation [47]. The meter mechanism had to be immersed in the liquid in the measurement chamber. Therefore it was no longer appropriate to use water as the transmission fluid, which was replaced with liquid paraffin. Whilst this served to dampen and stabilise the movement, it introduced a frequency-dependent transmission loss between the entry window and the target. Nevertheless, this design was a significant step towards a practical portable milliwatt balance.

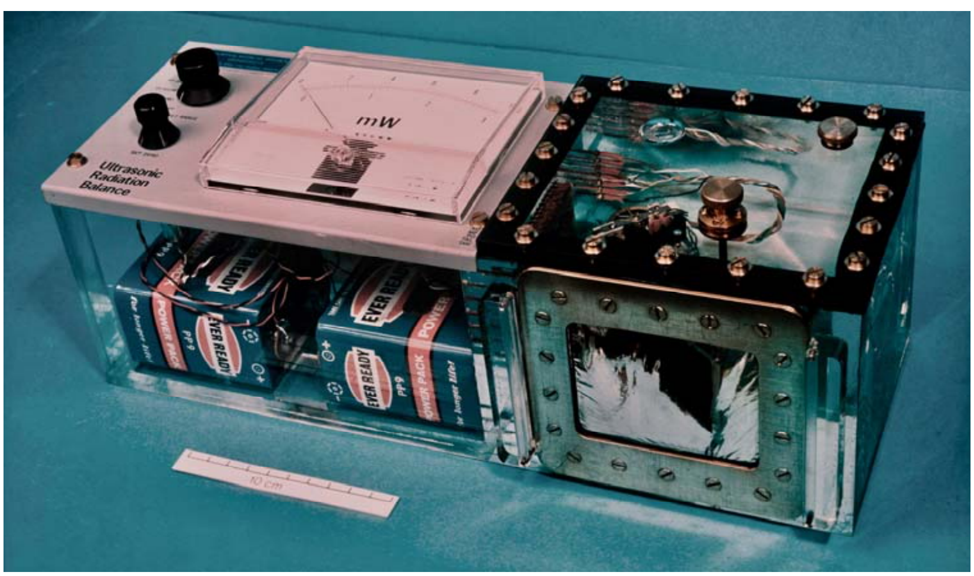

Fig 17. Farmery and Whittingham portable milliwatt power balance [45]

Following Whittingham's work, the first portable power balance to built in the Wiltshire Area Medical Physics Department, based at the Royal United Hospital Bath, UK, was based on the Newcastle design. Constructed by the senior engineer Mike Perkins it introduced a few changes. The transmission loss was minimised by making the overall size smaller and by using a low viscosity transformer oil. A mechanical iris was added to centre the transducer, so avoiding beam placement errors. Otherwise, the general design was the same, including an electronic damping circuit to aid the settling time. From this experience, Perkins carried out a design project for his Open University degree, creating a new design based on similar principles, but with changed geometry [48] (Figure 18).

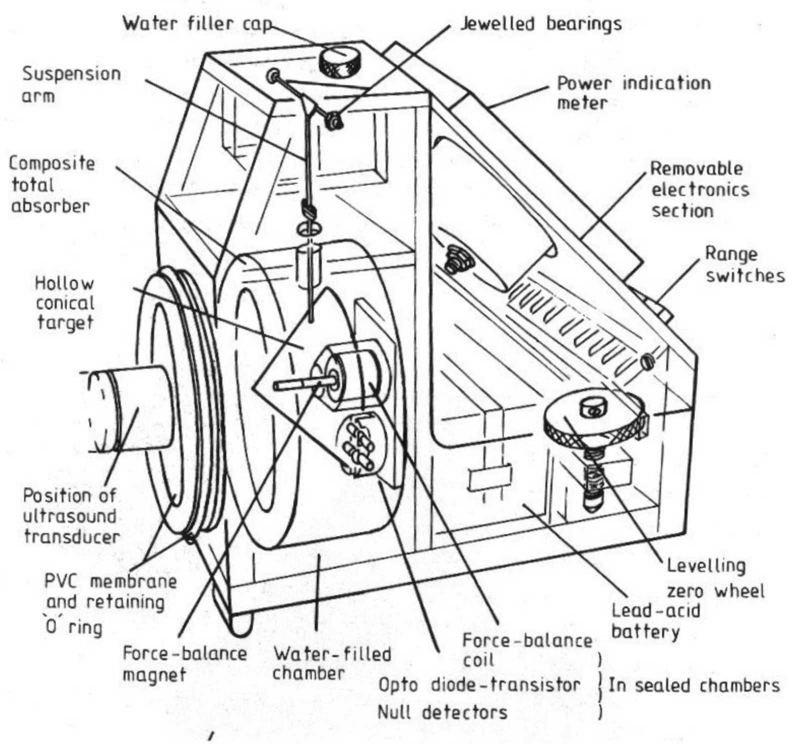

Fig 18. Ghost diagram of the Bath power balance. Perkins, 1989. (Michael Perkins)

The target of Mike Perkins' 'Bath' balance was made of a two cones of very thin monel metal separated by foam spacers. With a 70 mm diameter, it was large enough to ensure measurements could be made on the largest transducers. The cone was suspended from a jewelled bearing within a water-filled chamber with its apex towards an acoustic window. Reflected sound was absorbed in a cylindrical absorber made of carpet. Optical sensors detected movement. The restoring force operated through an external coil and a permanent magnet on the back of the target. The dynamic range enabled it to operate over scales from 0-1 mW up to 0-10W. Electronic zeroing was supplemented by an adjustable screw support, enabling the balance to operate on any surface. The horizontal beam arrangement minimised thermal instabilities. Absolute calibration was enabled using a horizontal balance arm attached to the top of the suspension, on which calibrating weights were hung. In a later simplifying adaptation, the conical reflecting target was replaced by a plane absorbing target made of carpet [49].

The balance was constructed in two compartments. Under normal use as a portable balance these were locked together. For work in the laboratory, it was possible to disconnect the measurement chamber from the electronics, allowing the target assembly to be placed into a separate larger water tank. This arrangement was used to study finite-amplitude loss in water [50]. Neither the Bath nor the Newcastle balance became a commercial product. Nevertheless, such was the interest in power measurement in other medical physics departments that ten more of the Bath design instruments were constructed and provided at cost, one being shipped as far as New Zealand.

With renewed interest in the calibration of ultrasound diagnostic equipment, National Standards laboratories invested in the development of primary standards for ultrasound metrology. For power measurement, this entailed the design of new sensitive radiation force balances, for which the servo feedback design offered significant advantages for sensitivity. One example was the balance developed in 1983 at the National Physical Laboratory in the UK, and designed for the measurement of powers below 200 mW [51].

In due course, the general availability of solid-state force transducers gave rise to simpler, electronic microbalances, leading to portable or semi-portable commercial balances of simpler designs. In one, the solid target was replaced by an absorbing liquid [52, 53]. Even so, in a wide review of radiation force balances that could operate in the diagnostic range, published in 1992, it was noted that 'there are relatively few off-the-shelf radiation force balances …. suitable for measurements of acoustic power of diagnostic equipment'. Only four out of a list of fifteen commercial balances offered measurement capability below 1 mW [54]. Sufficient sensitivity for diagnostic measurement remains a challenge, and the inherent issues of appropriate acoustic design, stability and calibration remain important.

