MEDICAL PHYSICS
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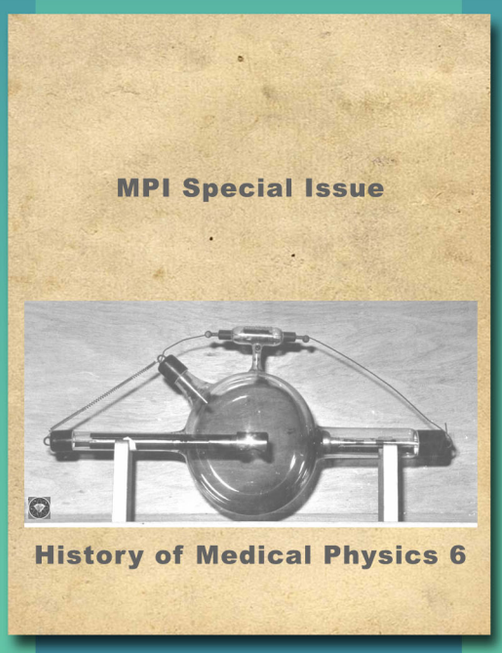

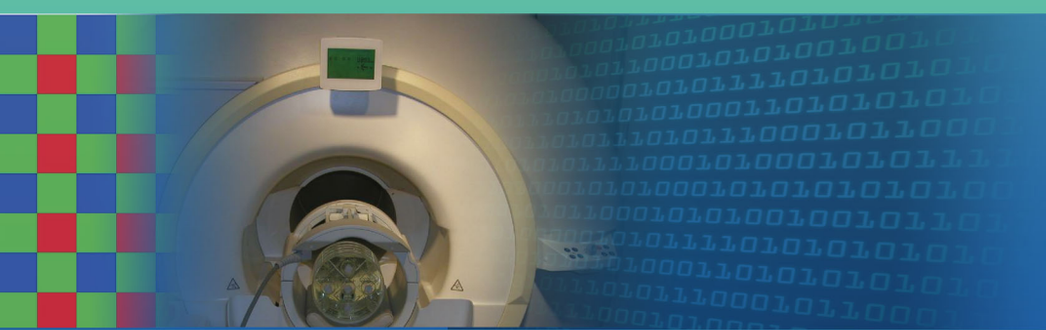

The Journal of the International Organization for Medical Physics

Special Issue 6, May 2021

MPI

MEDICAL PHYSICS INTERNATIONAL

THE JOURNAL OF

THE INTERNATIONAL ORGANIZATION FOR MEDICAL PHYSICS

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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 scientific, educational and professional nature. The e- journal is available free of charge to IOMP members.

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EDITORIAL

Slavik Tabakov, Perry Sprawls and Geoffrey Ibbott

MPI Special Issues Co-Editors

This Special Issue dedicated to Medical Physics History includes five new articles from the IOMP History Project. It continues the coverage of the History of Medical Ultrasound. Part II of this history, published here, covers the diagnostic ultrasound imaging, Doppler ultrasound and some therapeutic uses of ultrasound. An interesting part of this Special issue is the description of the difficulties that faced engineers and developers in creating equipment for the healthcare market. This is supported by a case study with the Diasonograph. In order to see the link between this Special Issue and the previous one (also related to Ultrasound), we have re-published here the introduction to the Ultrasound History Part I, made by Dr. Francis Duck, MBE, who coordinated both parts of this Medical Ultrasound History project.

We are very happy to see that the overall popularity of the History Project is growing - each MPI Special Issues has over 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 History topics extensively covered in MPI so far include:

Special issue 1- http://www.mpijournal.org/pdf/2018-SI-01/MPI-2018-SI-01.pdf

*X-ray Tubes Development; *Film-Screen Radiography Receptor Development; *History of Medical Physics e-Learning Introduction and First Steps

Special issue 2 - http://www.mpijournal.org/pdf/2019-SI-02/MPI-2019-SI-02.pdf

*Fluoroscopic Technology from 1895 to 2019; *The Scientific and Technological Developments in Mammography; *Review of the Physics of Mammography

Special issue 3 - http://www.mpijournal.org/pdf/2020-SI-03/MPI-2020-SI-03.pdf

*History of Dental Radiography ; *The History of Contrast Media Development in X-Ray Diagnostic Radiology; *Medical Physics Development in Africa

Special issue 4 - http://www.mpijournal.org/pdf/2020-SI-04/MPI-2020-SI-04.pdf

*A Retrospective of Cobalt-60 Radiation Therapy; *The Many Steps and Evolution in the Development of Computed Tomography; *Medical Physics Development in South-East Asia; *History of Medical Physics Education and Training in Central and Eastern Europe

Special issue 5 - http://www.mpijournal.org/pdf/2021-SI-05/MPI-2021-SI-05.pdf

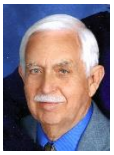
* Ultrasound – The First Fifty Years; *Measurement of Acoustic Pressure and Intensity Using Hydrophones; * Measurement of Acoustic Power and Intensity Using Radiation Force; *Development of Thermal Methods for Ultrasound Measurement

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 papers describing the MEFOMP activities, etc.

The Content of the Special History Issues of the Medical Physics International (MPI) Journal 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: PART II – INTRODUCTION

Francis Duck

Formerly, University of Bath

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.

Earlier this year, we presented the first four articles of a history of medical ultrasound physics in a special supplement of Medical Physics International: http://www.mpijournal.org/pdf/2021-SI-05/MPI-2021-SI-05.pdf. These articles formed 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. The first article documented the first fifty years of ultrasound up to 1950, during which time 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. The remaining three articles in this first supplement 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.

This second supplement includes five more articles on medical ultrasound, which move the history towards its clinical exploitation. In the first, Norman McDicken and Carmel Moran give a succinct overview of some early developments, which emphasises the unique aspects that were recognised by the early pioneers, namely precise dimensional measurement, an inherent ability to create slice images with a high frame rate and the ability to track structural movements in real-time, all attributes that were challenging to achieve using x-ray imaging at the time. The next two articles take a more detailed look at the difficulties that faced engineers and commercial companies in translating this potential into financially successful equipment. Tony Whittingham gives a detailed historical account of the creation and development of the Diasonograph, a unique ultrasound B-scanner, designed and developed in Scotland, that established ultrasound imaging as an integral part of modern gynaecological and obstetric care. Tom Szabo brings his personal knowledge as a research engineer to the description of the creation of Hewlett Packard's phased array cardiac scanner. Both these accounts record the contributions of the many talented engineers whose design skills broke new ground and continued to innovate. Also, independently, both narratives address the enlightened management approach that created time and finance for an enterprise whose outcome was still uncertain. The fourth article, by Peter Hoskins, takes a look at the history of another aspect of ultrasound, the use of Doppler shift in evaluating haemodynamics in normal and diseased cardiovascular structures. At first separate from imaging, Doppler brought new diagnostic information in the audio Doppler-shifted spectrum, and eventually pulsed Doppler opened the door to colour Doppler imaging in which anatomical and physiological information merged. Finally, Gail ter Haar's article is a reminder that, during the same period

of time that saw advances in ultrasound applications for diagnosis and for therapy, challenges were also being met and overcome for exploiting the destructive power of high intensity ultrasound for surgery and ablative therapy

You will find many details of the history of medical ultrasound in these articles that have not been documented elsewhere. Many histories start with the first arrival of an ultrasound instrument in the hands of a creative clinical user, and appropriately celebrate their achievements. Here, we look behind the scenes to those whose prior engineering skills and vision placed new tools in the hands of these clinicians.

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. In addition, may I add my personal thanks to Kevin Martin for his careful editorial scrutiny of these articles, helping to reduce to a minimum the residual errors that can inevitably slip though.

HISTORY OF MEDICAL ULTRASOUND

WN McDicken and CM Moran

Medical Physics, Centre for Cardiovascular Sciences, University of Edinburgh, Edinburgh, Scotland, UK

I. Introduction

This section deals with the technology that used single piezoelectric ceramic transducers to obtain information on soft tissues. It also seeks to identify lessons learned from applications as techniques advanced to multi-element array transducers. In the early days, engineering and clinical projects were pursued in universities and companies throughout the world. It is worth remembering that ultrasound technology was developed at the end of the vacuum tube valve era. Stability and reliability were problematic with valve circuitry as was the shear bulk of the instruments. Not surprisingly, digital technology impacted the field in an ever- increasing way.

To explore the very large output of scientific work, use has been made of bibliographies in text books and international conferences proceedings. Several text books with a historical content provide extensive bibliographies of medical ultrasound [1-7]. Reference 3 is a very large cumulative bibliography with 19,453 entries from pre-1950 up to 1978 [3]. These publications also give an appreciation of the large amount of biological research which was a feature of that era. International conferences, although they are selective in content, convey the range and enthusiasm of participants around the world and a flavour of early research both technical and clinical [4]. The internet also provides many scientific papers and excellent reviews but the amount can be over-whelming, but usually worth pursuing. All figures are reproduced from McDicken (1991) [8].

II. A-mode and M-mode scanners

A basic instrument used initially in medical ultrasound was called an A-scanner. With this type of unit, a single piezoelectric ceramic transducer was employed to transmit a short pulse of ultrasound, typically 2 or 3 cycles in length, along a narrow directional beam into the body. After transmission, this unit then quickly switched to a reception mode using the same transducer to detect echoes from targets of interest. The very high speed of ultrasound in tissue resulted in a rapid collection of echo data e.g. in 130 microseconds for a target at a depth in tissue of 10 cm. This rapid data collection is central to all pulse imaging techniques which can therefore operate at high frame rates. High frame rates are a very valuable feature of medical ultrasound. Using an average speed of sound in soft tissue of 1540 m/sec to calibrate scanners gives acceptable errors for range measurement in virtually all clinical applications.

There were a large number of A-scan instruments available in the 1960s since in the engineering industry they were manufactured for detecting echoes from flaws in engineering components – they often went by the name of ‘flaw detectors’. They were basically oscilloscopes with a pulse generator and receiver added. Part of the folklore of ultrasound is that engineering apprentices immersed their feet in buckets of water and used the flaw detectors to detect their bones. A few clinical applications were developed for example measurement of shift of the brain midline due to head injury or measurement of eye-ball length (Figure 1).

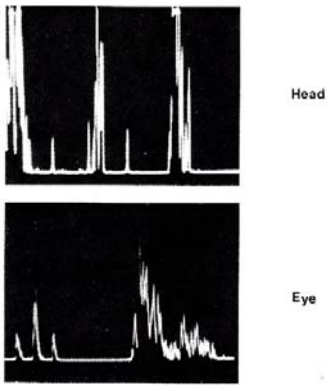


Fig. 1 An A-scan trace from the head. The echoes from the midbrain are positioned between the strong echoes from the skull. An A-scan trace from the eye shows cornea and lens echoes on the left and orbital echoes on the right.

A-scanners are no longer purchased as stand-alone clinical instruments. Pioneering applications of engineering in medicine were often not complex but were difficult to interpret. The A-scan facility remained a feature of imaging machines for around 30 years and were widely used to make biparietal diameter measurements in obstetrics. It was felt that greater accuracy of measurement could be obtained with it rather than with a B-scan image. High quality image storage displays had still to be incorporated into scanners.

The A-scan mode was considered to be safe for both the operator and the patient since MHz frequencies are rapidly attenuated in air and there had been no reports of harmful effects. Studies of bio-effects of ultrasound ran in parallel with development of imaging technology.

Early A-scan applications showed that it was essential to exclude even a very thin film of air at the transducer/tissue interface. Exclusion was readily achieved with oil or gel in the interface. This simple means of exclusion permitted good acoustic coupling of single element or array transducers. Ease of coupling made contact scanning possible even when the transducer was moved across the skin. Stand-alone A-scan units never found much routine clinical application but they did provide a means of gaining experience of the use of ultrasound in tissue. Echoes from static structures remain fixed on a display. Motion of structures such as heart walls and valves could be observed and recorded (Figure 2)**.** The latter technique, known as an 'M-mode', is fast and is still a feature of modern echocardiography. M-mode traces can be presented simultaneously with other physiological signals such as an ECG, phonocardiogram and Doppler ultrasound flow spectrograms (Figure 3).

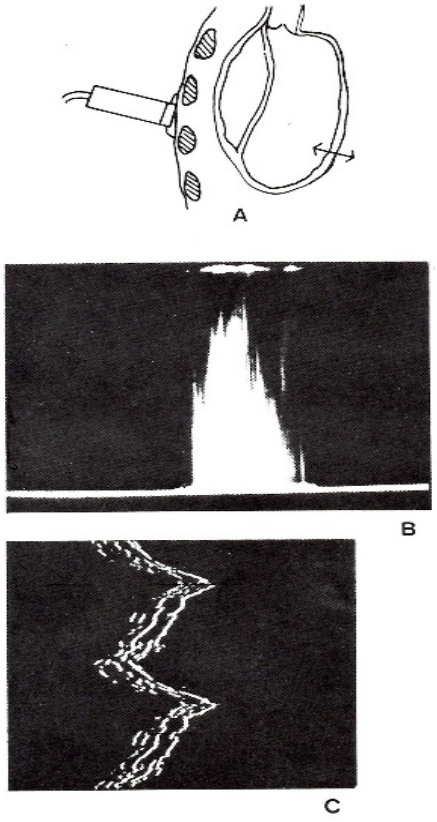


Fig. 2 Principle of M-scanning. (a) Transducer directed at moving structure of interest and held fixed, (b) echoes may be observed on an A-scan display but this does not give a record of motion, (c) echo dots sweep up the screen to provide a trace of position versus time.

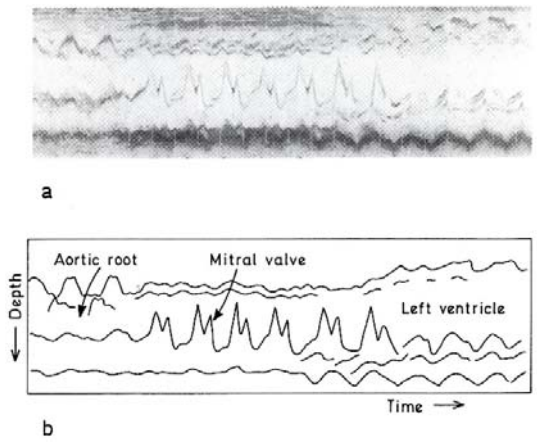


Fig. 3 Adult heart action recorded by M-scanning: write-out on a fibre-optic chart recorder. The direction of the beam was slowly altered during the recording.

III. Water-bath B-mode scanners

Early ultrasound imaging scanners often employed water-baths to couple the machines to the patient. In a water-bath scanner, a depth of water was interposed between the transducer and the patient either by immersion of the patient or by enabling access via a thin membrane on the side of the tank (Figure 4).

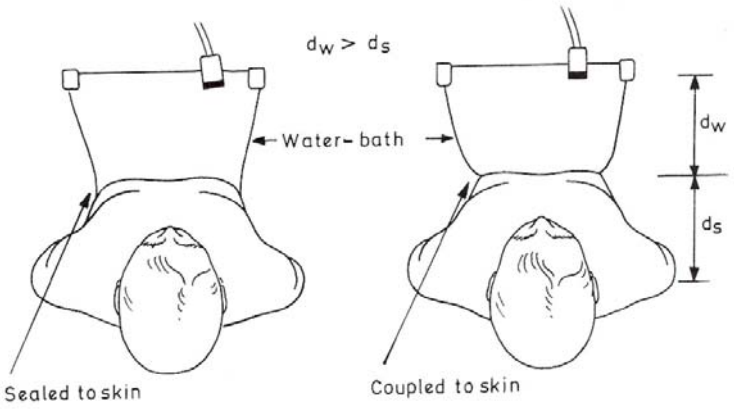


Fig. 4 Water-bath coupling techniques.

To avoid multiple reflection echoes in the water overlapping in time with echoes from tissue, the depth of water had to be greater than the depth in tissue of the structures of interest. The walls of the bath were often lined with rubber which was an excellent acoustic absorber reducing weak background multiple reflection signals. Water is a uniform low attenuation medium in which a transmitted field could form with good focusing. The velocity of ultrasound in water at room temperature is close to the average velocity of ultrasound in soft tissue i.e. 1540 m/s. It is easy to appreciate why water was used as the coupling medium in the first imaging machines. Water-bath scanners were popular for scanning soft tissues which were mobile such as breasts, eyes, testicles and infants.

As technology developed, lessons were learned which benefitted the next generation of scanners. A water-bath scanner, known as the 'Octoson' is a particularly good example of this process (Figure 5). In this machine, 8 large saucer-shaped transducers, each of diameter 7 cm, were mounted on a cradle in a water-bath. The position of the cradle could be altered by a drive mechanism to select the scan-plane. The 8 transducers could oscillate singly or collectively to perform a simple or compound scan. In a later version of this machine, the single-element transducers were replaced by annular array transducers to give an extended focal range. The large aperture transducers provided well-focused beams. The patient lay on a membrane across the top of the water-bath (Figure 6).

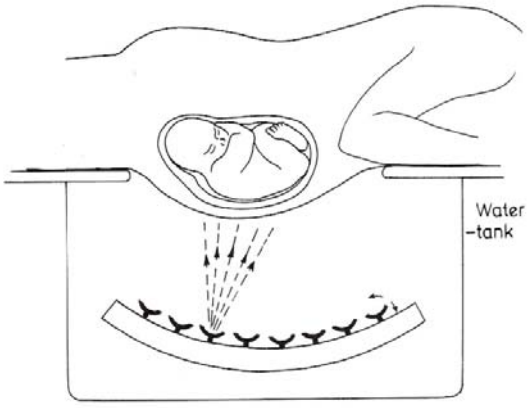


Fig. 5 Basic structure of the UI Octoson scanner.

