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].

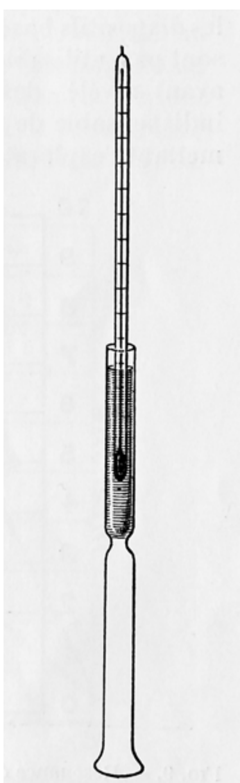

Fig 1. André Dognon's *tube calorimétrique*. 1937 [3].

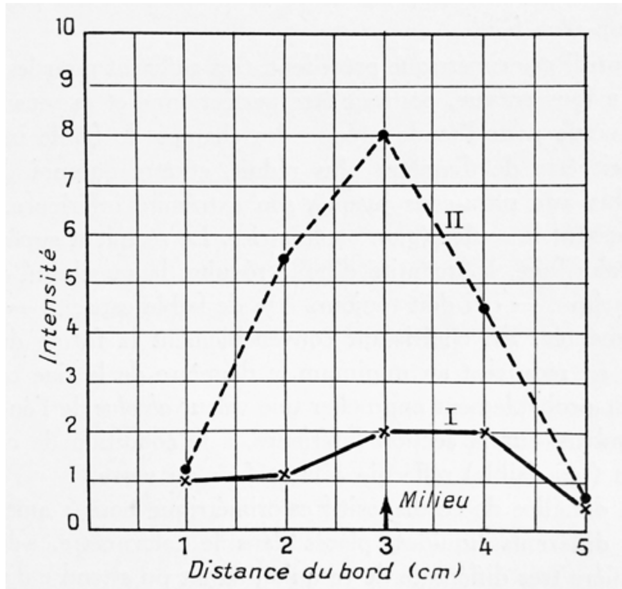

Fig 2. Relative intensity beam profile using a *tube calorimétrique.* [3]

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].

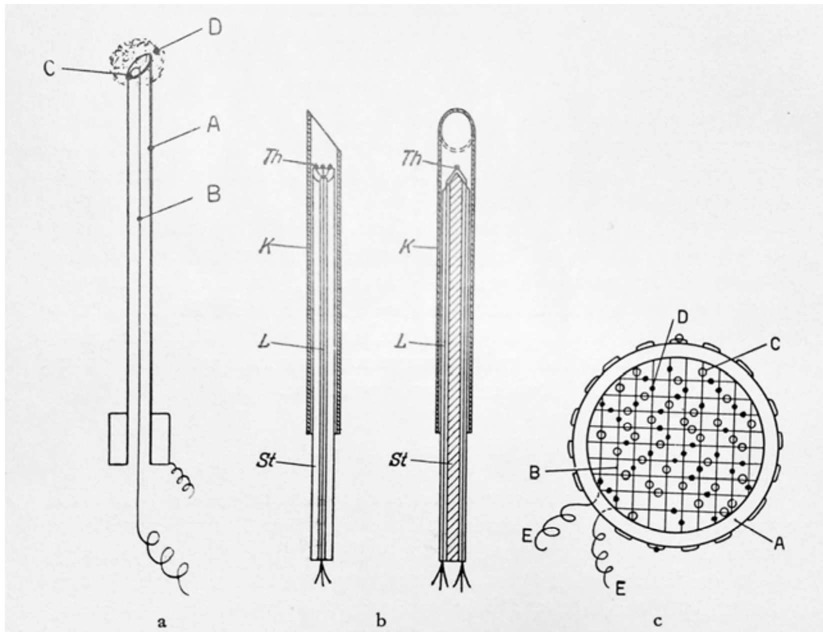

Fig 3. Pohlman's thermocouple probes, 1950 [5].