Physiotherapy equipment was less of a challenge, with several reliable commercial balances becoming available. This area also gave rise to one of the simplest and cheapest power balances, designed for quick operational checks of the output of physiotherapy equipment, which soon earned the name of the 'tea caddy'. This was simply an open topped beaker with a balanced vane, set at 45º. An attached pointer gave an approximate indication of acoustic power.

VIII. Regulatory changes

Until the late 1970s it had been accepted that exposure to ultrasound was best quantified by measuring the total acoustic power. Calorimetric methods had failed to dislodge radiation force as the preferred method of measurement. Whilst it was understood that local intensity could be derived from hydrophones, from which total power could be derived by scanning through the beam and integrating, practical and theoretical challenges inhibited this approach as a technique that could be realistically implemented. Hydrophones were used to check frequency and bandwidth, pulse length and frequency, but no more.

Fig 19. The Perkins 'Bath" Balance fitted with a 1 cm aperture for measurements for Thermal Index for scanned beams.

The introduction of regulatory changes in the USA in 1976 served to move the emphasis away from the measurement of total acoustic power. Manufacturers were required to establish 'substantial equivalence' on acoustic output to equipment sold before that date through the Food and Drug Administration 510(k) process. Acoustic power was not included as one of the criteria on which equivalence would be based. Instead, local maxima of intensity anywhere in the exposed region were set, with a range- and frequency-dependent factor applied to compensate for transmission loss through tissue. This approach meant that the emphasis in acoustic metrology moved away from the measurement of acoustic power and towards the use of hydrophones for measuring acoustic pressure and derived intensity. Whilst those concerned with the safe use of diagnostic ultrasound in hospitals and universities continued their efforts to develop improved means for measuring acoustic power, there was no regulatory drive to develop better commercial instruments. The FDA has never added total acoustic power for regulatory purposes.

Regulatory interest in the measurement of power resumed with the publication, in 1992, of the Safety Standard for Diagnostic Ultrasound Equipment, by the American Institute of Ultrasound in Medicine and the National Electrical Manufacturers Association (AIUM/NEMA) [55]. This document established acoustic power as the basis for estimates of tissue temperature rise, creating a definition of the Thermal Index as W/W_{DEG}, in which W is the acoustic power used, and W_{DEG} is the power required to raise the tissue temperature by 1°C under identical conditions of exposure.

Power measurement was reinstated in the metrology agenda. However, the relationship between total acoustic power and tissue temperature rise is not straightforward, especially for focussed, scanned beams from arrays. Some formulae depended not on the total power from the whole array, but instead on the power through a region across the array limited to 1 cm. As a result, apertures to limit the beam had to be added to the means for measuring power (Figure 19).

IX. Accuracy and sources of error

Radiation force remains the preferred method for measuring total acoustic power for medical ultrasonic applications [56]. Decades of experience have given clear evidence of possible sources of error. Absorbing targets are now recommended, using a specified, fully absorbing material which may be used over the full range of medical frequencies. The use of degassed water is mandatory. The distance between source and target should be short to minimise transmission losses at higher frequencies and also from finite amplitude effects, and to limit acoustic streaming. Good temperature control means that the velocity of sound, required for calibration, is known accurately, and convection instabilities are limited. Closed measurement chambers prevent disturbing air currents. Shock-absorbing mounts prevent disturbance from vibrational noise. The relationship between radiation force and power holds in pulsed beams [57], so integration of total power in beams with short pulses may be achieved so long as the mechanical time constant of the balance is long enough. Errors associated with poor beam alignment may be mitigated by careful experimental technique. The most sensitive techniques, used for powers below 1 mW, generally compensate for base-line drift by cycling the power on and off, averaging the differences during each cycle.

Nevertheless, some systematic sources of inaccuracy may remain. The acoustic load on the transducer is different when coupling onto a membrane as opposed to water coupling. The simple relationship between acoustic power and radiation force assumes a plane wave front. As Klaus Beissner has pointed out, the focussed beams used for many medical applications can decrease the radiation force on a target for a given total power. He has proposed appropriate correction factors [58]. There remains the question whether practical targets, absorbing or reflecting, confirm to their ideals when calibration is carried out with an externally applied known force or weight. The design of an absorbing target that will transfer all the force in the direction of the beam remains a challenge. However carefully a reflecting target is manufactured, unsuspected surface acoustic effects may occur. For these reasons, standard test sources are now available to deliver known powers for periodic checks, and regular calibration traceable to a National Standards Laboratory is recommended. An International 'Key Comparison' now allows considerable confidence in this chain of traceability [59].

X. Summary

Radiation force was first used to measure ultrasonic power by Paul Langevin as part of his development of pulse-echo location during the 1914-18 war. The torsion balance was used in many early ultrasonic studies to investigate intensity. The substantial growth of ultrasound therapy in the late 1940s gave rise to new designs for the measurement of total acoustic power, especially in Germany. The standards laboratory at PTB created a balance with a floating target to support its programme of type-testing, widely copied by others. Robust portable commercial balances with mechanical force detectors were also available by 1950. In the 1960s, servo-controlled balances of various designs replaced chemical micro-balances for the measurement of powers in the mW range emitted by medical diagnostic equipment. Regulatory changes in the USA briefly inhibited interest in power measurement, which was reinstated following the establishment of the thermal index for safety management in the 1990s.

Author	Date	Measurement method	Beam direction	Target
Langevin[1]	c.1915	torsion pendulum	Horizontal	Reflecting 90° plane
Boyle [4]	1928	torsion pendulum	Horizontal	Reflecting 90° plane
Baumgardt[13]	1937	movement	Vertical up	Reflecting 30°cone
Atlas-Werke	c.1939	movement	Vertical up	Reflecting 30°cone
Pohlman[17]	1939	gravitational	Vertical up	Absorbing 90° plane
Fiedler[18]	1950	mechanical	Vertical down	Reflecting 45° roof
Atlas- Werke[20]	c.1950	mechanical	Vertical down	Absorbing 45° roof
Oberst & Rieckmann[22]	1952	float movement	Vertical down	Reflecting 30°cone
Dognon[40]	1953	servo feedback	Vertical up	Reflecting 90° plane
Henry[24]	1957	float movement	Vertical up	Absorbing 30°cone
Kossoff [30]	1962	float movement	Vertical down	Reflecting 30° cone
Wells [29]	1964	movement	Horizontal	Reflecting 45° plane
Hill [31]	1970	gravitational	Vertical down	Reflecting 45° plane
Dunbar[42]	1976	servo feedback	Horizontal	Reflecting 45° plane
Farmery & Whittingham[45]	1978	servo feedback	Horizontal	Reflecting 45° plane
Shotton [27]	1980	float movement	Vertical down	Reflecting 30° cone
Cornhill[43]	1982	servo feedback	Vertical down	Absorbing 90° plane
Perkins [48]	1989	servo feedback	Horizontal	Reflecting 45° cone

Table 1. Selected examples of radiation force balance designs. The angle given is that between the target and the beam direction.