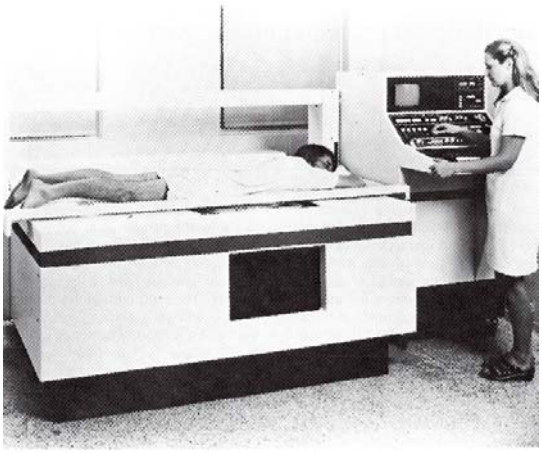


Fig. 6 The UI Octoson scanner (courtesy of Ausonics).

A large part of the success of the Octoson was due to the grey-scale images resulting from the electronic processing of the echo signals. This grey-scale processing preserved the weak echoes which had been removed as noise in most scanners. The preservation of weak echoes in an image demonstrated that they were generated by weakly scattering targets in the parenchyma of tissue. As a result of this preservation, the images looked like slices through tissue and did not just show boundaries (Figure 7).

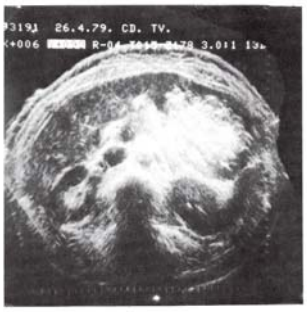


Fig. 7 Image from UI Octoson scanner (courtesy of Kossoff G).

These images improved the reputation of ultrasound imaging when X-ray CT and MRI imaging were making a significant impact. Compound scanning had been developed to accurately show distinct tissue boundaries and

anatomical boundaries such as the fetal head and the liver/kidney interface. Simple sweep grey-scale images demonstrated that it was often better to have a small amount of local image distortion rather than blurring due to misregistration in compound scanning. The importance of well-focused beams, both on transmission and reception, was illustrated by the clarity of small liquid-filled structures (e.g. cysts and blood vessels). Improved transducers and grey-scale signal processing handled boundary and parenchymal echoes well. In other words, full use was made of the dynamic range of the echo signals. The speckle pattern of the weak echoes gave some information on the nature of the scanned tissue but it is probably of more value for assessing tissue motion such as that of the myocardium. Complete images produced by single sweep scans showed that compound scanning is not essential. This had major implications for hand-held, real-time scanners which were starting to be developed.

Grey-scale images were initially recorded using photographic film which did not allow the build-up of the image to be observed as the echoes were received and was therefore inconvenient to use. The advent of grey-scale scan converters, with which the build-up of the echoes in the image could be observed, provided the technology that ultrasound imaging required. Today echo signals are stored in digital memory which permits imaging processing. For a few years bi-stable display cathode ray tubes had been employed since they presented the large structures clearly but they were essentially a blind alley.

IV. Real-time mechanical scanners

It became obvious in the early 1970's that if fleeting structures like heart valves could be presented in an image to the operator, interpretation of cardiac echoes would be greatly facilitated. The simplest approaches were to make a single-element transducer rotate (Figure 8), rock (Figure 9) or oscillate (Figure 10) with the transducer closely coupled to the skin surface. With a careful choice of casing plastic, oil and structural design, the transmitted pulse was not degraded compared to an M-mode or B-mode and hence image resolution was preserved. The image sector angle of a rocking transducer was usually kept below 90^0 to minimise vibrations. Rotating transducers could readily produce a large 180^0 field-of-view with a high line density at low frame rates, giving a quality of image similar to that of a B-scanner. This was a goal of real-time scanners since, understandably, there was a reluctance to accept lower quality images. Oscillating transducers, angular or linear, performed well up to about 30 frames/sec, above which vibrations became problematic. The rotating transducer approach was therefore preferred at higher frame rates. An example of a commercially produced transducer consisting of four transducers on a rotating wheel is presented in Figure 11. A linear oscillation with a rectangular field-of-view suited some applications and a number of elegant devices were produced.

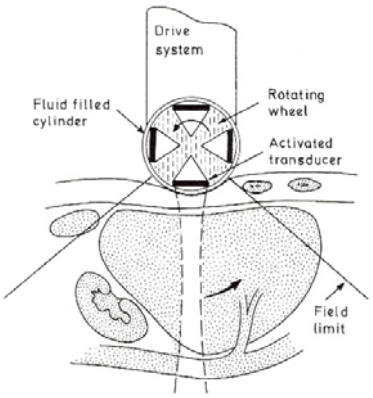


Fig. 8 A schematic diagram of a rotating transducer, real-time B-scanner.

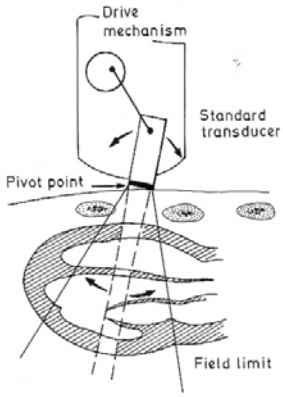


Fig. 9 A schematic diagram of an oscillating transducer, real-time B-scanner. The oscillations produce a sector scan.

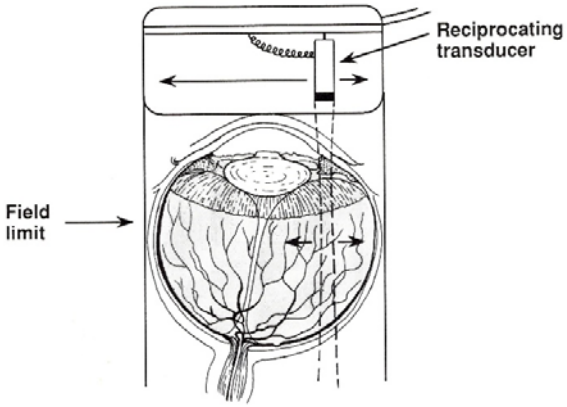


Fig. 10 A schematic diagram of an oscillating transducer, real-time B-scanner. The oscillations produce a linear scan.

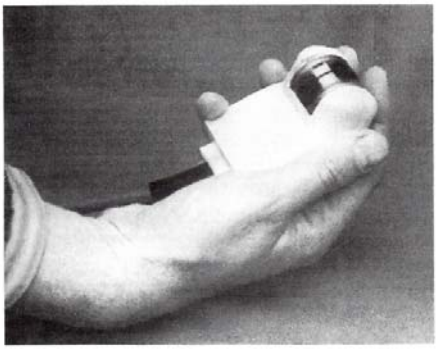


Fig. 11 A mechanical transducer assembly in which four transducers rotate in a thin-walled oil-bath (courtesy of BCF).

The advent of hand-held imaging transducers, which gave good image quality, made scanning quicker and easier. Imaging moving structures and changing scan-planes meant that the sensitivity controls (Gain, Time-gain-compensation – TGC, Power) were difficult to optimise with manual controls. Approaches to automatic control were introduced whereby information, related to echo signal magnitudes and rates of attenuation, was used to set the sensitivity in different regions of the image. After some initial concerns about loss of control, expressed by clinical users, it was demonstrated that well-balanced reliable images could be quickly produced.

There were attempts at mechanical 3D imaging at this time. However clinical operators with hand-held transducers developed skills in mentally imaging 3D anatomy as they viewed changing planes of scan. This

removed the immediate need for 3D real-time imaging. The latter awaited the demands of heart scanning where changing volumes are of interest and these demands were accommodated by 2D phased arrays.

The versatility of real-time ultrasound technology is amply exhibited by a range of special invasive devices which have been developed. Over the history of medical ultrasound both real-time mechanical and electronic array technologies have been used. This versatility continues to be exploited in Doppler duplex, tissue biopsy and contrast agent applications. The new equipment remains relatively inexpensive.

IV. Plan position indicator (PPI) imaging

In PPI, a transducer rotates and scans through 360^0 at right angles to the axis of a rod which is inserted into the body (Figure 12). Since the transducer can be placed close to the site of interest, high frequencies can be used. The transducer is contained in a shallow oil bath or balloon to avoid friction at the tissue. The transducers are often labelled 'transrectal', 'transvaginal', 'transuretheral' or 'transoesophageal'. Transoesophageal probes give good access to the heart by avoiding lung and bone. As for rotating element B-mode scanners, careful choice of oil and window plastic results in little distortion of the transmitted ultrasound pulse. Rotating transducers are also employed in flexible endoscopes.

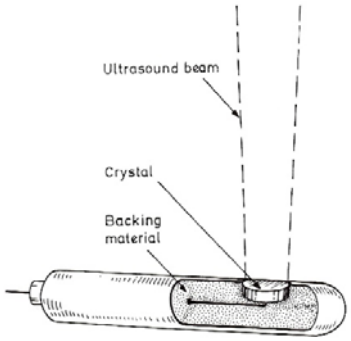


Fig. 12 Transducer for 360° radial invasive scanning.

V. Catheter scanners

Reducing the diameter size of PPI endoscopes from say 15 mm to 2 mm and increasing the frequency from 3 MHz to 30 MHz gave rotating element catheters a 360^0 field-of-view. Such catheters continue to be applied to the study of artery walls. Again liquid-filled balloons are used to make good acoustic coupling. This catheter technique is known as intravascular ultrasound (IVUS) [9]. Due to the risk of infection, catheters are required to be disposable and hence inexpensive. A concern using catheters which viewed only sideways and not forwards was that they may block a stenosed artery. Forward-viewing oscillating-transducer catheters were produced which demonstrated that easily understood images, pulse-echo and Doppler, could be produced of structures ahead of the catheter tip.

VI. Biopsy needle guidance

A big attraction of real-time ultrasonic imaging was that the tip of a fine needle could be observed and followed to the tissue site of interest. The needle may or may not be passed through a guide channel or attached to the transducer. Again sterility must be retained. This continues to be a very valuable technique since a collected tissue sample can then be passed to the pathology laboratory. This method of getting an accurate diagnosis has in many cases removed the need to try to interpret ultrasound signals from tissue.

It is possible to attach a small piezoelectric element to the tip of the needle and get a signal which identifies it accurately in the real-time image. To avoid expense, most operators work with a normal needle and rely on identifying the tip echo signal. However there can be ambiguity as to the exact source of the tip echo.

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AUTHOR BIOGRAPHY

Professor William Norman McDicken graduated from University of Glasgow with a BSc in physics in 1962 and a PhD in nuclear physics in 1965. He was an assistant lecturer for one year at University of Glasgow before becoming a medical physicist for the West Regional Hospital Board in Glasgow (1966-1974). During this time he organised and ran a new course on medical ultrasound. In 1974 he became a medical physicist in Lothian Regional Hospital Board, Edinburgh becoming the Professor of Medical Physics and Medical Engineering at University of Edinburgh in 1988. He held this role until 2005 whereupon he became Emeritus Professor of Medical Physics and Medical Engineering, University of Edinburgh, a post he still retains. Professor McDicken is a past president of British Medical Ultrasound Society and a winner of the Hospital Physicists' Association's Manufacturers' prize.

Professor Carmel Mary Moran BSc MSc PhD FInstP FIPEM graduated in Physics and Applied Maths from Queen's University Belfast in 1986, MSc from University of Aberdeen in 1988 and PhD from Institute of Cancer Research, University of London in 1991. From 1991, she has worked at the University of Edinburgh, formerly in Medical Physics and latterly in the Centre for Cardiovascular Science. Her research interests include the development and characterisation of ultrasonic contrast agents for diagnosis and therapy and the applications of high frequency ultrasound. She established, and currently directs, the preclinical ultrasound imaging facility at the University of Edinburgh. She is a member of the EFSUMB safety committee and is a Past-President of the British Medical Ultrasound Society.

THE DIASONOGRAPH STORY

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I. Background.

Up to the time when Ian Donald started to experiment with ultrasound in Glasgow there had already been many significant developments in diagnostic applications of ultrasound around the world. A number of researchers had managed to acquire industrial flaw detectors and publish the results of their A-mode investigations of tissues. In 1949, R P McLaughlin and G N Guastavino at the Argentinian laboratory of the American electronics company RCA, published a paper describing their own pulse echo instrument for detecting foreign objects in tissue, including the example of a stone embedded in an excised kidney [1]. In the same year, George Ludwig, a medical officer, and Francis Struthers, a physicist, both at the Naval Medical Research Institute in Bethseda, Maryland, measured the acoustic properties of a range of tissues and demonstrated gallstones implanted in dogs [2]. Also in 1949, John Wild, an English surgeon working at the University of Minnesota in the USA, measured changes in bowel wall thickness [3]. In 1953, cardiologist Inge Edler and physicist Hellmuth Hertz, in Lund, Sweden, experimented with an industrial flaw detector, borrowed from a shipyard, and interpreted moving echoes from within the heart. By 1954, they had invented the M-mode technique for recording and measuring echo movements and had published M-mode echo recordings of the hearts of living patients, establishing what was to become the diagnostic technique of echocardiography [4]. In 1956, G Mundt and W Hughes described their use of a flaw detector for A-mode examination of in vitro enucleated eyes and patients with intraocular tumours [5].

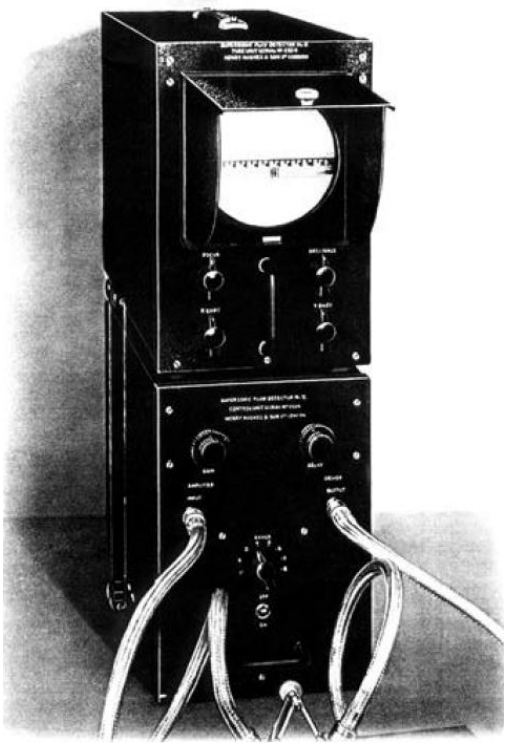

Fig. 1 The Kelvin and Hughes Mark IIb Flaw Detector.

In the late 1940s, Valentine Mayneord, Professor of Physics as Applied to Medicine, at the Royal Cancer (later Marsden) Hospital in London, picked up on Wild's early work and started to investigate the application of ultrasound A-scans to the brain, a very challenging organ to study with ultrasound due to attenuation, reflection

and refraction by the skull. R C Turner, an electronic engineer in Mayneord's department, using a Kelvin and Hughes Mark IIb industrial flaw detector (Figure 1), was sometimes able to show that an echo from the midline of the brain could be displaced towards or away from the probe if there was a space occupying lesion in either hemisphere [6]. Turner demonstrated his findings to Swedish neurosurgeon Lars Leksell during a visit by the latter to Mayneord's department [7][8]. Later, in 1950, Leksell borrowed a similar Kelvin and Hughes Mk IIb flaw detector but failed to improve much on Turner's patchy success [9]. However, when he replaced the Kelvin and Hughes instrument with the Siemens flaw detector that his cardiologist colleague Elder had used, he found

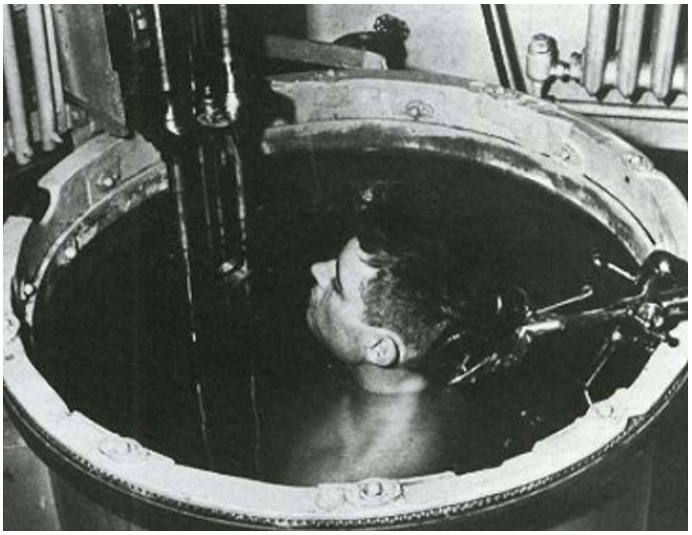

Fig. 2 Water bath compound scanning system of Howry et al., c 1957. This water-bath was based on the gun turret of a WW2 B-29 bomber. Photo courtesy of AIUM.

he could demonstrate shifts of the midline echo more clearly. His results, published in 1955 [10], led to the establishment of echoencephalography as a valuable diagnostic tool.