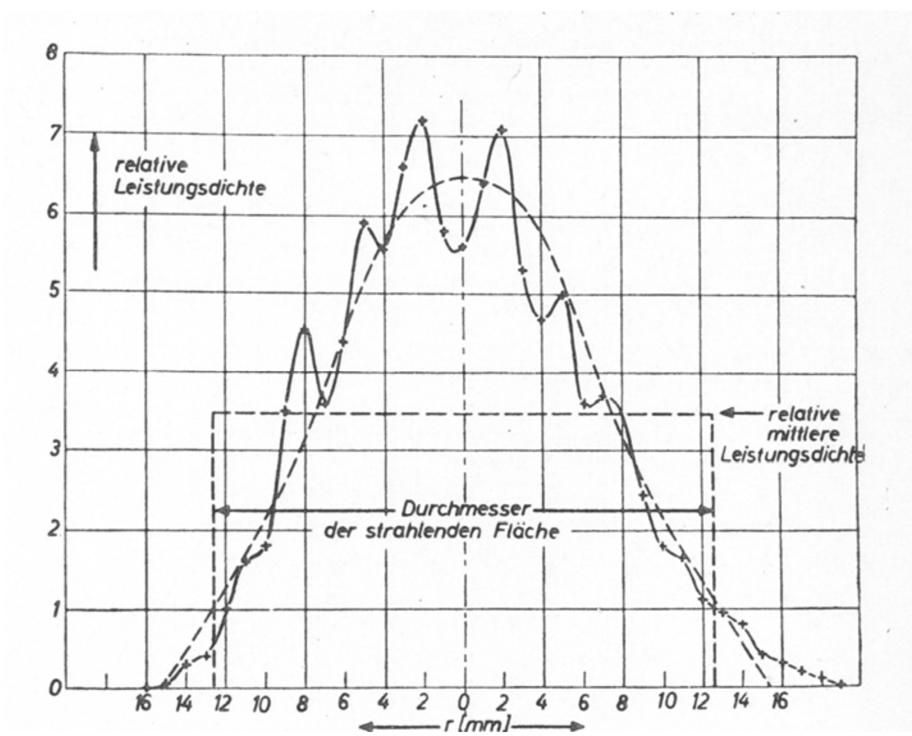

Fig 4. Near-field relative intensity profile (continuous line) in the near field of a 25 mm diameter transducer, measured using a thermal probe at PTB, Brunschweig,1952. The dashed curve was calculated assuming radial symmetry and omitting diffraction. [7]

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).

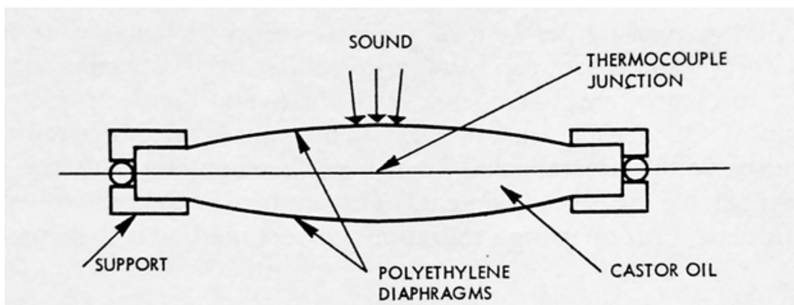

Fig 5. Intensity probe using a thermocouple in castor oil. Fry & Fry. 1954. [12] (©Iliffe Books Ltd)

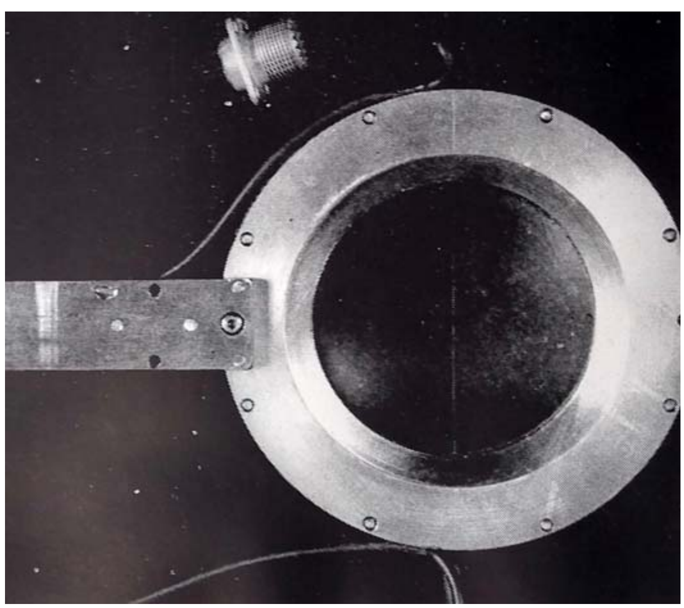

Fig 6. Beam's eye view of the Fry thermal probe.