Acknowledgements

I wish to add my sincere thanks to Dr Volker Wilkens from Physikalische-Technischen Bundesanstalt (PTB) in Braunschweig, for providing documents, photographs and further details from the earliest years of PTB's activities in metrology for medical ultrasound. In addition, Dr Bajram Zeqiri, of the National Physical Laboratory, Teddington UK, was kind enough to read and comment on an earlier draft. The kind permission of Illife Books Ltd to reproduce figures 12 and 13 from B Brown and D Gordon, *Ultrasonic Techniques in Biology and Medicine* (1967) is acknowledged.

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ULTRASONICS METROLOGY III. THE DEVELOPMENT OF THERMAL METHODS FOR ULTRASOUND MEASUREMENT

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The temperature rise that occurs when energy from the transmission of ultrasound is absorbed in the propagation medium has been used to quantify exposure from the earliest laboratory investigations in the 1930s. This article is an account of how thermal and calorimetric techniques evolved as part of ultrasound exposure measurement for medical applications.

I. Early thermal probes

Thermal methods for assessing variations in ultrasonic intensity were explored in several laboratories during the 1920s and 1930s. Soon after Alfred Lee Loomis opened his private laboratory in Tuxedo NY, William Richards, from the Chemistry Department of Princeton University, spent time there exploring whether a copper-constantan thermocouple might be used to measure intensity. He wrapped one junction in rubber, the second in a glass tube, and noted the change in differential temperature as the exposed rubber was heated by ultrasonic absorption. Instabilities in the very intense beams under investigation (which the called 'supersonic' to distinguish them from ultrasonic beams of lower intensity) caused them to abandon the thermal approach in favour of a hydraulic system connected to a capillary tube [1]. Not long afterwards, Nikolai Malov, a Moscow physicist who was investigating ultrasound beams in different liquids, reported greater success using a 15µm iron wire resistance thermometer embedded in rubber [2].

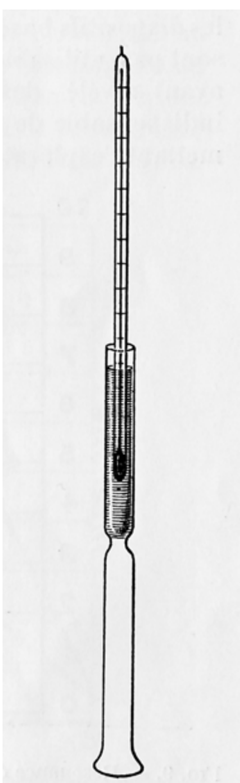

Fig 1. André Dognon's *tube calorimétrique*. 1937 [3].

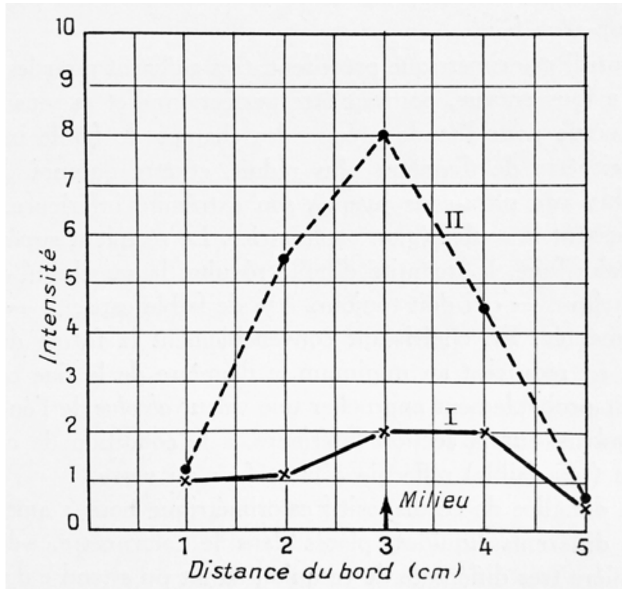

Fig 2. Relative intensity beam profile using a *tube calorimétrique.* [3]

André Dognon was the first to investigate the temperature rise caused by ultrasound in a medical context [3]. He was agrégé professor of physics in the Faculty of Medicine in Paris, working with the Biancani brothers, assistants in radiology at the *Hôtel Dieu*. They agreed with Richards' assessment that cavitation could cause radiation pressure measurements of power to be unstable, so they investigated calorimetry as an alternative. Using a glass tube filled with paraffin oil as a crude thermal sensor, which Dognon called a '*tube calorimétrique'*, they measured the temperature increase generated in the oil using a small mercury thermometer (Figure 1). Whilst such an arrangement is of no value for absolute measurements of intensity, they could use the technique to set up resonant conditions for maximum output and were even able to plot an approximate beam profile (Figure 2). By using a standard exposure time (10 s or 30 s) they compared the temperature rise in small volumes of various liquids and also in soft tissue samples. In this way they demonstrated that water and gelatine gel heated very little. On the other hand, wax and fat heated quickly. A 2 ml volume of liver or brain tissue in the tube increased in temperature by about 10 deg C after 30 s exposure [4].

II. Thermal probes for acoustic intensity

The rapid expansion in the use of therapeutic ultrasound after the Second World War ended in 1945 drove a need for improved methods for ultrasound measurement. By 1950, Reimar Pohlman (1907-1978), a Siemens physicist who had pioneered ultrasound therapy and who was by then living in Switzerland, reported several new initiatives in which thermocouples, usually copper-constantan, were mounted for applications relevant for therapeutic applications of ultrasound (Figure 3) [5]. The simplest device had been reported by Pätzold and Born in 1947 and used to plot the sound distribution in a focussed beam they used for treatment [6]. It consisted of a thermocouple mounted at the end of a cannula, the tip encased in wax or Plexiglas (polymethyl methacrylate). A similar thermal probe was used at Physikalische-Technischen Bundesanstalt (PTB) in Braunschweig, in their programme of type-testing for therapy ultrasound equipment, introduced by Oberst and Rieckmann [7] (see Figure 3a). The absorbing sphere on the PTB probe was about 1 mm diameter, and this enabled detailed mapping of the relative intensity profiles in the near field (Figure 4). Pohlman and Fiedler also built a remarkable thermocouple array, which is shown in Figure 3c. Copper and constantan wires were soldered into a mesh, and selected junctions were encased in an absorber, creating an array of heated and reference junctions with which to evaluate the beam [8].