Cross-sectional (B-mode) images of human subjects were first achieved by two groups in the USA. One group was led by radiologist Douglass Howry, in Denver, USA, while the other consisted of Wild, mentioned above, and his electronic engineer colleague, John Reid, in Minneapolis. Between 1949 and 1954, Howry's group constructed a number of 'Somascope' instruments which required the subject to sit in a water bath, across or around which an ultrasound transducer was automatically scanned [11]. In the later versions, the direction of the ultrasound beam was changed between passes of the transducer, producing 'spatially compounded' images in

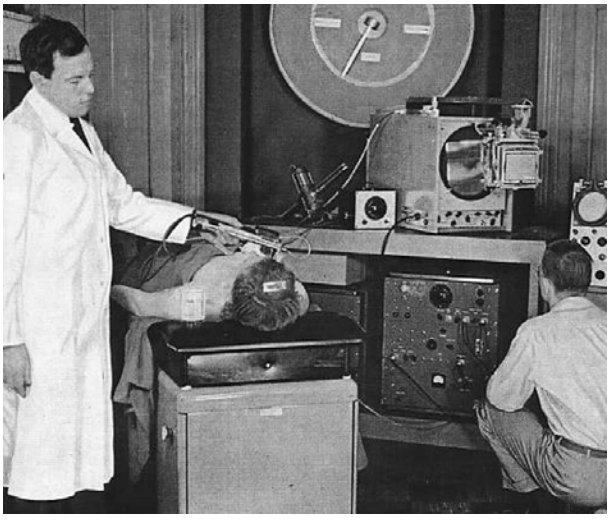

Fig. 3 John Wild (left) demonstrating the hand-held 15 MHz contact scanner developed by himself and John Reid (right) c 1953. From the front cover of Electronics magazine, March 1955.

which each anatomical target was insonated from several directions and the echoes summed [12] (Figure 2). This resulted in improved spatial resolution and better delineation of specularly reflecting tissue interfaces. The need for immersion was avoided by Wild and Reid who, in 1953, built the world's first hand-held ultrasound contact scanner, which they used successfully to produce grey-scale images of breast tumours [13]. This 'Two-dimensional Echoscope', as Wild called it, had a small water-filled chamber with a flexible rubber membrane forming its base, mounted beneath it (Figure 3). A 15 MHz transducer was driven back and forth within this chamber by an electric motor and worm screw drive.

II. Ian Donald.

Before his move to Glasgow, Ian Donald (Figure 4) held the post of Reader at the Institute of Obstetrics and Gynaecology at the Royal Postgraduate Medical School, Hammersmith Hospital, London. During his previous clinical career, he had demonstrated a talent and enthusiasm for innovation by developing improved mechanical devices for the diagnosis and treatment of respiratory problems in neonates. Donald had read Wild's 1951 paper in the Lancet [14] and was able to speak with him directly when Wild visited the Royal Postgraduate Medical School during an extended visit to London to deliver the University Lecture in Medicine. Wild showed Donald slides of some of his A-mode and cross-sectional (B-scan) ultrasound images. They also discussed [15] the possibility of using ultrasound to image the gravid uterus, Wild indicating that, in view of its cystic nature, a lower frequency would be more appropriate than the 15 MHz that he had been using. Wild had also visited Mayneord to hear of his work with the brain, so he knew of the Kelvin and Hughes flaw detector and suggested something similar might be suitable if Donald wanted to take further the possibility of obstetric applications. Donald was unable to attend Wild's lecture himself, but Mayneord, who Donald knew through his interest in his work, gave him an account of it [15]. Apart from describing his ultrasound breast imaging work in his lecture, Wild had included a discussion of how ultrasound might be applied to the lower abdomen. He had also commented on the safety of using ultrasound on live subjects, advising that a positive but cautious approach to safety was justified, given the absence of evidence of tissue damage at the ultrasound intensities used.

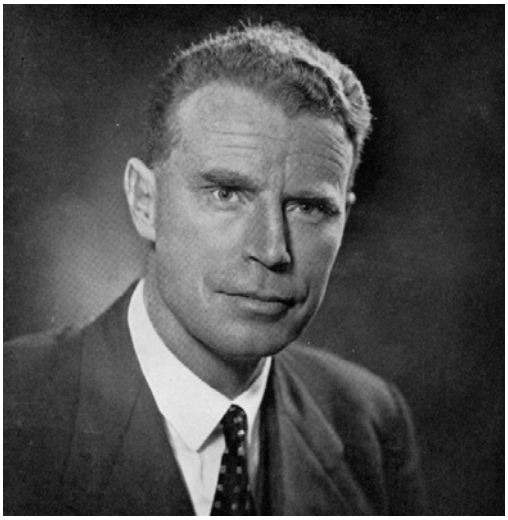

Fig. 4 Ian Donald. Photo courtesy of the BMUS Historical Collection.
Photograph held in NHS Greater Glasgow and Clyde Archives.

In September 1954, Ian Donald took up his post as the Regius Professor in Obstetrics and Gynaecology at the University of Glasgow. For a while he collaborated with John Lenihan, of the Western Hospital Board's Regional Department of Medical Physics, and in particular with physicist and engineer Ronald Greer, continuing his efforts to develop novel respiratory equipment and to investigate the respiratory changes during the first breath of the newborn. However, it proved difficult to make further substantial progress into the problem of neonatal respiratory distress [16] and Donald realized a change to a new project might be timely. He was aware that differentiation between ovarian cysts and fibroids was an important clinical problem, with potentially life-threatening consequences for the patient, so he decided to experiment with ultrasound to see if it could help.

One of the first things he did was to borrow a powerful ultrasound generator "from a friend of a friend in a scientific instrument factory near Paisley" [17] and noted that red cell destruction depended on exposure time. He concluded that the degree of cell damage was directly proportional to the heat generated [18] and considered that the use of ultrasound would be safe as long as no significant heating occurred. In the spring of 1955, through an introduction by a grateful patient, he was invited to the Renfrew factory of the boiler making company, Babcock and Wilcox, where he was given a demonstration of a flaw detector, made by Kelvin Hughes Ltd [18]. Although Donald was probably not aware of the significance, this was their latest, much improved, Mark IV model with a single hand-held probe containing two piezoelectric transducers, one for pulse transmission and one for echo reception; these were arranged in a shallow 'V' configuration so the crossover region of their beams extended several centimetres from the probe [19]. He noticed that the technicians used their thumbs several times a day to check that the instrument was working satisfactorily, reinforcing his opinion that there was no significant hazard from ultrasound exposure. A technician also demonstrated that the echo from the bone could be identified and that its position along the time-base trace on the A-scan display shifted back and forth as the probe was pressed in and out against the thumb.

In July 1955, Donald arranged a second visit to Babcock and Wilcox, this time using the flaw detector himself to examine uterine fibroids and a large ovarian cyst, freshly removed from patients that morning. Donald reported that the results from the flaw detector were as he had expected from his reading of the published literature [18]. This gave him encouragement, although Fleming and Nicholson have since argued that his interpretation of the echo patterns may have involved a degree of wishful thinking and that he may have been lucky to find the controls already set appropriately [20]. Shortly afterwards, Donald visited Prof. Mayneord at the Marsden Hospital. He found the team somewhat discouraged by the difficulties they had encountered, to the extent that they had decided to replace their Kelvin and Hughes flaw detector with an A-scan machine they were building themselves. They were, therefore, in a position to offer their Kelvin and Hughes Mk IIb flaw detector as a loan to Donald [21].

On his return to the Western Infirmary, Donald enlisted the help of Greer and, together, they tried to reproduce the results that Donald had achieved during his second visit to Babcock and Wilcox and to move on to investigating the intact abdomen. Unfortunately, they had little success, largely because the Mk IIb flaw detector they were using was inferior to the later, Mk IV, model that Donald had used at Babcock and Wilcox. Its performance had been further compromised by a modification made by Turner, while working on the midline shift project in Mayneord's Department. In its original form, the Mk IIb machine had separate transmit and receive probes in order to prevent the large excitation voltage applied to the transmit transducer from temporarily overloading and paralyzing the receiving amplifier. Unfortunately, it proved extremely difficult to hold these two probes close together on the curved skull of the patient. By replacing the two probes with a single probe for both transmission and reception, Turner had solved this ergonomic problem but in so doing he had reintroduced the paralysis problem, making it impossible to detect echoes from within 8 cm of the probe face. Donald and Greer tried introducing water offsets between the probe and the patient in order to overcome this serious limitation, both in the form of open-ended tubes with a rubber membrane at the patient end [22] and in the form of water-filled sealed condoms[18], but these did not prove to be suitable as a long term solution for clinical use. Despite further help when obstetrician John MacVicar (seen on the right in Figure 12), then a registrar in the Department of Midwifery, joined the team sometime in 1956, they still could not obtain consistent echo patterns from within the abdomen, nor reliably interpret them.

III. Tom Brown and Kelvin and Hughes Ltd.

This rather unsatisfactory state of affairs continued until Tom Brown (Figure 5), a twenty-three-year-old engineer with Kelvin and Hughes Ltd, at Hillington, Glasgow, heard that a professor was using one of the company's flaw detectors to examine patients. Brown had previously impressed his employers by his work in helping to develop a semi-automatic flaw detecting system to the extent that they had sponsored him for a course in applied physics at the Royal College for Science and Technology in Glasgow (now Strathclyde University). Unfortunately, the mathematical content of the course had proven too challenging for Brown and he had to drop out after one year [23]. As he said himself: "I spent too much time playing snooker and generally enjoying the student lifestyle" [24}. His employers took him back, but he was now looking for a way to redeem himself in their eyes. The idea of applying ultrasound to medical diagnosis appealed to him so, one evening in late 1956, he telephoned Donald; his boldness was rewarded by an invitation to visit Donald at the Western Infirmary. Despite what he called a "rather comical demonstration with the water stand-off and all the rest of it" [24], Brown could see there were echoes coming back from within the patient's body, so he called his boss, Alex Rankin, Head of

Applications Engineering, to tell him of the potential new application. Rankin was already well disposed towards medical projects involving ultrasound, having provided support to Leksell in Sweden. He immediately arranged for a brand-new Mk IV flaw detector to be delivered to Donald [23].

Fig. 5 Tom Brown, pictured around the time he built the bed-table scanner. Photo courtesy of the BMUS Historical Collection.

The new instrument made all the difference to the success of Donald and MacVicar's A-scanning efforts (Figure 6). Probes were provided at ⅝ MHz, 1.25 MHz, 2.5 MHz and 5 MHz, but they soon established that a frequency of 2.5 MHz gave the best compromise between penetration and spatial resolution for obstetric patients. In addition, Brown's company was able to provide them with a Cossor oscilloscope camera to record the A-scans on 35 mm film [23]. Hitherto, their only means of recording A-scan traces had been by sketching them. Donald was very pleased, as the ability to produce accurate photographic records of the traces was important for publication of any noteworthy results.

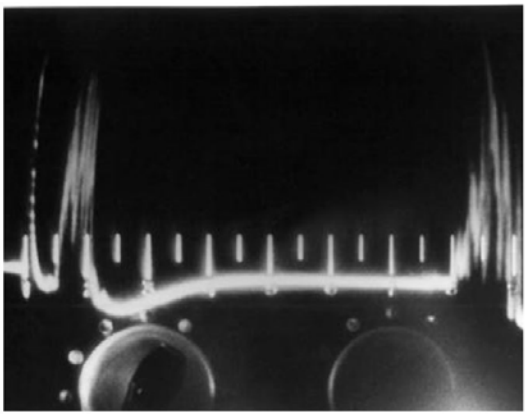

Fig. 6 A-scan of a simple ovarian cyst, c 1956. Photo courtesy of Tansey and Christie [24].

Donald also appreciated the involvement of an engineer with substantial practical experience of the most recent developments in industrial ultrasound technology. As a result, Brown was welcomed into the fold whilst Greer's involvement came to an end [23]. At first, Brown's involvement was purely informal, with him going to the Western Infirmary in the evenings, after his day's work at Kevin and Hughes at Hillington. There, he would perform any necessary maintenance and help MacVicar develop and analyse film from the day's clinical work. In order to learn how to interpret the echo traces, they would experiment by applying the probe to their own

bodies, as well as to rubber membrane acoustic windows built into the walls of water tanks containing surgically removed cysts and tumours [25]. Donald, however, had little time for such in-vitro experimentation, preferring that the probe be applied directly to real patients [26].

Donald was happy to arrange an abundant supply of patients and he drove the project forward with characteristic energy. By mid-1957, with Brown's assistance, he and MacVicar were providing a clinically useful service to gynaecological patients and starting to look at obstetric patients. Through their hard-won and growing expertise in interpreting the A-scans, they had largely solved the original challenge of differentiating cysts from solid tumours and they could even differentiate between different types of cysts. There had been occasional mistakes, as when a highly vascular fibroid had been mistaken for a cyst [27], but this all contributed to the learning process; consequently, the initial scepticism amongst Donald's clinical colleagues at the Western Infirmary was disappearing. The much-quoted turning point in the acceptance of the technique was when a swollen abdomen, thought by the Regius Professor of Medicine to be due to ascites, secondary to inoperable and terminal cancer, was being ultrasonically examined by Donald. Looking over Donald's shoulder, MacVicar, commented that the trace looked like that of an ovarian cyst. Although MacVicar was rewarded by an unseen kick from Donald for contradicting the Professor of Medicine's diagnosis, Donald accepted that an ovarian cyst was a possibility and arranged for a laparotomy, which confirmed that the mass was indeed a very large ovarian cyst. This was duly removed and, instead of being allowed to die, the patient made a full recovery [24][28].

A. The Bed-table Scanner

Notwithstanding Donald and MacVicar's satisfaction and excitement with the clinical value of A-scanning using the Mk IV flaw detector, Brown was convinced that a display that showed the positions of each reflecting interface was needed to fully exploit the potential of the ultrasound pulse-echo technique. He felt that, ideally, since the human body was three dimensional there should be a 3D representation of the echo sources. This was to remain his ambition throughout his life [23] [24] but, for now, he accepted that an image of a 2D cross-sectional slice of the patient would have to suffice. According to Brown [23], Donald was initially less enthusiastic about what 2D imaging could offer. This was despite the interest Donald had shown in radar and sonar during his wartime military service and the 2D ultrasound images he had seen in the publications of Howry's team in Denver and of Wild and Reid in Minnesota. He had even attempted, unsuccessfully, to build a 2D ultrasound system himself in his early days in Glasgow, although very little is known about it [22]. He felt the detailed echo information from within organs and other body masses that he was now obtaining had more diagnostic value than knowing the position of a reflecting interface [29]. For whatever reason, Donald chose not to show Howry's or Wild's cross-sectional images to Brown. Years later, Brown said "I think that had we been aware of what Howry was doing and had set out our stall to improve on Howry's work, we would have been stuck with immersion scanning" [24]. Having seen the trouble caused by attempts to use water offsets when he first visited Donald, Brown was very keen to avoid using water, either for partial immersion of, or acoustic coupling to, the patient in a hospital environment. He also wanted a system that could be used at the patient's bedside, as was possible with the A-scan unit. It is a matter of conjecture what the outcome would have been if Brown had known of Wild and Reid's relatively compact hand-held contact scanning system: perhaps he might have been inspired to try to build a low frequency development of it.

From his knowledge of the radar work undertaken by Kelvin and Hughes, Brown was familiar with "True Motion" radar displays, in which the screen acted as a fixed map on which the position of the transmitting aircraft or ship, as well as the positions of echo-returning targets, were updated after every sweep of the beam. He considered applying the technology of this technique to the medical situation [23] but radar experts at Kelvin and Hughes quickly made him realize this approach would be unnecessarily complex. Whereas, in the case of radar, the position of the transmitting ship or aeroplane had to be calculated from the echo data returning from land-based targets, in the medical situation the position of the probe could be measured directly by mounting it on a support arm or mechanism from a fixed point. From the known transducer position, the positions of echo-producing targets could be plotted on the screen of a cathode ray tube (CRT) using the standard 'plan position indicator (PPI)' method of radar displays. Brown's proposal was supported by his managers at Kelvin and Hughes, including the company's Chief Scientist, Bill Halliday. However, the crucial move was made by Donald, who was keen to put Brown's input on a formal footing [23]. Donald arranged to meet William Slater, Deputy Managing Director of Kelvin and Hughes, impressing him with his account of the clinical value of the project and its potential. The result was that half a day per week of Brown's time was allocated to working with Donald, along with a budget of £500 to make the first machine. Brown later said this figure turned out to be very elastic [23].

During 1957, Brown designed and built his prototype system [30] [23] in the research department of the Kelvin and Hughes factory at Hillington. He felt that a system in which the probe was in direct contact with the patient was required, in the same way that the probe of a flaw detector was applied directly to the test piece in an industrial setting. Mindful of the convenience of being able to use the system on a patient in a hospital ward, he chose to build his prototype around a wheeled bed table. A photograph of the resulting 'Bed-table scanner' being used by MacVicar to scan Brown's abdomen is shown in Figure 7.