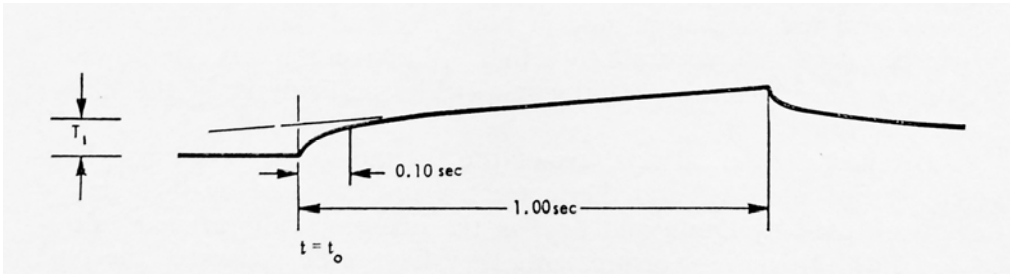

Fig 7. Temperature-time profile from Fry's thermocouple probe. (©Iliffe Books Ltd)

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, $\mathrm{d}T/\mathrm{d}t$. 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 $\mu I=\rho C(\mathrm{d}T/\mathrm{d}t)$. 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.

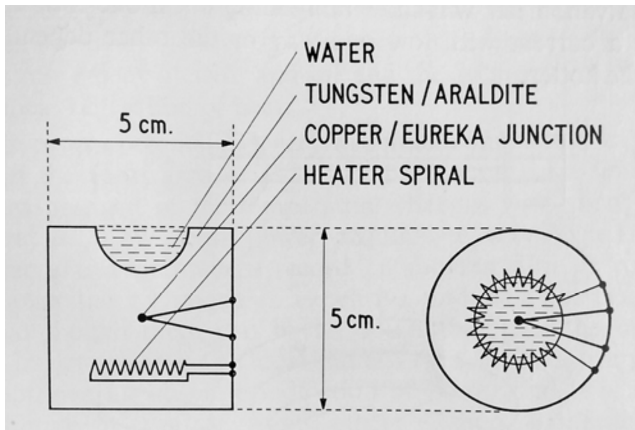

Fig 8. The Bristol ultrasonic calorimeter

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].

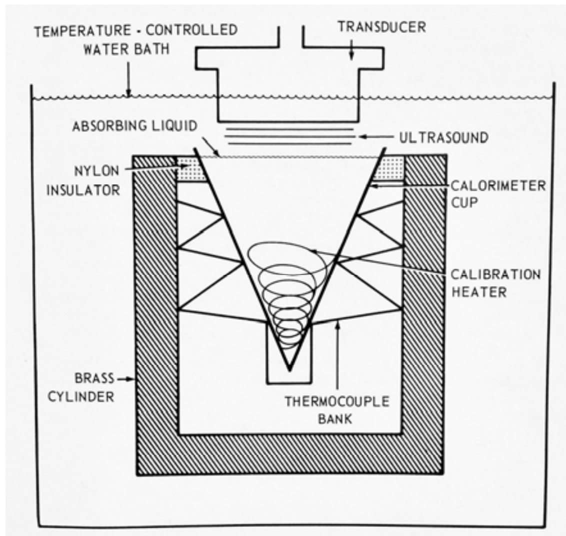

Fig 9. Stewart's steady-state calorimeter for therapy power measurements. 1974. [24]

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.