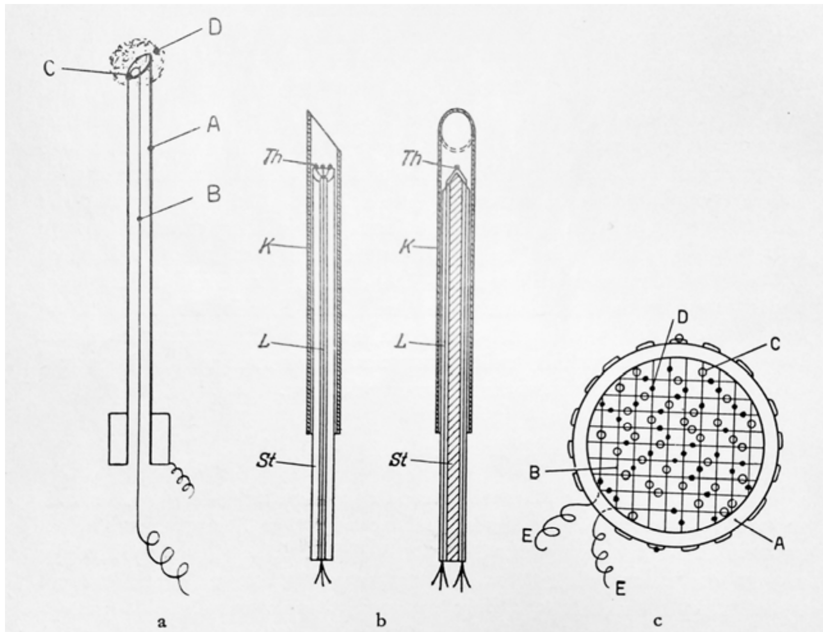

Fig 3. Pohlman's thermocouple probes, 1950 [5].

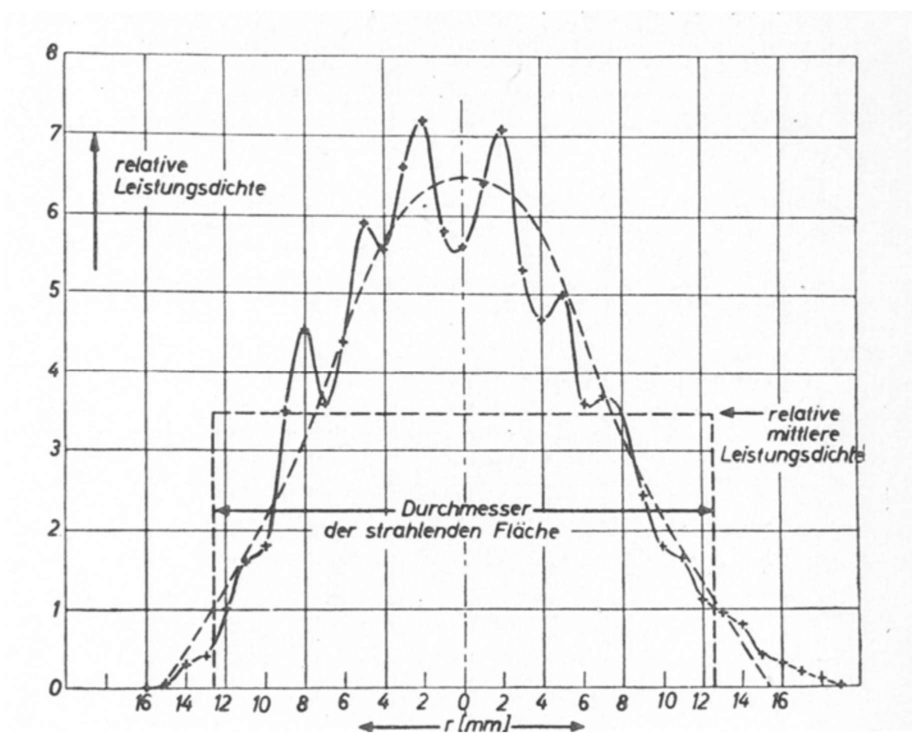

Fig 4. Near-field relative intensity profile (continuous line) in the near field of a 25 mm diameter transducer, measured using a thermal probe at PTB, Brunschweig,1952. The dashed curve was calculated assuming radial symmetry and omitting diffraction. [7]

Figure 3b shows another thermal probe reported by Pohlman, intended to determine the temperature rise within tissue during exposure, and so with no absorbing tip [9,10]. But, as Oberst and Rieckmann pointed out, there were problems that would continue to challenge later investigators. The needle material on which the thermocouple was mounted was inevitably exposed to ultrasound as well as the surrounding tissue, so heated from absorption and from friction, sometimes to a greater extent than the tissue itself. Later workers realised that the thermally conductive thermocouple leads drew heat away, a problem that became more significant with small heated volumes at a beam focus. As a result, the errors in a number of later experiments on tissue temperature rise were large. Eventually, fragile, miniature, unsheathed thermocouples were the only ones being recommended to make reliable *in-situ* measurements of temperature.

Elsewhere, there was interest in measuring the higher intensities used for surgery. The physicist William Fry founded the Bioacoustics Laboratory at the University of Illinois in 1946 and, working with Francis Fry and others, pioneered the use of intense focussed ultrasound for tissue ablation. To underpin this work they carried out a detailed theoretical and experimental analysis of the performance of thermocouples as thermal sensors [11,12,13]. Their probe, a thermocouple junction etched to 0.013mm, was mounted in a capsule of castor oil that acted as a liquid acoustic absorbing medium. With a diameter of about 7.5 cm, the capsule was large enough to allow the passage of the sound beam without disturbance from the ring support (Figures 5,6).

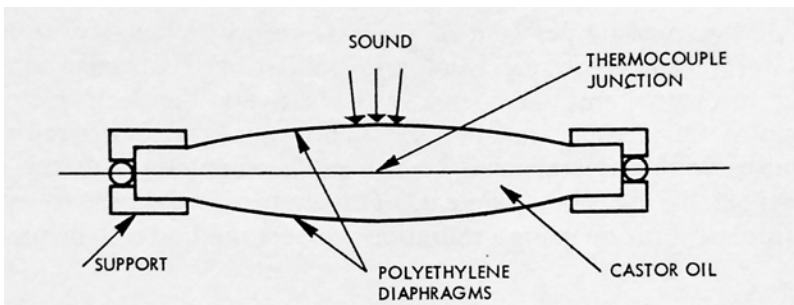

Fig 5. Intensity probe using a thermocouple in castor oil. Fry & Fry. 1954. [12] (©Iliffe Books Ltd)

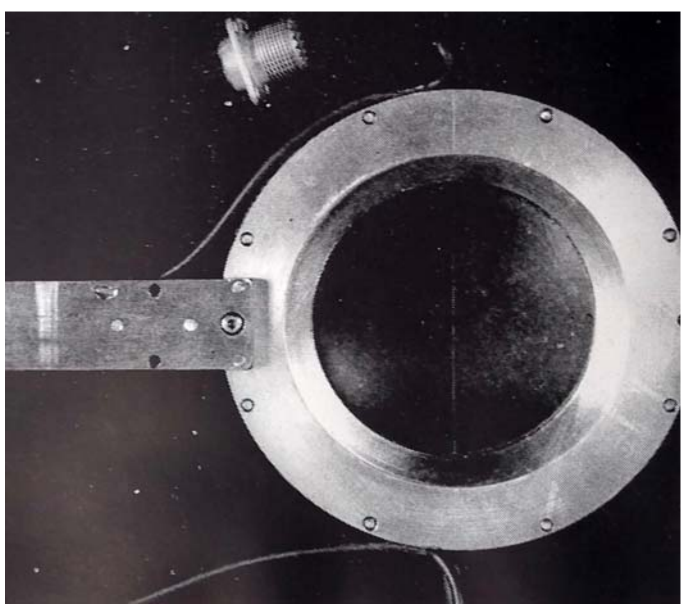

Fig 6. Beam's eye view of the Fry thermal probe.