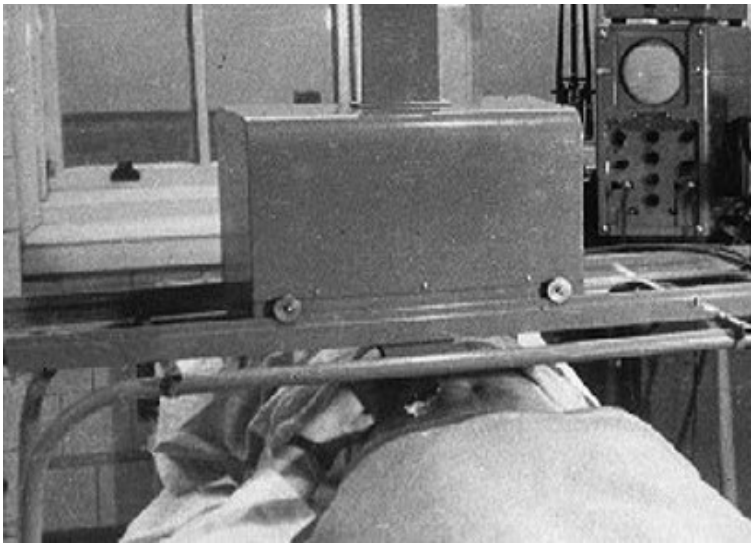

Fig. 7 Tom Brown's bed-table scanner, c 1957. Note the restricted room for the operator's forearm and hand (seen lower left). Photo courtesy of the BMUS Historical Collection

The user could move the transducer by hand anywhere within a fixed vertical transverse 'scan plane', keeping the transducer face lightly pressed against the patient's skin, which was kept lubricated with vegetable oil. The scan plane was defined by the position of a wheeled carriage that could be moved transversely on rails across the bed table. At any position within this plane, the transducer could be rotated or rocked through a large range of angles, thus giving the benefit of compound scanning, mentioned previously in connection with Howry's system. The scan plane could be moved longitudinally by simply moving the bed table above and along the bed on which the patient was lying. Three displays were provided: an A-mode display, at that time still considered essential by Donald and MacVicar: a B-mode display on a long persistence CRT screen, which the user could monitor as he moved the probe around in an exploratory fashion; a second B-mode display, this time on a short persistence CRT fitted with a camera for when the user had found a particular cross-sectional view that he wished to record. Figure 8 shows two of these displays: on the left is a Mark IV flaw detector used to display A-scans and, on the right, is a CRT with a Thompson-Polaroid Land camera attached for recording B-scans.

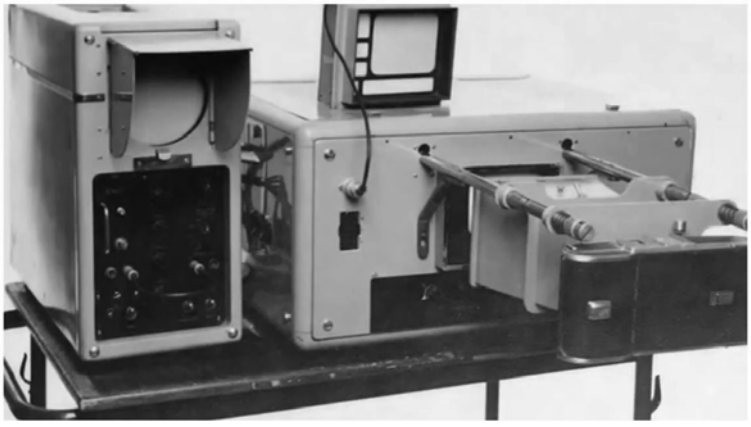

Fig. 8 Displays and camera from the bed-table scanner. Photo courtesy of the BMUS Historical Collection.

The hand-guided probe could be rotated about a horizontal spindle whose X and Y coordinates were measured by means of a system of wires and pulleys connected to the shafts of wire-wound X and Y linear potentiometers. The angle of the probe, and hence the ultrasound beam, to the Y axis of the measuring frame was measured by linking the rotation of the probe to the shaft of a sine-cosine potentiometer (Figure 9). By applying stable positive and negative voltages of equal magnitude (+v and -v) respectively to the diametrically opposite sides of the potentiometer's circular track, the resistance per unit length of which varied sinusoidally, the two wipers, arranged 90° apart on the potentiometer shaft, picked off voltages of v cosθ and v sinθ respectively, where θ was the angle of the probe to the Y axis of the measuring frame. Each of these voltages was applied to its own integrator circuit, producing two voltage ramps with slopes proportional to v cosθ and v sinθ respectively. A voltage proportional to the Y coordinate of the probe spindle was added to the v cosθ ramp and a voltage proportional to the X coordinate of the spindle was added to the v sinθ ramp. This provided vertical and horizontal voltage drives, respectively, for the time-base on the CRT used for the display. When the time-base had swept across the screen to reach a point corresponding to the probe face (with a correction to allow for the perspex block in front of the transducer) the transmitter was triggered. From this moment on, the position and orientation of the time-base trace on the CRT screen matched the position and orientation of the transmit-receive ultrasound beam within the scan plane. The speed at which the time-base spot of light was swept across the screen depended on the magnification required.

For unity magnification, for example, echoes from two targets a certain distance apart, both lying on the axis

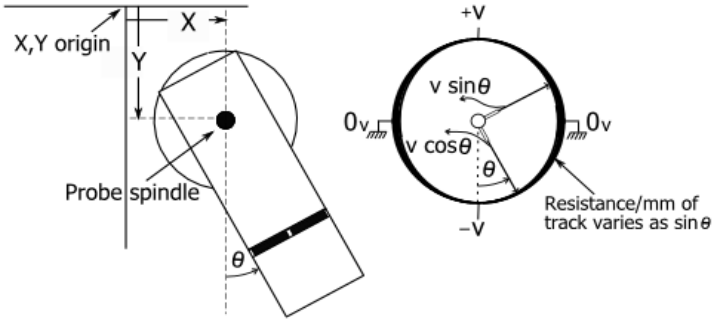

Fig. 9 Left: Measurement of probe angle Θ relative to the Y axis of the scan plane. Right: The 'cosine' wiper of the sine-cosine potentiometer is at the same angle as the principal axis of the probe.

of the beam, must be represented by echoes the same distance apart along the CRT time-base. In order to allow for the two-way travel of the sound pulse in the body, this requires the spot of light to be swept along the time-base trace at half the speed of sound in the body (approximately 1540 ms^{-1}). For a magnification of one half, the spot of light would have to be swept across the screen at one quarter of the speed of sound, and so on. This arrangement was similar to the measuring system used later in the Diasonograph machines. A then popular construction kit known as 'Meccano' was used to provide many of the smaller components, such as chains, sprockets and pulley wheels, but some of the larger and more critical components were manufactured by technicians in the model shop at the Hillington factory.

The probe was that of the Mark IV flaw detector. It housed two rectangular, 10 x 7 mm, unfocussed, air-backed barium titanate transducers, one for transmission and one for reception. They were mounted, side by side with their shorter sides adjacent, on one of the flat faces of a one inch (25.4 mm) thick cylindrical Perspex block, the opposite flat face being in contact with the patient [30]. As mentioned earlier, the use of separate transducers for transmission and reception avoided the problem of paralysis encountered in the modified Mk II flaw detector initially used by Donald. The transducer dimensions were chosen to give a compromise between beam width in the near field and angular divergence in the far field. Another consequence of using two transducer elements side by side was that the overlap of their beams increased with distance from the probe face, producing a two-way sensitivity that increased progressively with depth. This provided a degree of compensation for the effect of attenuation on echo strength from deeper targets and explained why swept gain (later often called TGC - time gain compensation) was not considered a high priority. A resonant frequency of 2.5 MHz was chosen for the transducers, corresponding to a transducer thickness of approximately 1 mm, as the earlier in-vitro experiments made by Brown and MacVicar had indicated that this frequency would give an optimum compromise between spatial resolution and penetration for use in the lower abdomen. The transmission pulse generator and receiver were those of the Mark IV flaw detector [31]. The transmission transducer was shock excited by a pulse of approximately 2 μs duration, generated by charging a 100 pF capacitor to about 1.4 kV through a high resistance

and then triggering a thyratron to discharge it though the primary winding of a pulse transformer, across the secondary winding of which was connected the transducer in parallel with a 50 ohm damping resistor [30].

The other electronic circuits and components were also largely those used in the Mark IV instrument. Brown chose a pulse repetition frequency (PRF) of 50 Hz, partly because this was one of the standard PRFs of the Mark IV and partly because it represented what he considered to be a prudent compromise between safety and performance. A higher PRF would have allowed the probe to be scanned across the abdomen more quickly, without increasing the gaps between lines of echoes on the display, but it would also have meant that the patient would have received more ultrasound energy. On testing an early version of the apparatus by scanning a grapefruit suspended on a wire in a water tank, Brown realized the signal processor stage, the output of which was used to brightness modulate the time-base trace on a CRT, needed to be more sensitive. He described this circuit as being "where art and science tended to co-exist" [23], but he nevertheless managed to achieve a very acceptable dynamic range of the order of 60 dB.

The increased receiver sensitivity allowed him to introduce another safety feature in the form of a switched attenuator in the transmit voltage drive to the transducer. If the displayed echoes were too strong, the operator could reduce their amplitude by means of the attenuator, leaving the receiver sensitivity (gain) at its high level. He recommended that the operator should start with a high transmitter attenuation setting and, only if necessary, reduce it to the level needed to achieve a useful image, thus helping to ensure that the patient received no more ultrasound power or intensity than was necessary. Brown's estimate of the maximum acoustic intensity that his scanner was capable of producing in the patient was 1.5 mW cm^{-2} and the maximum acoustic power was approximately 1.0 mW [30]. These values are tiny in comparison to the corresponding figures for obstetric ultrasound scanners post-1990 [32] and were considerably less than those of either Howry [33] or Wild [34].

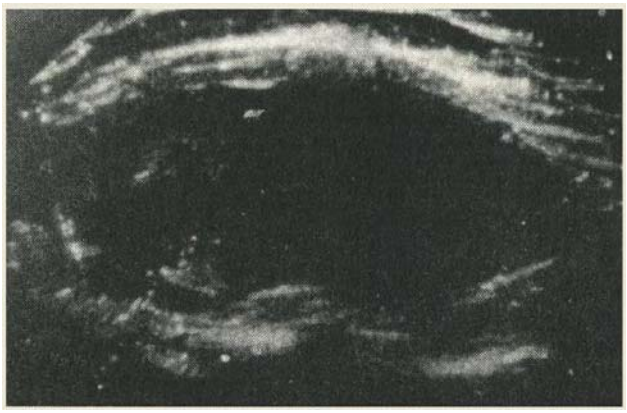

Fig. 10 Example of a scan produced by the bed table scanner. It shows a uterus containing a fetus (left) and a fibroid (right). Reproduced from Figure 17 of Donald, MacVicar and Brown, The Lancet, 1958 [30].

In late 1957, the machine was sufficiently developed to be put into use in the Western Infirmary, but Brown admitted to some disappointment at the quality of the images [35]. Figure 10 shows the scan of a uterus, obtained with the scanner, with barely recognizable echoes from an early stage fetus on the left and what was thought to be a fibroid on the right [30]. Brown and MacVicar frequently scanned each other to discover the limitations of the apparatus, how to get the best out of it, and which aspects of it required improvement. In observing clinicians using the system on patients, Brown recognized that the scanning technique varied considerably between operators and this had a large effect on image quality. Operator skill was not helped by the poor ergonomics of the system, which, Brown later admitted, had not been given much consideration during the design; in fact he described it as "ergonomically horrific", but added "it was all done, after all, on a £500 budget" [24]. For example, the user had to reach into the narrow space between the table and the patient (Figure 7) to manipulate the probe whilst turning their head away to see the display screens. Brown was able to improve results by making further modifications, including the updating of the amplifier with one from his company's new Mark V flaw detector which gave superior performance [31]. In order to overcome the restriction of being able to scan only in transverse vertical planes he later replaced the bed-table with another over-bed, structure, in the form of a wheeled steel framework. This allowed the operator to scan in planes perpendicular or parallel to the longitudinal axis of the patient's body, and in planes inclined to the vertical, as desired [36]. Another later improvement was the replacement of the Cossor oscilloscope camera with a Thompson-Polaroid Land camera. This allowed photographic records of scans to be viewed in the scan room within minutes, rather than having to

wait for a full roll of film to be exposed and then waiting a further period of hours for it to taken away, processed and returned. A potential disadvantage of Polaroid film was that it was less sensitive than conventional film. Initially, this resulted in the loss of weaker echoes but, after carefully reading Polaroid's technical literature, Brown solved the problem by designing an illumination box in which the Polaroid film could be briefly pre-exposed, increasing its sensitivity. These improvements all helped to foster marked and growing enthusiasm from Donald for the B-mode technique. By the time of submitting their June 1958 publication 'Investigation of abdominal masses by pulsed ultrasound' in The Lancet, Donald and MacVicar, with Brown's technical input, had used the bed-table scanner on 100 patients and had made 275 B-scan recordings [30].

In 1959, physicist Tom Duggan joined Donald's team to work on ultrasound, his salary being paid from a Scottish Hospital Endowment Research Trust grant that was intended to finance neonatal respiratory studies [31]. Between 1961 and 1962 he developed a fetal ultrasonic cephalometer, by means of which two bright 'pips' could be superimposed on the A-mode trace on a Kelvin and Hughes flaw detector [37]. The instrument was described as portable but it weighed 30 kg and had to be pushed about on a trolley [31]. The bright pips were placed at the leading edges of the two echoes from opposite sides of the fetal skull at the level of the parietal eminences. These echoes corresponded to the outside of the nearer side of the skull and the inside of the far side. The time elapsed between the generation of these two pips was measured electronically and converted to an estimate of the distance across the outside of the fetal head at this level, called the bi-parietal diameter (BPD). This was achieved by multiplying by a factor of 0.080 cm μs^{-1}, derived experimentally from measurements on neonates and post-mortem fetuses [38]. Because this conversion factor relates a distance to a (two-way) ultrasound time of flight, it is usually expressed by saying that the 'caliper velocity' is 1600 m s^{-1}.

The ultrasonic estimate of the BPD became an important index for monitoring fetal gestation and development, thanks to the efforts of John Willocks, a young doctor who had joined Donald's team about the same time as Duggan [37]. Later, Duggan joined Kelvin and Hughes, where he was involved with transducer developments, before moving on to an academic post at the University of Strathclyde and thence to the Regional Department of Medical Physics (now Clinical Physics and Bioengineering) at the West of Scotland Health Board. There, he was closely involved with the introduction of ultrasound teaching and development laboratories and with supervision of an ultrasound maintenance service [24].

Meanwhile, both Brown [23] and Donald [39] were frustrated by the variations between operators in probe manipulation, artefact avoidance and other scanning skills, as these were limiting the success of their cross-sectional imaging project. In Scotland in the mid-1950s, it was unthinkable that a young male engineer without any medical qualifications could be allowed to scan patients himself, particularly on gynaecology and obstetrics wards, so Brown was unable to demonstrate to others how to get more consistent results [23]. Mindful of his prior success in helping to develop an automatic industrial flaw detecting system, it seemed to Brown that an automatic clinical scanning system could provide the solution to the inconsistency problem as it would greatly reduce the influence of the operator on the scanning procedure. Even if it proved too complex and costly to consider as a prototype production machine, it would at least demonstrate the scanning action needed to produce good images. Donald agreed and asked Kelvin and Hughes if they could provide an "apparatus which automatically scanned the surface of the abdomen at a standard rate and rocking speed" [39].

B. The Automatic Scanner

Brown was aware that Kelvin and Hughes were investigating new probe designs in which the two rectangular transducer elements of the Mark IV design were replaced by a single disc-shaped transducer element, offering a much greater sensitivity than had been possible with the overlapping beam arrangement of the twin element probe. Also, by this time in the late 1950s, new, more sensitive, piezoelectrics such as lead zirconate titanate (PZT) were becoming available. Brown was keen to take advantage of these development as he was aware that attenuation, and hence signal loss, was very much more of a problem in tissues than in the metal structures for which flaw detecting technology had been developed. Not only would a higher sensitivity improve the dynamic range of the detected echo signals, but it would have safety implications as it would mean that pulses of lower energy could be transmitted. The large impedance mismatch between the transducer and the patient's tissue meant that the absorbing backing behind the transducer was more critical in suppressing 'ringing' than it was for industrial applications. This backing was normally made from epoxy resin in which dense metal particles were suspended but its performance could be degraded by gas bubbles trapped within it during the curing process. Brown and Clive Ross, a colleague at Kelvin and Hughes whom Brown described as "very gifted" [23], used a centrifuge to drive gas bubbles in the uncured epoxy resin away from the transducer as well as to give the backing an inclined rear surface. In Brown's own words, this allowed them to produce "some quite respectable single transducer probes" [40]. Brown was by now aware of the work of Howry's team, including their use of

concave lenses to improve beam shape, so Ross experimented with different lens designs, finally settling on a conical design, having decided it gave better results than lenses with the more conventional spherical curvature. The new scanner was provided with a range of focused transducers, with the frequencies that were standard for Kelvin and Hughes flaw detectors, namely 1.5 MHz, 2.5 MHz and 5.0 MHz. Brown later said that perhaps a frequency of 3.5 MHz would have been optimum in terms of the compromise between spatial resolution and penetration for gynaecological and obstetric applications but that this omission was not too serious in view of the other limitations of the equipment at the time [40].