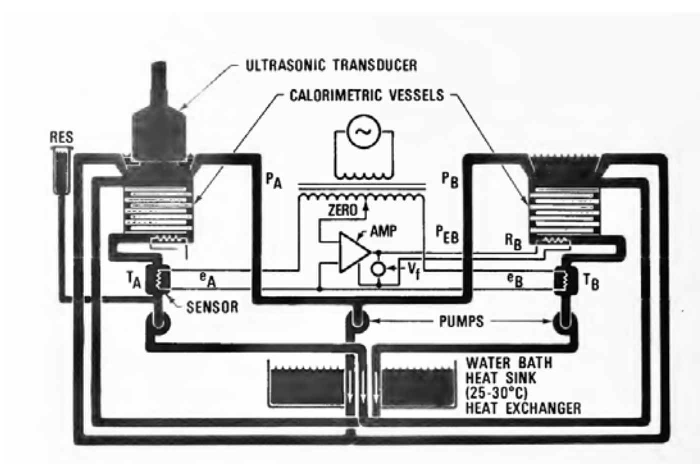

Fig 10. The NBS two-chamber servo-controlled calorimeter 1976. [31] (Reprinted courtesy of the National Institute of Standards and Technology, U.S. Department of Commerce).

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.

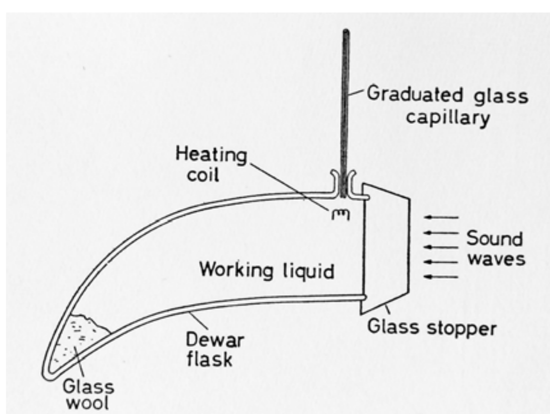

Fig 11. Mikhailov's expansion calorimeter, 1957. [35]

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.

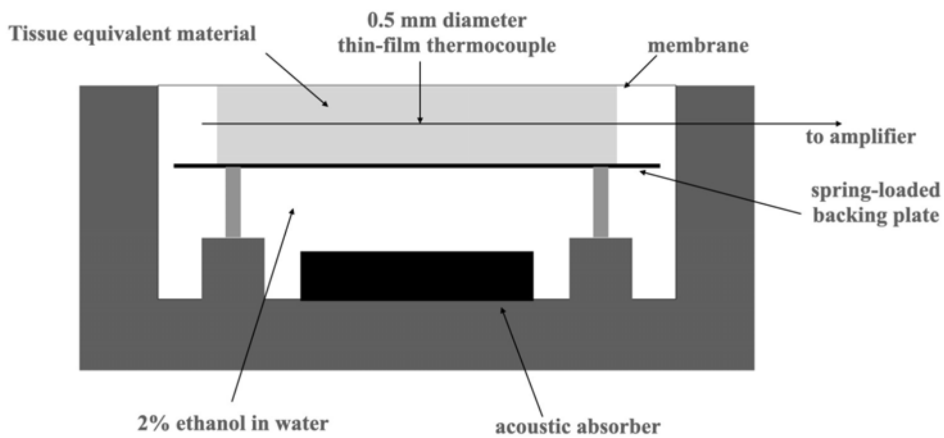

Fig 12. Thermal Test Object (redrawn). [49]

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].

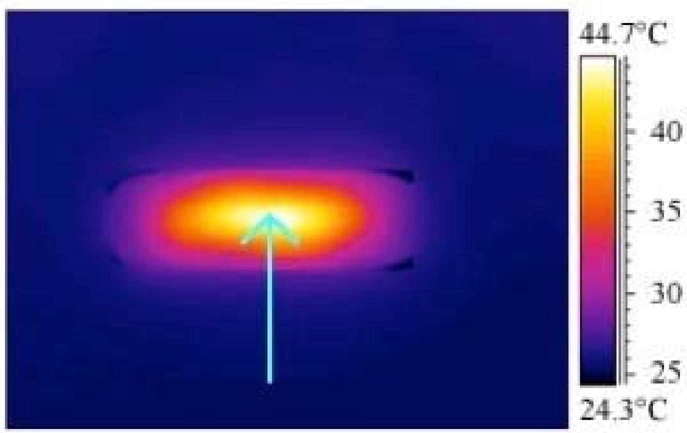

Fig 13. Thermal image of the front surface of an ultrasound array in Doppler mode, operating in air. The arrow identifies the point of maximum temperature rise. [57] (British Medical Ultrasound Society)

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.

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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.