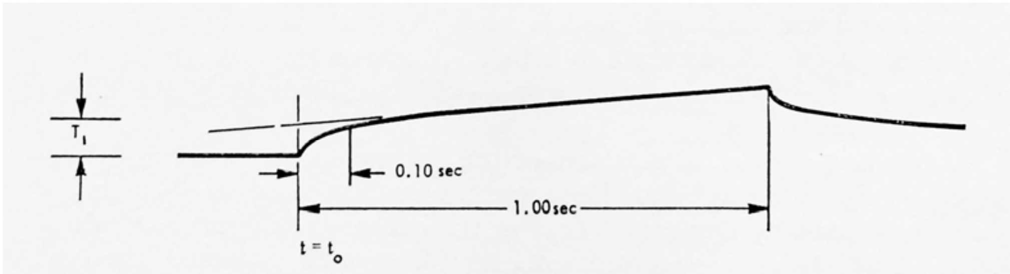

Fig 7. Temperature-time profile from Fry's thermocouple probe. (©Iliffe Books Ltd)

When exposed to ultrasound for 1 s, the thermocouple voltage viewed on an oscilloscope showed two distinct time periods, an initial rise due to viscous heating of the thermocouple followed by a linear region due to absorption of ultrasound (Figure 7). The local intensity could be determined from the gradient of the second, linear, region, dT/dt. Absolute values of sound intensity, I, were calculated from the acoustic absorption coefficient of the castor oil, μ, and its specific heat capacity, ρC, using $\mu I=\rho C(dT/dt)$. Intensities up to 20 W cm^{-2} at 980 Hz were measured using this device. In principle this method allows absolute measurement of intensity. Nevertheless, uncertainty about the specific heat capacity of the heated liquid lead to a recommendation of cross-calibration against a radiation force radiometer using a spherical target. Conversely, knowing the intensity allowed the ultrasonic absorption coefficient of tissue to be determined under a variety of conditions [14].

Several thermal probes have been described which use embedded thermistors in place of thermocouples as the heat sensing elements, once they became generally available in the 1950s. Care was needed to select a limited temperature range over which a linear response could be assumed. The thermistor has much higher sensitivity to temperature changes than a thermocouple so may be used in beams with lower intensities. The thermistor is coated with a thin layer of absorbing medium. Morita (1952) constructed thermal probes approximately 2 mm in diameter by embedding a thermistor (0.3-0.5 mm dia.) in various sound absorbing materials including pitch, paraffin wax, vinyl resins and varnish [15]. The sensitivity of the probes was reduced when the diameter of the probe was reduced to less than 1 mm. A sound intensity of 1 W cm^{-2} gave a temperature rise of approximately 20 °C.

Martin and Law constructed a range of thermistor probes using various glues, varnishes and resins to form the absorbing coating [16]. The pattern of temperature rise when exposed to a burst of ultrasound was similar to that of the thermocouple probe, i.e. a short rise due to shear viscosity near the boundary between the thermistor and coating material, followed by a linear region due to ultrasound absorption. Over the range of intensities used for physiotherapy, the rate of temperature rise is in the linear region during the first two hundred milliseconds of exposure and thus is proportional to intensity.

Thermistor probes were generally larger than those using thermocouples. However, the response of either is slow in comparison with a piezoelectric hydrophone and so they can only be used to measure time-averaged values of intensity and need to be calibrated against an alternative measurement method. Nevertheless, for those of us who were making and testing our own ultrasonic transducers in the 1970s, the improved sensitivity of an epoxy-coated thermistor over a thermocouple offered a cheap and convenient device with which to make a quick assessment of acoustic output and beam pattern.

Small thermal sensors continue to find a place in acoustic metrology. In 1996, a method was described in which the intensity was derived from the steady-state temperature measured at the rear surface of an absorbing block, thermally insulated by air [17]. Calibrated against hydrophones, they can provide a simple and sensitive means for investigating intensities of the order of 1 mW cm^{-2}. Volker Wilkens of PTB described the performance of a thermal probe of this type [18,19]. Measurement uncertainties for time-averaged intensity lie in the range ±20% to ±30%, comparable with estimates from hydrophone measurements. Identification of the location and value of the maximum time-averaged intensity can be almost impossible using a hydrophone in the switched, multi-functional fields now used in diagnostic ultrasound. Under these conditions, a calibrated thermal sensor may be preferred. Measurements are limited by thermal noise, estimated to be 3 mK. For the sensors used in his study, this translates to noise equivalent intensities of between 0.7 W m^{-2} and 4.1 W m^{-2} (0.07-0.41 mW cm^{-2}).

III. Calorimetry

Thermal probes were developed to measure local intensity, either in a liquid bath or within tissue. An alternative thermal technique is for the measurement of total acoustic power from the temperature rise caused by the total deposition of acoustic energy in an absorbing target. The attraction of calorimetric methods lies in the capture of energy directly, so no particular regard is needed for the details of the acoustic beam. On the other hand, complete conversion of acoustic energy to heat is needed, and any uncontrolled losses of thermal energy must be excluded. These design criteria lead to large target volumes, and consequent small temperature rises, generally limiting the technique to measurements above about 100 mW at best.

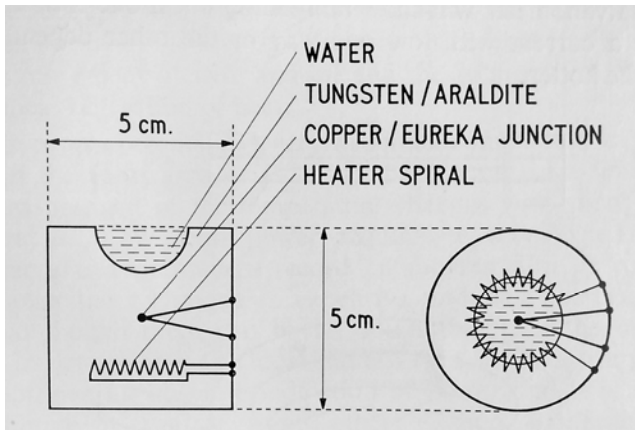

Fig 8. The Bristol ultrasonic calorimeter

A. Non-flow calorimetry

An early calorimeter was described by Peter Wells. He had been appointed by Herbert Freundlich, head of the Medical Physics Department in Bristol, UK, in 1959. Freundlich was familiar with ultrasound. His father, the German colloid chemist, Herbert Freundlich senior, had studied thixotropy of colloids and gels using ultrasound at the Kaiser Wilhelm Institute in Berlin. In 1932 he had suggested that bone marrow might be heated while whole bone itself was unaffected and proposed its use for thermal therapy [20]. He emigrated after resigning for refusing to dismiss associates who were not racially appropriate, moving first to University College, London and then on to Minnesota in 1938. His physicist son stayed in England, studied in Cambridge before establishing the first department of medical physics at Bristol General Hospital.