The electronics of the automatic machine were mostly identical to those of the updated bed-table scanner [40]. An important exception was that the receiver amplifier from the Mark IV flaw-detector could no longer be used as, following the change to single transducer element probes, the transmission pulse produced too much receiver paralysis. When developing the new Mark V flaw detector for Kelvin and Hughes, one of Brown's colleagues, John Woods, had found that the solution to this problem was to design a separate tuned RF amplifier for each probe frequency. Consequently, separate plug-in RF amplifiers were provided for use with each probe of the automatic scanner. A feature of the new amplifiers was that the amplifier gain could be made to increase with time after transmission at a rate set by the operator. Without this feature, attenuation due to scattering and absorption in tissue would mean that B-mode echoes produced by a single transducer element probe would exhibit a general trend to diminish in brightness with depth. The new feature, known as 'swept gain', meant that this attenuation could be compensated for in a controlled way by the operator. Previously, in the bed-table scanner, using probes containing two transducer elements, a fixed degree of such compensation had been provided by the increasing beam overlap with depth. Now, by judiciously adjusting the swept gain controls, the general brightness of echoes could now be made more uniform at all depths. (Note that swept gain did not affect the difference in brightness between a strong echo and a weak echo at similar depths).

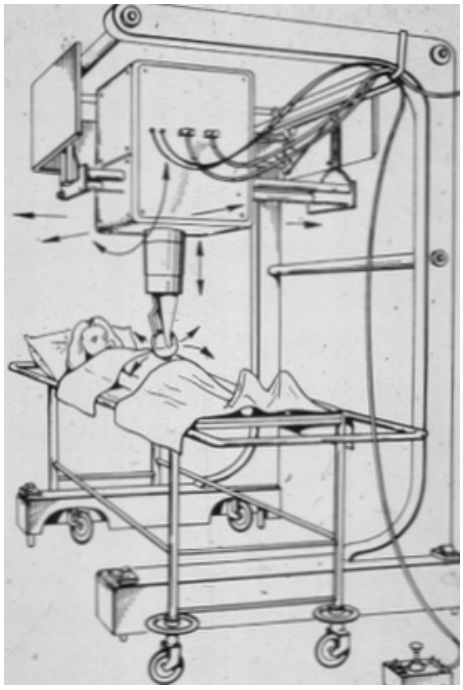

Fig. 11. Drawing of the automatic scanner from Brown [23], showing all the motorized movements. Courtesy of the BMUS Historical Collection

A diagram of the automatic scanner is shown in Figure 11. The finished machine, being used by Donald and MacVicar; is shown in Figure 12; a clearer view of the silver ball probe is inset. The ingenious way in which the probe was scanned and rocked automatically across the curved form of the patient is, perhaps, best described in Brown's own words [23]:

> "The 'business end' of the automatic scanner consisted of a probe holder in a 'silver ball' - which looked a little like the kind of soap dispenser once commonplace in public toilets. This was mounted on a vertical, telescoping motor-driven column, such that it moved up and down to keep

the probe face in contact with the patient's skin. It too was shiny chromium plate. A pressure sensing switch ensured that it maintained contact with the skin, but with minimal pressure.

The silver ball rocked to-and-fro on an axle, driven by a system of cranks and connecting rods. Slightly indelicate looking soft plastic "nipples" on either side of the ball touched the skin when the probe axis had moved about 30 degrees to the perpendicular to it, and - almost as sensitive as the real thing - caused the rotary motion to stop and then reverse its direction.

Each time the nipple touched the patient's skin, another set of relays and motors were activated to inch the vertical column sideways. This, when combined with the compensating up/down motion to keep the probe in contact with the skin, caused about a 15 mm tangential displacement between successive "sweeps" of the rotating probe.

In this way the probe gradually 'walked' across the abdominal surface, rocking to-and-fro as it went, carrying out what was actually quite a thorough, and highly consistent compound scan.

Of course, it was not quite as simple as that. To enable it to work properly on the steep flanks of the often rotund ladies being examined, an automatic changeover mechanism operated at about the 45 degree point on either side of the vertical, so that the horizontal drive then controlled the pressure, and the vertical drive did the 'inching'. The distance the machine 'inched' each time was regulated by a profiled cam system, so that it remained constant, irrespective of the average angle of the probe to the vertical. Nowadays it would all be done by microprocessor, but then it had to depend on cams, switches and relays - and I guess it was the sort of thing which would have delighted Heath Robinson.

There was a 'joystick' controller in a box on the end of a cable, by means of which the operator could position the probe for the start of the scan, but when he had done so, his task was over. He simply pressed the 'Auto' button, and the machine did the rest.

When it had finished, (about 90 seconds later on a big lady), it would switch itself off, and then ring a bell to summon him back. It may seem unlikely, but such was the confidence which developed in the machine, and the pressures of nicotine addition, that the bell became a necessity."

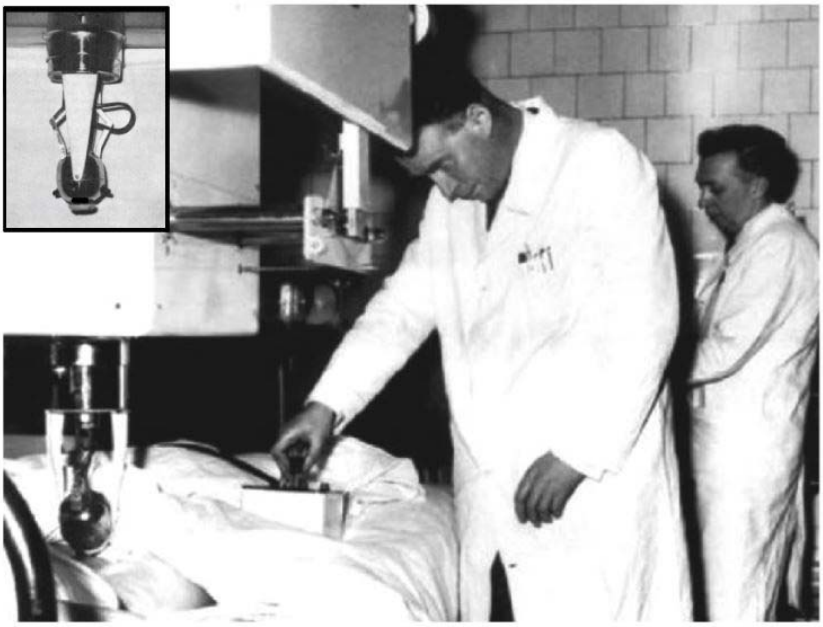

Fig. 12 The automatic scanner being operated by Donald (left) and MacVicar (right). The ball probe, viewed from another angle, is inset. Photo courtesy of the BMUS Historical Collection.

The large size of the somewhat intimidating box suspended above the patient was, in part, due to building-in the capability to scan transverse, longitudinal and all intermediate planes, as well in planes tilted away from the vertical. The tilt facility was little used, however, as users were not practised in the interpretation of the anatomy in such oblique views. The other reason for the bulk and complexity of the system was the incorporation of an optional facility for automatically stepping the scan plane in small increments perpendicular to itself in order to

acquire a volume set of echo data. This reflected Brown's enduring ambition to produce a scanner incorporating both 3D echo acquisition and 3D display. In later years he was to work with Sonicaid Ltd, based in Bognor, UK, to produce the radical "3D Multiplanar Scanner", which produced three-dimensional stereoscopic virtual images of body tissue but met with little commercial success [41]. Conscious of the disastrous consequences if the heavy box ever were to fall onto a patient, Brown incorporated safety features such as ratchets and cams that would prevent it falling more than a few millimetres should its support chain break. As might be expected with such a complex mechanical system, malfunctions did sometimes occur. The most alarming being on one occasion while scanning a "very stout" patient. The silver ball "started to dig in because the soft flabby fat stopped progress across the abdomen and the probe oscillated on one spot, burrowing deeper into the six or eight inches of fat" [24]. This was attributed to the patient, understandably, drawing away from the advancing ball, thereby inviting it to advance further. A part may also have been played by congealed olive oil around the probe, inhibiting it from moving up and activating the pressure sensing switch [42]. A white nylon ring was fitted later around the ball, increasing its contact area and thus reducing the pressure experienced by the patient. Another problem was electrical 'snowstorm' interference over the screen, caused by sparking across the contacts in the DC electric motors. When this happened, somebody would be sent from Kelvin and Hughes to clean the contacts and to fit suppressors and electrical shielding, as necessary.

While work on the automatic scanner was proceeding, a financial crisis had to be overcome. The £500 budget Brown had been given by Kelvin and Hughes had already been spent many times over. In December 1959, without any immediate prospect of commercial sales, and in view of Donald's estimate that it could take a further 15 years for diagnostic ultrasound to become routine, Brown was told by his manager at Kelvin and Hughes that the company could not justify further financial backing for the project, although they had assured Donald that they would complete the automatic scanner. Donald promptly sought the support of the University of Glasgow and was immediately promised £750 towards the research project. Moreover, the University advised him to approach the Scottish Hospitals Endowments Research Trust, which he did with Slater, and was rewarded with a donation of £4,000. The Trust also advised him to apply to the National Research Development Corporation (NRDC), an organization set up to assist British industry to compete internationally. This resulted in the NRDC committing a total of £10,000 to Kelvin and Hughes towards the development of diagnostic ultrasonic scanning over several years [43]. The crisis was over.

In July 1960, the automatic scanner was exhibited at the Third International Conference of Medical Electronics, held at Olympia, London, but failed to attract any commercial interest. At this meeting, Donald and Brown met Howry for the first time, initiating a long lasting, mutually supportive collaboration. Despite there being no real hope of it being a marketable product, owing to its sheer complexity, Brown rightly described the automatic scanner as "a lovely machine" and felt it had established a benchmark in image quality. As he had hoped, it demonstrated effective scanning technique to trainee operators, helping to improve consistency of scanning expertise. It was the means by which Donald and MacVicar developed their image interpretation skills and understanding of the clinical role of diagnostic ultrasound, producing around 3,000 scans of reasonably

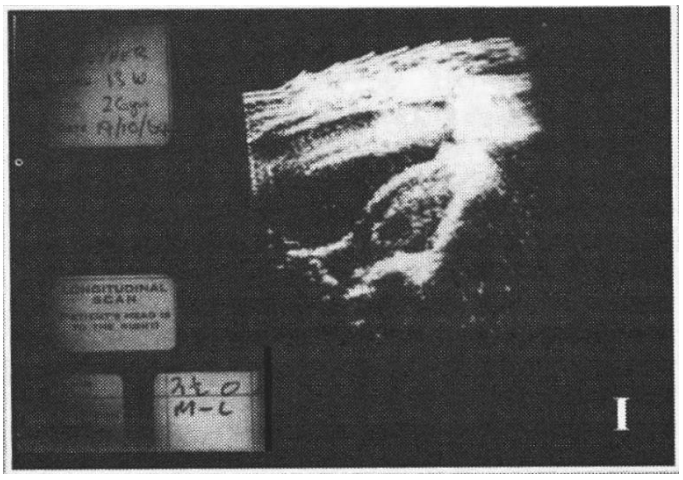

Fig. 13 Photographic record of the scan of an early pregnancy, obtained with the Automatic Scanner. Cards showing patient and scan details were included in the same exposure using an arrangement designed by Tom Brown. Photo courtesy of the BMUS Historical Collection.

consistent quality between 1959 and 1965 [24]. Figure 13 shows an example of an early pregnancy scan obtained

using the automatic scanner, together with scan and patient information written on cards that were included in the photographic record, a technique devised by Brown. It was replaced by the first manually operated Diasonograph (see Section IV below), but it went on to do non-obstetric/gynaecological service in the hands of radiologists Ellis Barnett and Pat Morley in the Glasgow Western Infirmary.

C. The 'Sundén' (Lund) machine.

Since 1953, Lund University had pioneered diagnostic applications of ultrasound, both to the brain by Leksell [9], and to the heart by Edler and Hertz [4]. Professor Alf Sjövall, Head of Obstetrics and Gynaecology, had taken an interest in this work and in May 1958 he had instructed a young doctor, Bertil Sundén, to use Leksell's equipment (a Krautkrämer flaw detector) to investigate the potential of ultrasound for his own discipline [44]. Donald visited Sjövall in June of that year to learn about the upcoming technique of laparoscopy and during his visit he spoke about his own ultrasound work. On hearing of the achievements in Glasgow, Sjövall arranged for Sundén to spend three weeks with Donald to learn what he could of the new ultrasound B-scan technique, which at that time was based on the bed-table scanner. On his return, Sundén obtained a grant from the Swedish Medical Council to buy a similar scanner for use in Lund. Much to the delight of Kelvin and Hughes, an order was placed in 1959 for an agreed cost of £2,500. Although, by this time, the automatic scanner was well advanced [23], Sundén requested a copy of the bed-table scanner he had used. This was never intended to be anything but a prototype and so a replica was out of the question. However, there was no doubt that only a machine with a hand-guided probe would be acceptable. Consequently, Brown and his managers at Kelvin and Hughes turned away from the automatic scanning approach for this and any future orders and decided to design a more refined manual contact scanner, with improved performance and ergonomics.

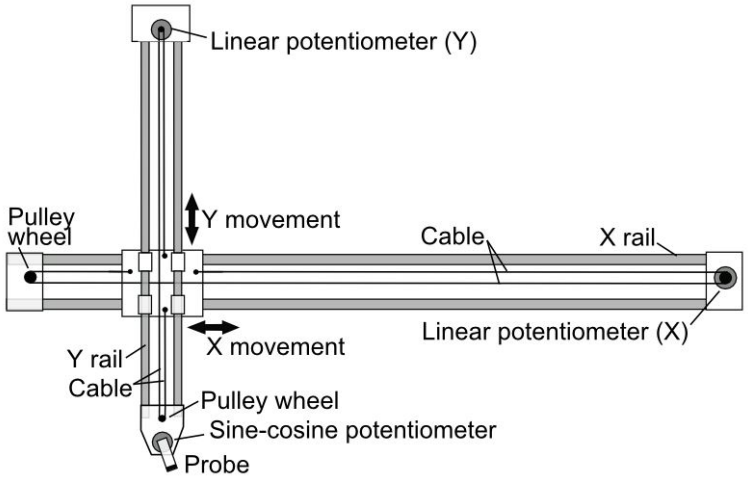

Fig. 14 Illustrating the principle of the system used to support and constrain the probe within the scan plane and to obtain the X,Y and angle coordinates of the probe relative to the scan plane.

Through his recent friendship with Howry, Brown was aware of the possibility of using an articulated arm (see section VIII *B*) to support the probe and measure its coordinates, but he continued to favour his original Cartesian coordinate method because of its intrinsically more rigid and accurate, albeit heavy and bulky, nature [23]. Figure 14 illustrates the general principle of the system of rails used to allow the probe free movement within a firmly defined scan plane and to measure the probe's X,Y coordinates and the angle of the probe to the Y axis. A photograph of the actual mechanics of a later version of the scanner (NE4102) is shown later in Figure 23b, but those in the Sundén scanner were basically similar. One of the designers at Kelvin and Hughes produced preliminary drawings of a machine to Brown's rough specification, but the company had a background in making equipment for industrial use and Brown felt the drawings did not suggest the kind of machine that would be appropriate for a clinical environment. Thanks to a mutual contact in the form of his sister in law, Brown met Dugald Cameron (Figure 15), a final year industrial design student at the Glasgow School of Art. Brown recognised that Cameron's talents and training could be just what was needed, so Cameron was commissioned to produce drawings showing how the aesthetics and the ergonomics of the design could be improved. Having first established that the machine should be planned around a single standing operator,

Cameron set about making the design, as far as possible, both ergonomically convenient for the operator and non-intimidating to the patient. An initial sketch by Cameron for the design is shown in Figure 16.

As may be seen in the photograph of the completed machine (Figure 17), an inverted-U outer frame was

Fig. 15 Dugald Cameron, in the east basement of the Mackintosh Building of the Glasgow School of Art, using an air brush to produce presentation drawings for the Sundén scanner. Photo reproduced by the kind permission of Dugald Cameron.

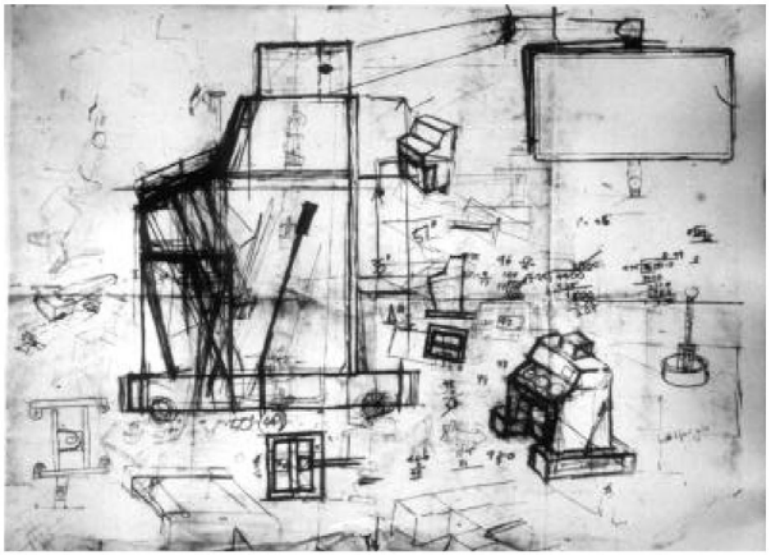

Fig. 16 Original sketch by Dugald Cameron of Kelvin and Hughes' concept for the Sundén machine. The Glasgow School of Art. By kind permission of Dugald Cameron.

supported over the patient by a hinged arm from a substantial column standing to the side of the patient. The arm, and all it supported, could be moved up or down, counterbalanced by a weight inside the support column [23]. This hinged arm allowed the outer frame to be positioned, as required, transversely or longitudinally with respect to the patient. The outer frame was free to rotate about a vertical axis at the end of the hinged arm, allowing the user to select a transverse, sagittal or intermediate scan plane orientation.

Fig. 17 The only known photograph of the original Sundén prototype in use in Lund in 1962. Photographed by Clive Ross, soon after he had installed it. Reproduced by kind permission of Mrs J Ross.

A chain system kept the angle of the outer frame constant with respect to the patient's longitudinal axis as it was moved transversely or longitudinally. Between the vertical arms of the outer frame, was a relatively slim rectangular box, enclosing the probe coordinate measurement frame and support arm. The position and orientation of this conspicuous 'probe support box' left the operator in little doubt as to the position and orientation of the scan plane. The box could be tilted about a horizontal axis to allow non-vertical planes to be scanned, for example in order to scan from the surface of the abdomen up into the rib cage or down into the pelvis. From the bottom of the box emerged the probe support arm, at the lower end of which was the probe holder. One of a range of probes could be easily inserted into this by means of a simple bayonet fitting. The probe could be rotated through ± 135° with respect to the 'Y' axis (scan plane vertical) within the scan plane by the operator, the rotation being transferred to, and measured by, a sine-cosine potentiometer located several centimetres above the probe on the same support arm. Separating the probe and the sine-cosine potentiometer in this way allowed the probe-holder to be smaller and neater, making it easier for the operator to grip and rotate the probe. The 'X' and 'Y' coordinates of the spindle, about which the probe rotated within the scan plane, were both measured by cable and pulley systems linking the spindle to linear potentiometers (Figure 14). The design also included the provision of accessible stowage for the range of probes and storage for the bottle of olive oil used for acoustic coupling. The job of overseeing the mechanical side of the project was entrusted to Brian Fraser, a senior development engineer within Kelvin & Hughes who had established a reputation as a practical designer in the field of marine instrumentation, leaving Brown free to concentrate on the electronics [23].

The electronics and their controls were housed in a console attached to the base of the support column on the side opposite the patient. A separate display console was suspended above the electronics console, on its own arm extending from the column. Although the structure was perfectly stable anyway, this gave the reassuring appearance of counterbalancing the heavy scanning frame suspended above the patient on the other side of the column. The display console accommodated two CRT screens for displaying B-mode scans, one with a long persistence phosphor for the operator to view and one with a short persistence phosphor for the Polaroid camera. This camera played a more fundamental role than simply providing a visual record of the images; it was intrinsic to the spatial compounding process since the final brightness of any target on the photograph was determined by the sum of several partial exposures, one for every time the ultrasound beam hit the target from a new direction [24]. It was, by now, considered that there was much less need for a dedicated A-mode display, although provision was made for an A-mode scan to be displayed on either of the B-mode screens if desired. As in the automatic scanner, Brown's continuing concern for safety meant that, as in the two previous scanners, the

receiver gain was set as high as electrical noise would allow and a switchable attenuator was provided that enabled the user to progressively increase the voltage drive to the transducer until usefully large echoes could be detected. Perhaps because this scanner was going out onto a world stage, his caution over safety was even greater and the PRF was reduced to 25 Hz, half that used in the automatic scanner [23]. An undesirable effect of this was the creation of gaps between lines of echoes whenever the probe was moved too quickly across the patient's skin; such gaps are evident in the B-scan of twins shown in Figure 18.

Fig. 18 Scan of a twin pregnancy clearly showing the two fetal heads, obtained using the Sundén machine in 1956. Photo courtesy of the BMUS Historical Collection.

After extensive testing by Donald and MacVicar at the Western Infirmary, the 700 kg machine was delivered to Lund, Ross travelling with it to install it. According to MacVicar, the date of delivery was 10th March 1962 [45], although according to Maršál and Sundén in Lund, it was in the autumn of 1961 [46]. Whatever the date, Sundén made good use of the machine, achieving impressive results and promoting the use of obstetric and gynaecological ultrasound in Europe as well as gathering material for an MD thesis "On the diagnostic value of ultrasound in obstetrics and gynaecology", which was examined by Donald and published in 1964 [47]. General maintenance was carried out by local engineers but, in March, 1964, John Fleming, an electrical engineer who joined the staff of the Hillington factory in 1962 as a development engineer on medical ultrasound projects, was sent to Lund to carry out a major overhaul [48]. Amongst the jobs needed was the replacement of the sine-cosine potentiometer used to measure the probe angle to the scan plane vertical. Wear of the wire-wound track due to the constant use of the machine and, it was suspected, poor standards of potentiometer manufacture, meant that echoes were being increasingly misplaced on the scan. Intimate acquaintance with the electronics of the Sundén machine and its limitations was to prove valuable to Fleming when helping to develop improvements for future commercial machines. Notwithstanding minor problems, the Sundén machine was much more refined than its two predecessors but, sadly, unlike them it was not destined to a final resting place in the Historical Collection of the British Medical Ultrasound Society (BMUS) but was to end its days in the hands of a scrap merchant [23].

IV. Smiths Ltd.

Meanwhile, in the Hillington factory, gradual progress was being made towards a production version of the Sundén machine. Work was also proceeding towards the production of a portable A-scan instrument with electronic calipers, to meet an expected demand for fetal cephalometry arising from the work of Duggan and Willocks. For a while, progress was hindered by major organizational changes to the company. For several years, Smiths Group had held a controlling interest in the company. In 1961, this resulted in the name Kelvin and Hughes changing to Smiths Industries Ltd. Problems in other parts of the company were addressed at the same time and, in the words of Brown, "medical ultrasonics had to take a bit of back seat" [23] as he and his team were asked to spend most of their time working on the commercial version of the semi-automatic industrial testing system with which he had made his mark at the start of his career. However, overall, the takeover did

bring improvements to the resources needed to develop the new ultrasound production machines. These included the recruitment of two engineers, John Fleming and Angus Hall, to join Brown in 1962.

A. The Smiths' Diasonograph.

In 1963, Brown tasked Fleming with revising the electronics for the new production machine while he oversaw the redesign of the mechanical side. Brown again invited Cameron to be involved, this time formerly as an industrial design consultant, working with one of the company's mechanical engineers, David McNair. The brief was to design a scanner that was physically safe for the patient and straightforward to manufacture. Possibly mindful of Donald's preference for the term 'sonar' rather than 'ultrasound', Brown suggested that the new diagnostic scanner might be called the 'Diasonic Scanner' and that the A-scan instrument might be called the 'Diasonic A-scope'. In the end, the names chosen were 'Diasonoscope for the A-scan instrument and 'Diasonograph' for the B-scanner.

Fig. 19 One of the first production Diasonographs in Smiths' factory, Hillington, Glasgow, c 1963. Also pictured is Arthur Johnson, a draughtsman involved in the project. Photo courtesy of the BMUS Historical Collection.

A major change to the mechanics of what was to be known as the Smiths' Diasonograph was the replacement of the elegant hinged 'elbow, shoulder and wrist-joint' arm that supported the measuring frame on the Sundén scanner with something that was less difficult and costly to manufacture. As may be seen in Figure 19, in the new design the probe support box and its outer frame, above the patient, were mounted at the end of a pair of substantial parallel steel tubes which could slide in and out of the side of a heavy and very stable, slab-sided cabinet. Inside this cabinet, the linear bearings through which the tubes passed were mounted on a counterbalanced slide, which could be driven up and down by an electric motor in order to adjust the height of the measuring frame above the patient. Brown still had in mind an ambition for 3D scanning at some future time so he ensured the counterbalancing would be sufficient to cope with the extra weight if the probe movements were to be motorized at some stage. The outer frame and probe support box were designed on similar lines to those in the Sundén machine, although they, and indeed the whole machine, which weighed nearly a ton, had a greater bulk and a less elegant look. It was not long before the machine acquired the nickname 'Dinosaurograph'. Reflecting the technical background of its makers, measuring scales were attached to the outer frame and box so that the coordinates and orientation of the scan plane could be recorded. In practice, little use was made of these, as operators preferred to use anatomical features on the patient, such as the umbilicus or symphysis pubis, to record the position of the scan plane. Brown's original intention was that the new machine would have a motorized couch that could move the patient longitudinally beneath the scanning frame [24]. Apart from being more convenient for the operator, a motorized couch would also have made it easier to modify the machine to automatically acquire sets of closely spaced transverse scans, should there ever be sufficient interest

in 3D scanning. However, cost considerations meant that a simple modified hospital trolley was used in the production version of the scanner.

The electronics of the Sundén machine were based on thermionic valves (vacuum tube devices). Fleming was aware of the recent progress in semiconductor devices and their many advantages, including greater robustness, smaller size and lower power consumption. He therefore offered to transistorize everything, but Brown felt such an undertaking would be too expensive and time consuming and instructed him to stick with valve technology. Fraser later commented [49] that the decision as to whether or not to adopt solid state technology around that time was difficult since, although manufacturers were aware that valves would soon be obsolete, solid state technology was not quite adequately developed. Fleming redesigned the power supplies from scratch, but in revising the timing and transmitting circuits he made direct use of circuits from the company's latest (Mark VII) flaw detector. The probe frequencies chosen were the same as used in the Mark VII flaw detector, namely 1.5 MHz, 2.5 MHz and 5 MHz, as this allowed him to use the new switchable amplifier of the Mark VII to replace the three separate plug-in amplifiers used in the Sundén machine. Besides being more convenient for the operator, this meant there was no need to provide protective storage for the amplifiers that were not plugged in. The probes were quick and easy to change, thanks to their simple bayonet fitting. Similar probe fittings were used on all subsequent evolutions of the Diasonograph; Figure 20 shows a photograph of the probe holder on the next incarnation, the NE4101 discussed in Section V *A*, below.

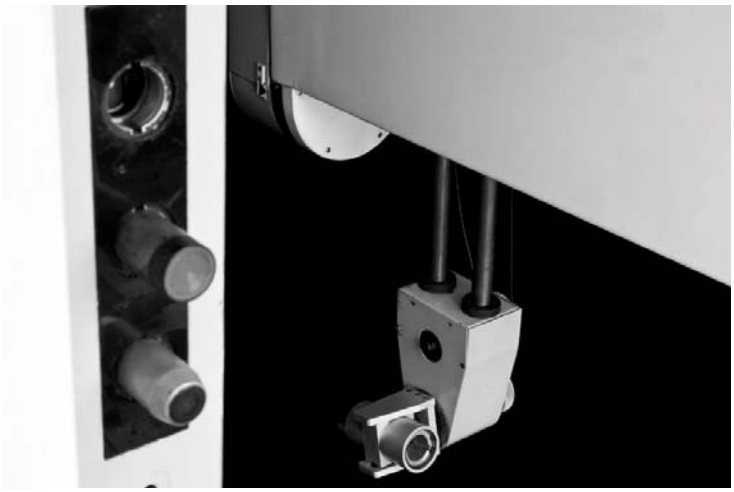

Fig. 20 The probe holder and probe storage facility on the NE 4101 Diasonograph. Photo courtesy of the Science Museum Group (https://collection.sciencemuseumgroup.org.uk/)

Cameron's original design for the Diasonograph was based on the assumption of a single operator, able to take advantage of a motorized couch. The decision not to incorporate a motorized couch meant that single-handed operation became difficult, although not impossible, and that, normally, two operators would be required - one to scan with the probe and another to operate the controls. In keeping with Brown's philosophy of trying to keep the scanning process as 'doctor-proof' as possible [24], there were three sets of controls on the electronic console, each requiring a different level of expertise (Figure 21) [50]. Covers were provided so that only the appropriate level controls could be revealed. The rightmost control panel housed the primary controls, consisting of the on/off switch, the sensitivity switch (in the form of a variable transmitter voltage attenuator) and the frequency selection switch (which changed the frequency band of the receiver RF amplifier). To the left of this panel was the panel containing the secondary controls, intended for more experienced operators. The main control was a switch to select either A-mode, M-mode (known then as 'Time-Position' (TP) mode) or B-mode (known then as 'Cross-sectional Display' or 'Section Scan'). There were also controls specific to each mode, such as 'scale', 'vertical shift' and 'horizontal shift' for B-mode, 'gain', 'reject' (suppression of weak echoes) and baseline time 'delay' and 'expansion' for A-mode, and 'sweep start' and 'sweep rate' for M-mode. The tertiary controls were less accessible, behind a fold down flap below the primary and secondary control panels. These were primarily for use by service engineers when calibrating the equipment and for technically advanced users, for example if engaged in experiments and research. Surprisingly, to users of modern diagnostic scanners, these tertiary controls included those for adjusting the swept gain.

A 'probe dispenser' was built into the outer frame, with a socket for storing each of the three probes (Fig. 20). An interlock mechanism between the frequency selection switch and the probe dispenser ensured that only the

probe that matched the setting of the frequency selection switch could be released for use. Conversely, it ensured that the frequency selection switch could not be moved if the probe matching its current setting was absent from its socket on the probe dispenser.

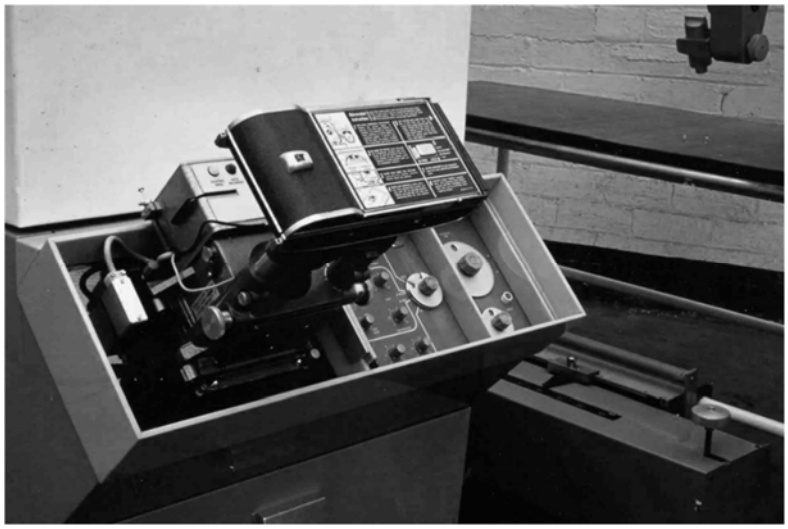

Figure 21. The controls on the console of the Smiths' Diasonograph, with the three levels of control and Polaroid camera, designed by Dugald Cameron. Photo reproduced by the kind permission of Dugald Cameron

The Smiths' Diasonograph was the world's first production model of an obstetric ultrasonic scanner, producing images that were superior to those of the Sundén machine and its predecessors. The Ministry of Health helped to stimulate the market by placing an order for four machines for research into the clinical application of ultrasound in UK centres. The first was installed in Glasgow's soon to be opened Queen Mother's Maternity Hospital in December 1963. Donald enjoyed experimenting with all the controls, including the tertiary ones, to the extent that he trained an assistant to scan patients with the machine, while he operated the controls. That assistant was Mrs Ida Miller, the "very efficient" wife of a local GP of Donald's acquaintance, recruited by Donald to help organize the scanning department [22] [51]. Over time, he allowed Miller to scan patients without his being present, thereby making her possibly the first sonographer in the UK, if not the world. In view of her lack of any formal medical qualification, he arranged later for a CCTV link between the scan room and his office so that he could exercise at least a nominal degree of supervision [52].

In 1965, Smiths Industries underwent internal organizational and management changes which did not suit Brown [23]. He left to work as Chief Engineer of Honeywell's Medical Equipment Division in Hemel Hempstead, working on equipment associated with heart surgery and coronary care, but no ultrasound. Fraser took over from him as leader of Smiths' ultrasound department, continuing to produce Diasonographs, of which twelve were sold by 1966.

Despite the sale of Diasonographs, by this time the Hillington factory of Smiths was losing money. It had also suffered a blow in losing a legal dispute with Automation Industries (USA) over global rights to patents concerning contact transducer systems, included some naming Brown as inventor [23]. The outcome was that, in 1967, Smiths decided to close their Hillington factory, giving up their investment in medical ultrasound. Determined to protect the British lead in obstetric and gynaecological ultrasound that he had done so much to establish, Donald turned again to the University of Glasgow, asking for permission to employ electronic engineers to maintain and develop the ultrasound equipment in Queen Mother's Hospital [53]. Much to Donald's amazement, the Principal, then Sir Charles Wilson, authorized him to set up the University Department of Ultrasonic Technology, complete with two full-time staff.

Fleming was pleased to accept a post as Research Technologist in the new department and encouraged his former colleague, Angus Hall, who had left Smiths two months earlier, to join him. The third member of the team was Jonathan Powell, a technician whom Willocks described as a "mechanical wizard" [53]. As a bonus, a large quantity of instruments and electronic components were acquired from Smiths for a nominal sum. They set to work correcting a variety of small but troublesome problems with their Diasonograph [54]. As on the Sundén machine, the wire-wound sine-cosine potentiometer was giving trouble. They found a permanent solution to this by sourcing, from the USA, a sine-cosine potentiometer with a plastic conductive track. Earthing problems were

also addressed, thereby curing a problem of transient oscillation which had occasionally spoilt the images. This, in turn, allowed them to increase the receiver gain, resulting in improved penetration and sensitivity [55]. The Smiths' Diasonograph remained in service in Donald's department at the Queen Mother's Hospital until 1972, by which time an NE 4102 Diasonograph (see below) had been installed there [56]. About this time, Donald underwent a series of three major heart operations in a four-year period, retiring soon afterwards in 1976.