One of Peter Wells' first tasks was to develop methods of calibration for ultrasound exposure of the inner ear, used by John Angell-James for the treatment of Meniéres disease [21]. The Bristol calorimeter was a simple and effective means for measurement of the power emitted by the miniature transducers used in this treatment (Figure 8). Sound was absorbed in an epoxy resin block, loaded with tungsten powder to increase its absorption coefficient, which was used to embed a copper-constantan thermocouple and an electrical heater coil. The temperature reached after a specific time was calibrated against temperature increases using electrical power. Error sources included incomplete transfer of energy at the water epoxy interface, and unknown thermal loss to the surroundings. Nevertheless, 6% agreement with radiation force measurement of acoustic power was claimed [22].

The Bristol calorimeter was a rather simple example of a non-flow calorimeter, in which the beam is operated for a short time during which the temperature change is measured. A more complex example was described by Curt Wiederhielm in 1956, which could be used over a frequency range from 0.5 MHz to 20 MHz, either to measure total power or to probe beams. The acoustic energy was captured in water held in a reflecting flask and the timed increase in temperature noted. Air bubbles induced mixing. An aperture facilitated beam plotting for intensity. Calibration used an integral electrical heater. Poor sensitivity limited the use of this device to powers of 600 mW and above [23].

Harold Stewart, from the Bureau of Radiological Health, FDA, Maryland, gave a review of calorimeters in 1974 [24]. One novel arrangement used a thermal sensor placed at the focus of a parabolic mirror. A second design is shown in Figure 9. The absorbing liquid, carbon tetrachloride, was placed in a conical cup which ensures an extended path-length to enhance absorption. Acoustic streaming assists thermal mixing. The temperature difference between the exposed liquid and the surrounding water-bath is measured using chromel-constantan thermocouples. A coil is immersed for electrical calibration. Other non-flow calorimeters have also been described [25,26,27], including an extremely simple device suitable for use by physiotherapists to monitor the output of their equipment, nicknamed 'calorimeter in a coffee cup' [28].

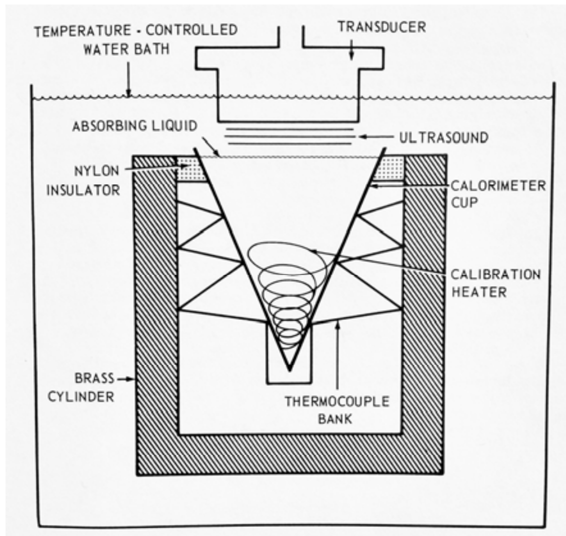

Fig 9. Stewart's steady-state calorimeter for therapy power measurements. 1974. [24]

B. Flow calorimetry

Greater sensitivity can be achieved with flow calorimeters, which are operated in steady-state conditions, the measurement of interest being the temperature difference between two streams [29]. An early example of such an ultrasound calorimeter, reported by Szilard, used two chambers, one heated electrically, while the second chamber is exposed to ultrasound, and the thermal responses are compared [30]. By the mid-1970s there was enough interest in exposure measurement for diagnostic systems for national standards laboratories to invest in the design and construction of sensitive devices for power measurement as low as 1 mW.

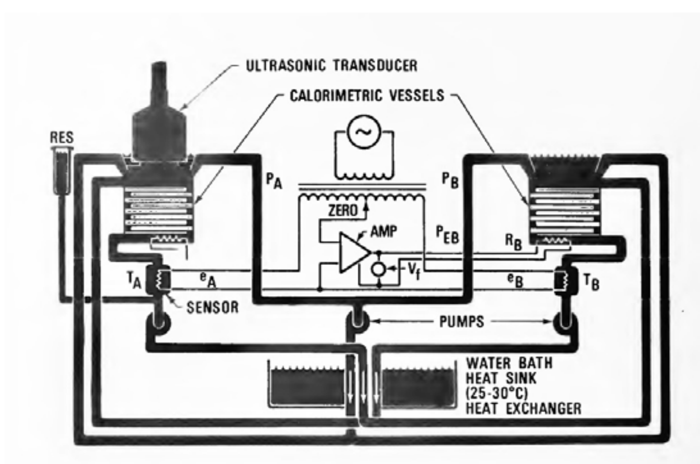

Fig 10. The NBS two-chamber servo-controlled calorimeter 1976. [31] (Reprinted courtesy of the National Institute of Standards and Technology, U.S. Department of Commerce).

The US National Bureau of Standards (NBS) flow calorimeter was built specifically to measure the output power from medical ultrasonic systems [31] (Figure 10). Two thermally-insulated flow chambers were connected in series, one exposed to ultrasound and the second heated electrically. Solid acoustic absorbers were made from four wafers of silicone elastomer and four wafers of butyl rubber. Heat transport used an inert per-fluorinated liquid. An electronic feedback system automatically adjusted the power of the heater, based on the temperature imbalance between the outflows from the two chambers. The whole calorimeter assembly was placed in a temperature-controlled water bath. The sensitivity of the design allowed the measurement of power over a range of 0.5 mW to 10 W. The main loss of heat, as much 1.5%, was judged to be by conduction through the flange and water well, and a correction was applied.

It was recognised during this study that damped piezoceramic transducers, designed to generate very short pulses for diagnostic work, were lossy, and that a proportion of the electrical energy was converted to heat in the transducer itself. A correction for this additional source of heat was introduced, based on the assumption that a damped transducer was 50% efficient. As will be noted below, this transducer self-heating later became a challenge both for manufacturers designing to FDA limits on acoustic output, and for the formulation of an informative 'thermal index' with which to advise operators of potentially hazardous increases in tissue temperature.

With care, the NBS calorimeter achieved an uncertainty of ±(7% + 0.2 mW). However, considerable care was needed with the operation of the calorimeter and in the application of corrections for thermal leakage, both ingress and loss. The project served to emphasise the challenges that had to be overcome before calorimetry could take its place as a viable alternative in serious ultrasonic metrology. Comparative reviews discussed the influence of thermal parameters in calorimetry [32], and its comparison with radiation force for measuring acoustic power [33].

C. Expansion calorimetry

Dognon's '*tube calorimétrique'* was not the only thermal approach to ultrasound exposure measurement that emerged in the early experimental phase of ultrasound. In 1935, Johannes Greutzmacher reported a method using a form of air thermometer to test the performance of a novel focusing transducer [34]. A glass rod was dipped into the focal zone and connected outside the water-bath to an air-filled glass bulb. On exposure, the thermal expansion of the air-filled bulb was communicated to a water manometer, indicating changes in intensity.