V. NUCLEAR ENTERPRISES LTD.

In 1967, after a period of competitive negotiations, during which Brown visited and briefed Sam Davis, Under-Secretary at the Ministry of Health in London [24], Smiths' interest in medical ultrasound was acquired by Nuclear Enterprises Ltd, a small but successful Edinburgh-based company that made gamma cameras, amongst other products. They immediately recruited Brown back from his position with Honeywell and also employed Fraser, who had been made redundant when Smiths had closed the Hillington factory. Donald was hired as a consultant.

A. The NE 4101.

The first Nuclear Enterprises version of the Smiths' Diasonograph was marketed as the NE 4101 [59]. Built with the help of stock inherited from Smiths, the mechanics and electronics were virtually identical to the Smiths' Diasonograph but the layout of the controls was different. There was no longer a distinction between 'primary' and 'secondary' controls, and the PRF was switchable between 50, 150 and 300 Hz. The NE 4101 had a superior performance to the Smith's Diasonograph, particularly in regard to registration (positional accuracy of targets on the B-scan image) when compounding (insonating from different directions). The improvement was due to the combination of many small factors, including the use of higher quality components and more careful assembly [57].

Around 1968, Brown started to go through what he described as "a difficult personal patch, probably reacting to past domestic problems and a very uncomfortable period with Honeywell", and for a time he was fairly incapacitated [23]. In 1970 he left Nuclear Enterprises to become a Research Fellow in Medical Physics at the University of Edinburgh [24]. In 1973 he joined Sonicaid Ltd. to pursue his ambition to develop a 3D ultrasound scanner, as mentioned in section III *B*. The resulting '3D Multiplanar Scanner' [41] did not enjoy much commercial success and production ended in 1979, when Brown gave up his medical ultrasound activities to work in the oil and gas industry. In 1999 he moved back into the medical world, becoming Quality Manager at the Radiological Protection Centre, St George's Hospital, London. He retired in 2002, setting up a small firm, NoStrain, in 2005 to help sonographers who suffered from musculoskeletal disorders as a result of scanning [58].

B. The NE 4102.

Brown's departure from Nuclear Enterprises meant that Fraser again became the main driving force for further development of the Diasonograph to what was to be the NE 4102 (Figures 22 and 23) [60]. The electronics console, previously forming the base of the support cabinet, now became a separate movable unit. This allowed the operator to sit closer to the patient whilst still viewing the screen and gave more freedom of choice in the layout of the scanning room. Instead of one display screen, serving whichever mode the operator selected, there were now two screens, one with a short persistence phosphor, particularly suitable for producing photographic records with a Polaroid camera, and one with a variable persistence phosphor. In place of cathode ray tubes with associated circuitry built in-house, fully assembled and tested display units were bought-in from Hewlett Packard. Push-button switches next to each screen determined whether that screen displayed A-mode, M-Mode or B-mode. An electronic caliper, as pioneered by Duggan, complete with a prominent digital display, could be used with either display screen. The caliper velocity could be set by the user - a facility which sometimes led to confusion when comparing successive BPD measurements of the same fetus at different hospitals [61]. Although A-mode scans were no longer used much to interpret echo patterns, they were sensitive indicators of rapid tissue movement. A valuable application of this was confirmation of fetal life by directing the probe onto the fetal or embryonic heart; M-mode could then be used to provide a photographic record.

Like the display units, the power supplies for the electronics were bought-in as ready-built and tested modules, in this case supplied by Standard Telephones and Cables Ltd. Using ready-built and tested components like these meant that the electronics were more tightly specified and reliable, leading to noticeable improvements to the images, including superior registration to that of the NE 4101. A signal processing option known as

'Differentiation' was introduced. As its name implies, this involved differentiating the rectified echo pulses, so that a very short spike was produced by the rapidly rising leading edge of each demodulated echo. This spike was mixed with the original demodulated echo signal, thereby improving axial resolution, albeit at the cost of a substantial loss of grey tone discrimination. Three controls were provided to set the swept gain characteristics,

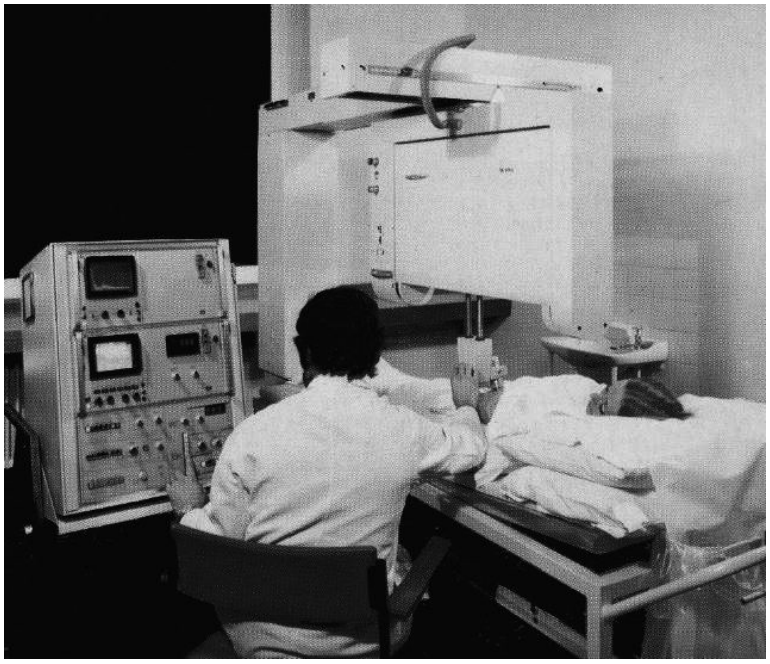

Fig.22. The NE 4102 Diasonograph. From Bulletin No. 64 - New Diasonograph NE4102 Diagnostic Ultrasonic Scanner. Nuclear Enterprises Ltd, 1972 [60].

arranged around a graphical representation of how the sensitivity increased with target depth. One control, labelled 'Initial attenuation (dB)' set the sensitivity close to the probe, a second, labelled 'Delay (mm)' set the depth range over which this initial sensitivity was maintained, and the third, labelled 'Slope (dB/cm)', set the rate at which the sensitivity increased with depth beyond the end of the delay.

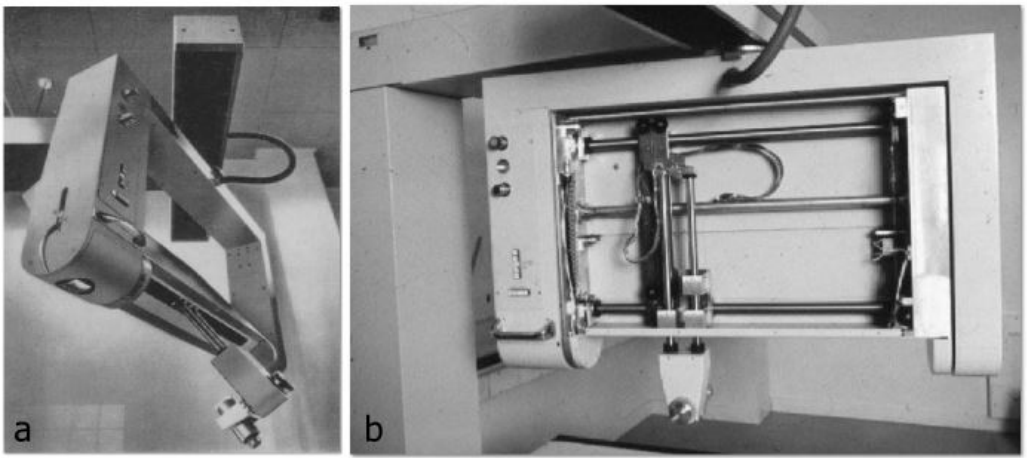

Fig. 23. a) Patient's view of the probe, the probe support box and the outer frame of the box, on an NE 4102. From Nuclear Enterprises Bulletin No. 64 [60]. b) View of the interior of the probe support box. The X and Y guide rails shown schematically in Figure 14 are prominent. Photo courtesy of the Design Council Slide Collection at Manchester Metropolitan University Special Collections.

As in previous Diasonographs, in order to minimise the ultrasound energy delivered to the patient, the sensitivity was varied by keeping the gain of the RF receiving amplifier as high as electronic noise would allow and using the sensitivity controls to vary the transmission excitation voltage In line with the growing confidence in the safety of diagnostic ultrasound, the standard PRF was increased to 600 Hz although a 'Velocity controlled' option was provided whereby the PRF varied between 60 and 1000 Hz according to the speed at which the probe was scanned across the patient, transmissions stopping altogether if the probe was held still in one position. The scanner's valve circuitry had been overdue for modernization and Fraser gave the task of re-designing the electronics using semiconductor technology to Alan Cole, described by Brown as "a very gifted electronics engineer" [23]. By the time the NE 4102 Diasonograph went on sale in 1972, the only valve remaining was a thyratron, a switch in the form of a gas-filled tube, used to discharge a capacitor across the transducer element. This was soon replaced by a newly developed solid-state switching device, the silicon controlled rectifier (aka thyristor) [57].

An example of a B-scan image of an early pregnancy obtained with this scanner, shown in Figure 24, illustrates the greater detail that could be seen as a result of the improved electronics. Combined with the rigidity and mechanical precision which had always been at the heart of the Diasonograph mechanics, this improvement in registration allowed more accurate size measurements to be made of anatomical structures. Hugh Robinson, First Assistant to Donald from 1972, took advantage of this to obtain accurate measurements of the crown-to-rump lengths of embryos and thereby establish a valuable method for assessing embryo growth [62]. The NE4102 was a marked advancement on the NE4101 and by November 1972 a hundred had been sold [63] in over forty countries around the world [64]. An NE 4102 installed in The Queen Mothers Hospital was in service until 1979 [56].

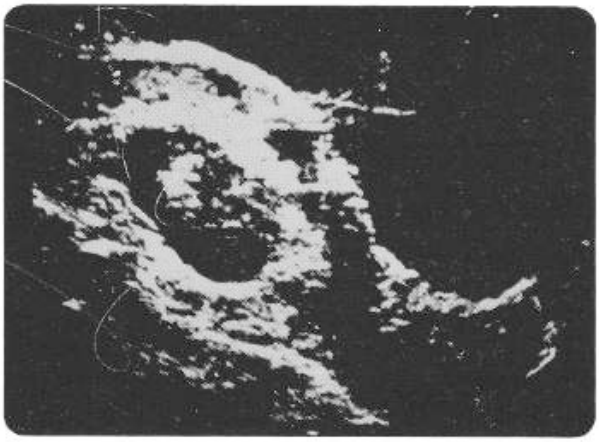

Fig. 24. B-scan of an early gestation sac, obtained with an NE 4102 Diasonograph. From Nuclear Enterprises Bulletin No. 64 [60].

C. Analogue Scan Converters.

In 1975, a major greyscale update was introduced for use with the NE 4102. Similar upgrades were being incorporated into scanners from other manufacturers at the same time, sometimes being retrofitted by medical physics departments [65]. These took advantage of the recent advances in image storage tubes, a popular example being the 'Lithocon' tube, as used in the PEP 400 and PEP 500 analogue scan converters from Princeton Electronic Products, USA [66]. At one end of an evacuated glass tube was a conducting 'target' consisting of a single crystal of silicon, on which was a mosaic of thousands of tiny non-conducting islands of silicon dioxide. There were three basic modes of operation: write, read and erase, carried out consecutively. In write mode, an electron beam emanating from a cathode, was accelerated and scanned across the target, using an X,Y timebase sweep, matching that of the CRT displays. By modulating the accelerating voltage of the electron beam with the amplified echo signal, negative charge was deposited onto the islands in proportions to that signal. By the end of each ultrasound scan, a charge pattern equivalent to the B-scan image had been painted onto the insulated mosaic. In read mode, this charge pattern could be read non-destructively and displayed on a TV monitor by scanning the same electron beam across the mosaic in a TV raster pattern. However, in this mode the voltage of the target was held only a few volts positive, so that the number of electrons reaching it was strongly influenced by the local negative charge on the island(s) immediately beneath the electron beam. The flow of

these electrons from the target formed the output current used to modulate the brightness of the TV monitor. In erase mode, another raster scan of the electron beam was performed, with the target voltage raised to several volts positive, so that electrons from the beam could flood onto all the islands of the mosaic and raise all their voltages to that of the cathode.

When first introduced, the operator had to monitor the build-up of the B-scan image on a long persistence CRT, since the scan converter had to remain in write mode over this time. Only when the scan was complete, could he/she select read mode to see the greyscale image from the lithocon target on the TV monitor. This limitation was soon overcome by multiplexing between read and write every few lines of the TV raster scan. Gaps in the image during the intervals of reading were avoided by writing for the first (say) 2 ms of a TV field, reading for the next 2 ms, writing for the next 2 ms, and so on, but on the next field reading for the first 2 ms, writing for the next 2 ms , and so on. The brief transitions between reading and writing bands produced thin dark lines across the screen, giving a so-called 'venetian blind' effect, but this was a small price to pay for the ability to see the progressive build-up of the grey scale image as the probe was moved across the patient. Analogue scan converters revolutionized the display of ultrasonic images as they could produce bright images with about ten levels of grey, giving a huge improvement in grey level differentiation between tissues. They also improved spatial resolution on compounded images due to the more controlled build-up of charge on a given island when the corresponding tissue target was repeatedly scanned from different directions. The density of the insulated islands in the target mosaic was sufficiently high to allow four separate images to be written, stored, read and erased independently without noticeable loss of spatial resolution. Read-out using a raster that covered the whole mosaic presented the four images together on the viewing monitor, a facility known as 'quad display' that was incorporated into some later Diasonographs. The relatively bright screen of a TV monitor meant that users could, at last, be freed from the requirement of working in relatively dark scan rooms. The TV format also meant that video storage and multiple viewing screens could be used, for example in consulting rooms or lecture theatres. Notwithstanding its advantages, the advent of B-mode images having a good range of grey tones was not universally welcomed. It was not unknown for some ultrasound users, who had learned to interpret images with no, or very few discernible grey tones, to deliberately throw away the new greyscale information by adjusting the dynamic range controls to produce images with the high contrast appearance they were used to.

A special version of the NE 4102, marketed as the NE 4102B, incorporated this greyscale facility, including the 'quad' display option, as standard, but a separate add-on package for existing NE 4102 scanners was made available as the NE 4104G Greyscale Storage Display [67]. An example of a greyscale B-scan image of an early pregnancy obtained with an NE 4102 equipped with an analogue scan converter is shown in Figure 25. Other accessories for use with the Greyscale Storage Display were introduced at the same time. These were: the NE 4106 Hard Copy Unit, based on a fibre optic recorder that produced approximately A4 sized (216 x 279 mm) grey scale images in just 12 seconds, the NE 4108 Video Cartridge Recorder and the NE 4210 Remote Photographic Facility, comprising a 6 inch (150 mm) TV monitor with a hinged adaptor to accept a Polaroid, or 70 mm, camera.

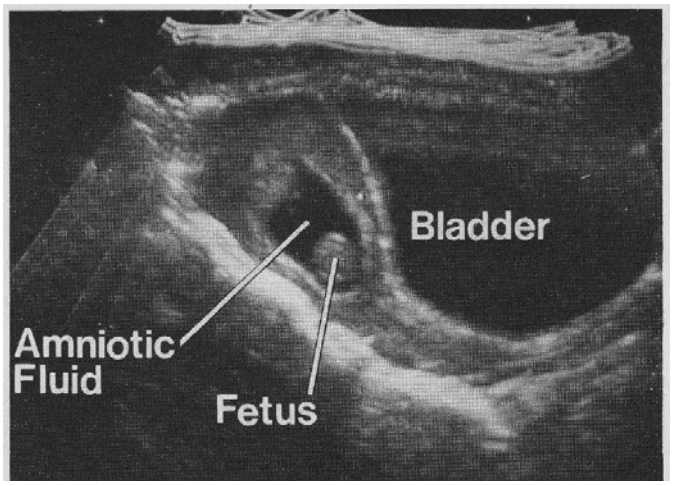

Fig. 25 Annotated greyscale B-scan of an early pregnancy obtained using an NE4102 fitted with an analogue scan converter. From Nuclear Enterprises Bulletin No. 434 [67].

D. The NE 4200.

In 1976, the NE 4200 was introduced [64]. Two high performance self-contained display units from Hewlett Packard were built into the electronic console, side by side. One was a short persistence HP 1332 for photography, but which could also display the TV image if required. The other was a variable persistence storage HP 1335, with a foot-operated erase switch. The NE 4200 could be ordered with greyscale imaging, based on a PEP 500 analogue scan converter and a separate TV monitor. The concept of greyscale enhancement of contrast resolution was taken further than on the NE 4102B, by providing an optional Colour Conversion Unit (NE 4204C). This comprised a 20-inch (508 mm) colour TV monitor and additional circuitry that added colour to the greyscale image. The signal amplitude range (window) over which the colour was applied could be varied by the operator to highlight subtle changes in echo amplitude, for example when looking for abnormal tissue, generating echo amplitudes that were only slightly different to echo amplitudes from surrounding normal tissue.

Different signal processing options could be selected by a row of four push buttons. One pair of push buttons selected between 'Greyscale' and 'Non-Greyscale'. The latter option provided what the brochure described as "outline-type scans of important structures" [64], but this option also made the machine more acceptable to those users, mentioned above, who were, at least initially, happier with the high contrast images they had become accustomed to. Another pair of buttons selected between 'Diff in' and 'Diff out', where 'Diff' stood for 'Differentiated', as explained earlier for the NE 4102.