This approach to expansion calorimetry was given improvement by Igor Mikhailov (1907-1894), from the ultrasonics laboratory in the University of Leningrad, in 1957 [35,36]. A simplified diagram of his device is shown in Figure 11. The ultrasound beam enters the tapered liquid-filled measurement capsule through a membrane. An absorbing material fills the tip. Heat deposited in the liquid causes it to expand up a calibrated capillary tube. Developments of this simple device included a double-walled vacuum vessel to minimize heat loss, and the addition of an electrical heater for calibration, depending on the time taken for the liquid to move between two fixed points on the capillary.

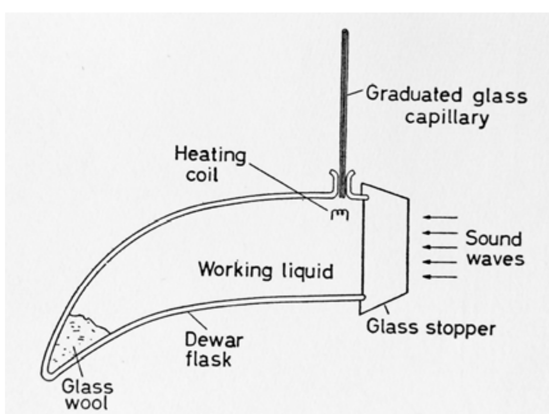

Fig 11. Mikhailov's expansion calorimeter, 1957. [35]

Most recently, Adam Shaw, from the NPL, has described how a target undergoing thermal expansion due to ultrasonic heating can be used to resolve the problem of power measurement in the highly focused beams used in high intensity ultrasound surgery [37,38]. The principles that underpin radiation force for the measurement of power depend on a plane-wave assumption, and this becomes progressively invalid with increased focusing. Under these conditions, the challenges of calorimetry become worth overcoming. Shaw's solution used a liquid target of castor oil, floating in a water bath. Instead of measuring the expansion of a closed liquid target as had been done before, the volume is allowed to expand causing a change in buoyancy. The rate of change in weight was measured using a commercial balance with 1 mg resolution and is proportional to the power. As with other calorimetric methods, the instrument may be calibrated using an electrical heater. The performance was validated for frequencies between 0.8 MHz and 3 MHz and for powers from 1 W to 300 W. Overall uncertainties at 1 MHz were estimated as ±3.4%.

D. Pyroelectric calorimetry

The other recent development in sensitive ultrasound calorimetry has been in the measurement of power in the milliwatt range by the use of a large area pyroelectric sensor [39,40]. Acoustic power is deposited in a specially designed highly-absorbing solid material, causing its temperature to rise. The temperature is measured with a polyvinylidene fluoride (PVDF) membrane, 28 μm thick, with large area gold electrodes. Heating at the interface between the absorber and the PVDF generates a pyroelectric voltage across the electrodes. The device can be operated differentially, the voltage compared with that generated by an identical, unexposed sensor in order to reduce the effects of background vibrations. Non-normal angle of beam incidence prevents the formation of standing waves. The electronics respond to the rate of change of pyroelectric voltage which is maximum immediately following switch-on and is proportional to acoustic power. The overall power to voltage conversion factor is typically 0.23 V W^{-1}, and the response is linear to within ±1.6% over a range of power from 1 mW to 120 mW. The device has a flat frequency response to within ±4% between 2.5 MHz and 10 MHz.

IV *IN-SITU* TEMPERATURE RISE AND THE THERMAL INDEX

The previous sections have concerned the historical development of thermal and calorimetric methods for the measurement of the acoustic intensity and power of ultrasonic beams. We now turn to a different question: what are the temperature rises caused within tissue when exposed to ultrasound?

Interest in such matters was renewed in 1992 with the publication of the so-called Output Display Standard by the American Institute of Ultrasound in Medicine and the National Electrical Manufacturers Association, AIUM/NEMA (ODS) [41]. The broad intention behind the creation of this American Standard was two-fold. Firstly, it set out a process by which the user of an ultrasound scanner could carry out their own risk assessment for exposure, based upon the displayed value of two safety indexes, Thermal Index (TI] and Mechanical Index (MI). Secondly, it enabled the FDA regulatory process to increase the upper intensity limits for some applications, especially for obstetrics, placing the onus on the user to use the safety indexes to manage exposure safely.

The Thermal Index was designed to be numerically equivalent to the greatest worst-case steady-state temperature rise in degrees Celsius anywhere in the exposed tissue. The soft-tissue formulations for temperature rise using the Pennes bio-heat equation to take account of perfusion [42], were based on the work of Wesley Nyborg of the University of Vermont to predict temperature rise in homogenous soft tissue [43]. Some details were published later [44], including the work of Ed Carstensen to predict temperature rise in bone [45]. Conceptually simple, the objective proved challenging to formulate, requiring a number of simplifying assumptions to reach a set of six formulae to cover the main conditions of clinical exposure. The best review of the rationale and development of the safety indices was later written by John Abbott, of ATL, who had taken a leading role in the AIUM/NEMA project [46]. Three separate conditions were considered: for heating bone, either close to the transducer or at the focus, and for exposure only of soft tissue. In each case, temperature rise was evaluated for both stationary and scanned beams. The underlying principle assumed that temperature rise could be predicted linearly from acoustic power. Nevertheless, some formulae required measurements of local intensity using hydrophones as well as the direct measurement of acoustic power. The tissue models used in the calculations were highly simplified, including the use of a particularly low soft tissue attenuation coefficient (0.3 dB cm^{-1} MHz^{-1}), a higher assumed value for the absorption coefficient, a fixed convection loss due to perfusion, and the assumption that half the incident energy was reflected at a soft-tissue bone interface.

One major outcome of the publication of the ODS was a progressive emphasis towards estimated tissue temperature rise as the primary metrological parameter in assessing ultrasound safety. Until its publication there had been a broad consensus that safety limits should be based on an acoustic quantity, the AIUM recommendation of a maximum time-averaged intensity of 100 mW cm^{-2} being widely accepted. As new equipment progressively broke this convention, especially in Doppler mode, and the FDA allowed equipment to operate with intensities considerably greater than this, questions about *in-situ* temperature rise started to replace questions about acoustical measurements of intensity and power, measured in water.

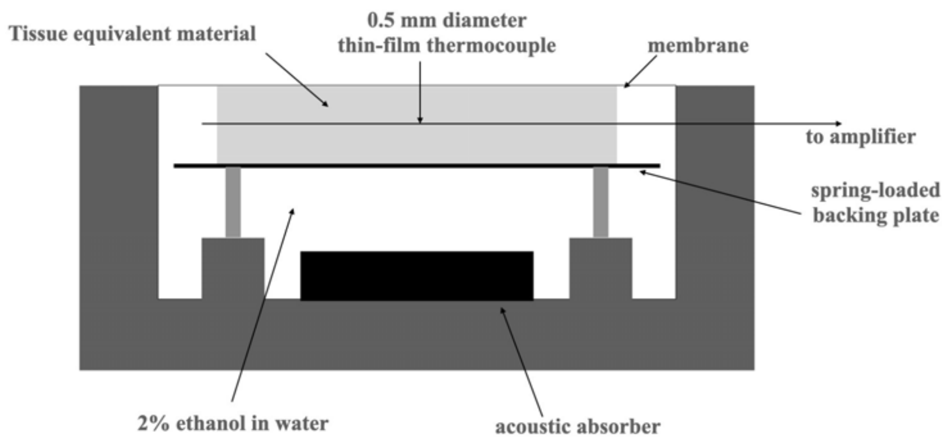

Fig 12. Thermal Test Object (redrawn). [49]

Reservations were voiced about the validity of many of the simplifying assumptions underpinning the calculations of Thermal Index, largely from scientists other than those in USA who had taken part in the formulation of the ODS. This resulted in several initiatives to explore the temperature rises that could occur in tissue exposed at permitted diagnostic exposures [47], and to compare the temperature rises implied by the Thermal Index with measured temperature increases in tissue-equivalent materials [48,49].

One response to the need for a standard means to measure the temperature rise in tissues in a diagnostic field was the Thermal Test Object, developed by Adam Shaw and David Bacon at the NPL UK (Figure 12) [50,51]. A small thin-film thermocouple was sandwiched between two blocks of tissue-mimicking material and held in place by a sprung backing-plate. While this arrangement was made available commercially, it suffered from ageing of the gel-based tissue mimic material. Nevertheless, the approach was usefully developed into a phantom designed to mimic the neonatal head to estimate the temperature rise at several locations [52].

V. Transducer self-heating

Our own measurements of temperature rise caused by the absorption of ultrasound led to a new appreciation of the significance of the transducer as a heat source [53]. We were exploring a new way of characterizing the output of diagnostic ultrasound equipment, using biophysical phenomena to complement purely acoustic quantification of exposure [54]. In exploring the location at which the temperature rise was a maximum, using a thermocouple embedded in agar gel, we observed that the temperature was always greatest when the probe was in close contact with the transducer, and never when it was placed in the region of the focus. Further investigation quickly showed that there were a number of conditions for which commercial scanners could cause the transducer surface to rise to a temperature where discomfort and pain was experienced by the skin.

As noted above, it had been appreciated that damped piezoceramic transducers are inefficient, with perhaps 50% of the electrical energy being converted to heat in the transducer. Design criteria at that time emphasised the limits set by the FDA on estimated *in-situ* intensities, time-averaged and pulse-averaged. There was no limit placed on total power or, equivalently, intensity at the transducer. As a result, the electrical power had been set without heed to the thermal load, resulting in the possibility of excessive increases in transducer surface temperature. In addition, the acoustic lenses used to improve focussing were often made from lossy elastomeric materials, resulting in heating localised at the surface.

Once this was appreciated, manufacturers were quick to limit the electrical power, controlling the surface temperature. Thermal cut-out mechanisms were introduced for oesophageal transducers. Specific limits for surface temperature for diagnostic transducers were established by the IEC for operation both in air and in contact with the skin [55]. Operation in air is now limited to less than 50°C and a 43°C limit is placed on operation in contact with tissue. One study showed that coupling to the skin resulted in a reduction to between 43% and 87% of the temperature rise that was reached in air, depending on the transducer and operating mode [56]. This means that the stronger of the two IEC controls is that set for contact operation. It is often these thermal limits that place an overall cap on acoustic output, rather than any limit placed on acoustic intensity.

The measurement of the maximum surface temperature operating in air is most readily accomplished using an infra-red radiometer, carefully positioned, or an infrared camera (Fig 13). Rob Hekkenberg, from TNO in The Netherlands, has reported the use of a thermal camera for both surface temperatures and the thermal distribution with depth using a split tissue-equivalent phantom [57]. The measurement of the surface temperature in contact with tissue offers more serious metrological challenges, requiring the use of a phantom with thermal and ultrasonic properties close to tissue and a thermocouple that minimises errors in temperature measurement and, moreover, is positioned where the temperature increase is greatest. The use of K-type rather than T-type thermocouples is now recommended to avoid distortion of the temperature distribution due to the high thermal conductivity of copper. For example, Justine Calvert used a 12 μm butt-bonded K-type thermocouple to explore surface temperatures generated by trans-vaginal transducers, for which excessive surface temperatures might pose a particular risk [58].

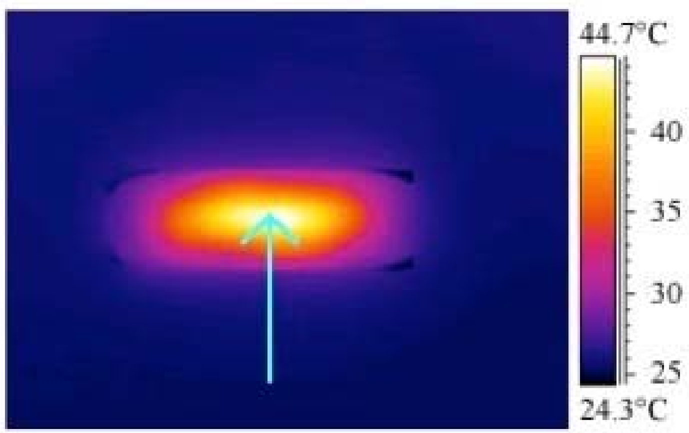

Fig 13. Thermal image of the front surface of an ultrasound array in Doppler mode, operating in air. The arrow identifies the point of maximum temperature rise. [57] (British Medical Ultrasound Society)

VI. Summary

Thermal methods have been used to probe ultrasound fields since the first laboratory experiments in the 1920s and 1930s. Mercury thermometers, air thermometers, liquid expansion, thermocouples, thermistors, pyroelectric sensors and infra-red radiometry have all been used for thermometry. The measurement of local intensity using the temperature rise in an absorber of known physical properties was established in the 1950s, when possible sources of error began to be better understood. A small sensitised thermal probe remains a useful non-direction device to map intensity in ultrasound beams. The challenge of measuring total acoustic power has given rise to many alternative designs of calorimeter. Sensitivity at the milliwatt level remains a design challenge. Starting in the 1960s, with increasing emphasis on thermal safety for diagnostic applications of ultrasound, the formulation of materials that are acoustically and thermally equivalent to a range of tissues, their use to measure the small increases in temperature caused by diagnostic ultrasound and the extrapolation to estimates of *in-vivo* temperature rise in practical situations has remained an experimental and theoretical challenge. The largest heat source is the transducer itself.

The kind permission of Illife Books Ltd to reproduce figures 5 and 7 from B Brown and D Gordon, *Ultrasonic Techniques in Biology and Medicine* (1967) is acknowledged. We are most grateful to Dr Bajram Zeqiri, of the National Physical Laboratory, Teddington UK, for his useful suggestions.

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AUTHORS' PROFILES

The biographical profiles of the authors are given elsewhere in this issue: Ultrasound - the first fifty years and Ultrasonic Metrology I.

www.mpijournal.org ISSN 2306-4609