Bayonet fitting probes with frequencies of 0.5, 1.5, 2.5, 5 and10 MHz were available. There was also a 2.5 MHz biopsy probe (NE 4167), principally intended for amniocentesis, having a central aperture sufficient for needles up to 1.96 mm in diameter. A handheld 2.5 MHz probe, for cardiology applications, was also available, intended for use with an optional cardiac module (NE 4103C), built into the console. This module allowed simultaneous presentation of M-mode, then known as Time-Position (TP) mode, ECG and PCG traces. The PRF remained at 600 Hz, as in the NE 4102, but it was increased to 1800 Hz in M-Mode when using the cardiac module or at the upper limit of Velocity Controlled mode. An NE 4200 installed in Donald's department in 1976 remained in service until 1985 [56], by which time real-time scanning was starting to become widely established, eventually to make so-called 'static scanners', including the Diasonographs, obsolete.

VI. EMI LTD.

By 1977, Nuclear Enterprises was producing fifteen Diasonographs every month, with a backlog of seven months of orders [68]. Production of ultrasound equipment was starting to dominate the company's activities and this change to the company profile did not sit easily with the directors. EMI, a large international organization with a proven track record in manufacturing medical imaging systems, had increased its shareholding in Nuclear Enterprises to ninety percent in 1976 [69] and in 1977 the directors of Nuclear Enterprises decided to sell their ultrasound division to them.

A. The EMISONIC 4200.

Soon after this takeover, the EMISONIC 4200 Diasonograph was launched. A brochure [70] photograph is shown in Figure 26a. This machine was basically very similar to the NE 4200, although there were a few minor changes, including the layout and style of some of the controls on the console. The 0.5 MHz probe was dropped from the range and a 3.5 MHz probe was added, part of an extended range of over sixteen probes with frequencies of 1.5 MHz, 2.5 MHz, 3.5 MHz, 5.0 MHz and 10 MHz. For probes of all frequencies, a choice of either unfocused, medium or long focal length was offered. A 2.5 MHz, unfocused biopsy/aspiratory transducer was also offered, having a 13 mm diameter and a central 2.4 mm aperture to accommodate a 14 gauge (2.108 mm diameter) aspiratory or biopsy needle. The acoustic power output and PRFs of the EMI 4200 remained as they were in the NE4200, except for a reduction in PRF to 1200 Hz when using the cardiac facility. As in the earlier 4102B model, a 'Quad' option that allowed four greyscale scans to be displayed on the TV monitor at the same time, each being independently written and erased. A new patient safety feature was added in the form of proximity detectors, built into the lower faces of the outer frame to ensure that neither the outer frame nor the probe support box could press down onto the patient. This and other features are indicated in an annotated view of the mechanical system shown in Figure 26b. A greyscale B-scan, obtained with an EMI 4200, of a sagittal cross-section of a liver with metastases, is shown in Figure 27.

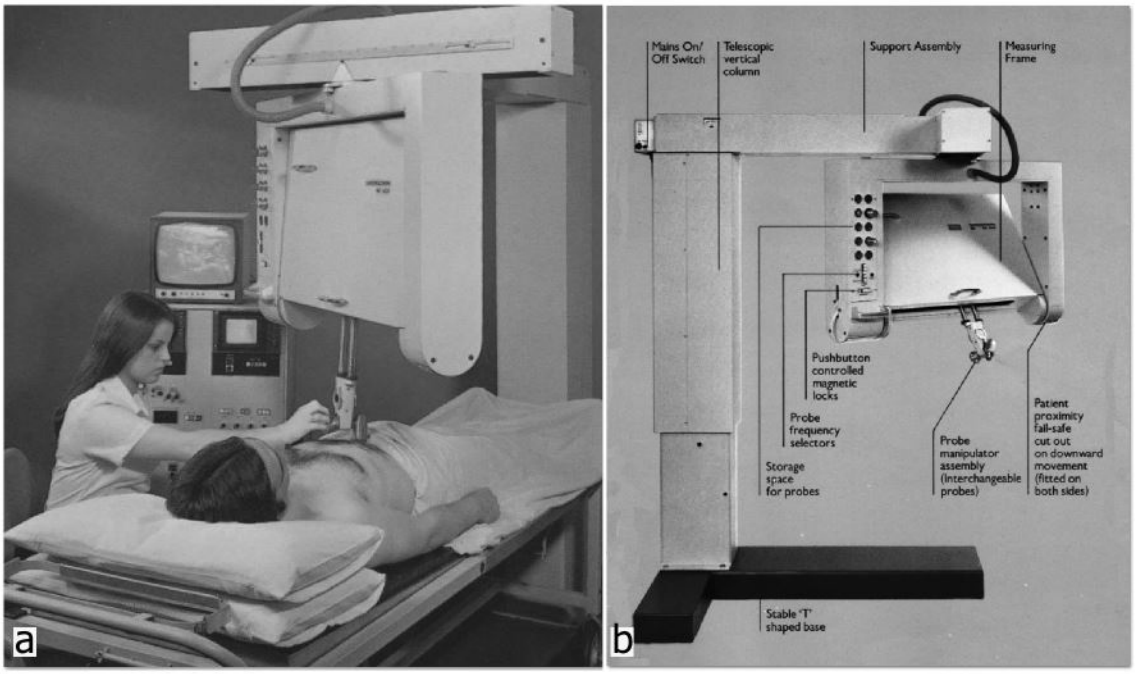

Fig. 26 a) The EMISONIC 4200 Diasonograph. b) Annotated image of the mechanical system. From Nuclear Enterprises Bulletin No. 112 (1977) [70].

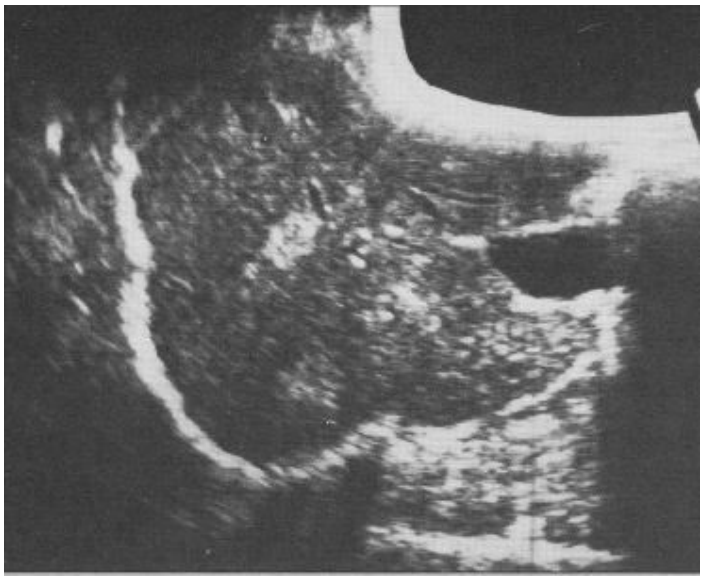

Fig. 27 B-scan of a liver with metastases, using the EMI 4200. From Nuclear Enterprises Bulletin No. 112 (1977) [70].

B. The EMISONIC 4201.

A second EMI Diasonograph, the EMISONIC 4201 (Figure 28) was launched in 1978 [71]. This machine had motorized movement of the outer frame and the probe support box, including automatic shifts (selectable between 2 mm and 40 mm) of the scan plane in a direction perpendicular to itself to facilitate acquisition of parallel sets of scans. A 9-inch (230 mm) TV monitor screen for greyscale images was built into the electronics console, alongside the short persistence and long persistence display modules. Every displayed and recorded scan could be automatically annotated with information such as the hospital name, patient number, scan serial number, scan plane coordinates and caliper reading. Potentially useful safety related information could also be

included automatically in any of the displays, such as the probe frequency, transmission attenuation, and even the total number of pulses delivered to each patient. The quoted values for maximum temporal average output power were slightly greater than on the EMI 4200. For example, for the NE 4238 long focus, 13mm diameter 2.5 MHz probe, supplied as standard on both machines, the maximum temporal average output power was quoted as 6.92 mW for the NE 4200 and 10.8 mW for the NE4201. Although only a small increase, and quite insufficient to produce significant heating, even of bone [72], this was an early pointer towards a more dramatic trend over the coming two decades for acoustic outputs to be increased in order to improve image quality [73]. Provision was made for the console to interface with the EMISONIC 4264 'Spinner' real-time sector scanner (Figure 37b). An EMISONIC 4201 was in use in the Queen Mother's Hospital, Glasgow, until 1987 [56].

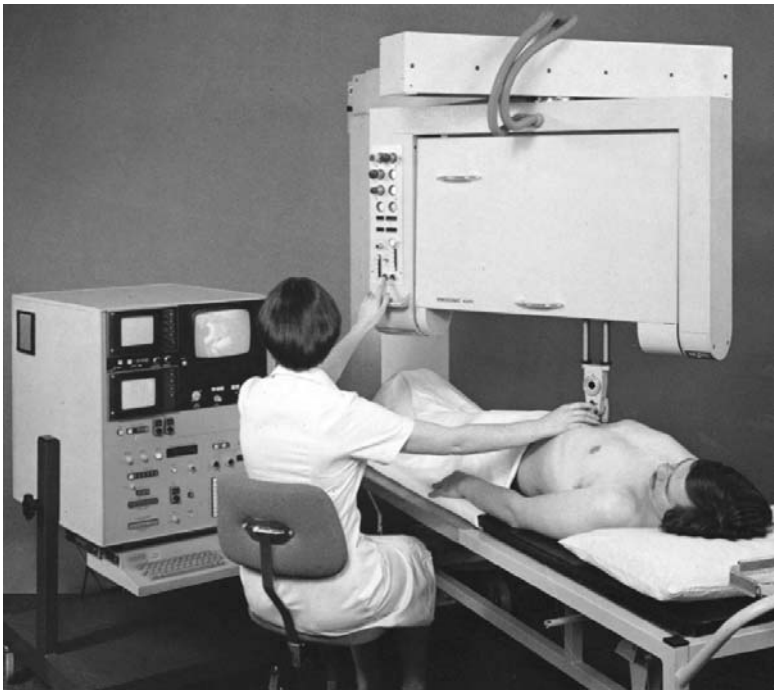

Fig. 28. The EMISONIC 4201. From Nuclear Enterprises Bulletin No. 116 (1978) [71]

VII. FISCHER LTD.

The final chapter of the Diasonograph story began in 1980 when EMI sold its medical ultrasound business to H. G. Fischer Inc., USA [69]. A new company, Fischer Ultrasound, was then formed, located in the former Nuclear Enterprises factory at Sighthill, Edinburgh.

A Fischer 4200S.

A modular ultrasound scanning system was produced, based on a console called the Fischer 4200S [74] (Figure 29a). This was similar to the console of the EMISONIC 4201 but without the long persistence monitor. The analogue scan converter was replaced by a digital scan converter, commercial examples of which were produced from 1976, although it took a few more years for their performance to match that of analogue scan converters [75]. This console could be combined with either the Diasonograph 4200 rectilinear B-scan mechanics, a more flexible articulated B-scan arm mounted on a substantial support column that was clearly from the Diasonograph stable [76] (Figure 29b), or a real-time hand-held sector-scanner probe (Figure 37c).

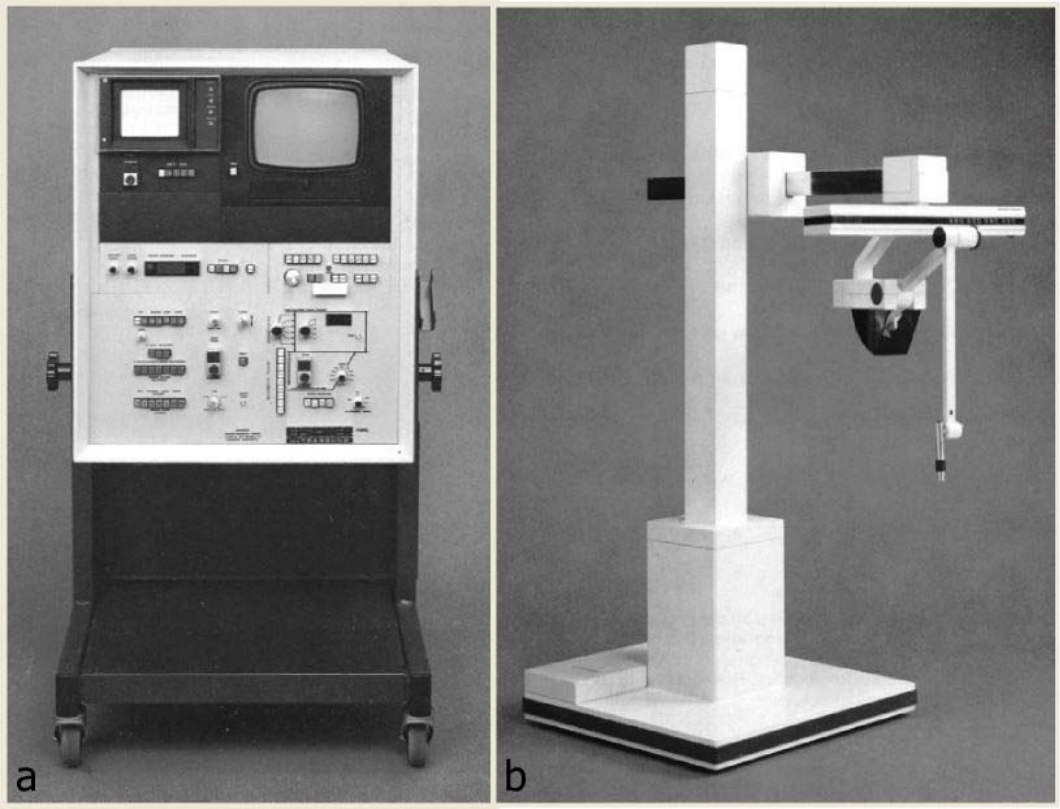

Fig. 29 a) Fischer 4200S console. From Fischer Ultrasound sales leaflet for 4200S [74]. b) Articulated arm option for the Fischer 4200S. From Fischer Ultrasound sales leaflet for the Articulated Scan Arm [76].

B. Other Fischer Ultrasound Products.

Fischer produced other diagnostic ultrasound products in the 1980s, principally real time systems such as the MARTI mechanical spinner [77], one of which was in the Queen Mother's Hospital between 1980 and 1986 [56]) and the LINUS linear array scanner [78]. Manufacturing continued in Edinburgh until 1995 [68].

VIII. CONTEMPORARY DEVELOPMENTS.

The purpose of this section is to lend some perspective to the Diasonograph story by briefly reviewing developments in medical ultrasonic imaging that were occurring elsewhere at the same time.

A. Other non-commercial B-scanning systems in the late 1950s.

In the late 1950s, before the Glasgow B-scanning work had resulted in a commercial system, researchers in other centres were also developing clinically useful ultrasound B-scanners. The water-bath approach was made more acceptable for clinical use by having the patient in acoustic contact with, but not immersed in, a water-filled bag or tank, within which the transducer was scanned. For example, in Japan, between 1954 and 1957, Surgeons Kenji Tanaka & Toshio Wagai at the Juntendo University, Tokyo, together with physicist Yoshimitsu Kikuchi and engineer Rokuro Uchida, built a scanner involving a water-filled bag that was lowered into contact with a patient lying on a couch [79][80]. In the USA, in 1957, Howry's group developed the 'Pan Scanner'. This consisted of a semi-circular water tank that half-surrounded the seated subject, allowing compound scanning of internal organs from a wide range of angles [81]. In 1960, at the Commonwealth Acoustics Laboratory (CAL) in Australia, later to become the Ultrasonic Institute, engineer David Robinson with physicist George Kossoff and obstetrician William Garret, designed and built a water-tank compound scanner, somewhat on the lines of the Pan Scanner but with a smaller angular width of arc [82]. It was specifically intended for obstetrics and had

much superior signal processing and compounding. The CAL group went on to develop systems for imaging other parts of the body, including eyes and breasts, setting world-beating standards in lateral resolution and signal processing. Some of this work is described in the next section (*B*).

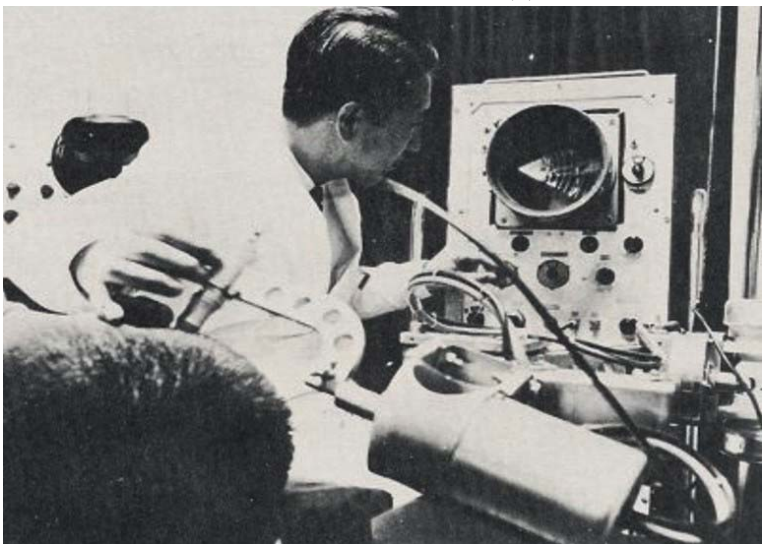

Fig. 30 The One-point Contact Sector Scanner of Kikuchi [83] being used by to scan a brain c 1957. It was also used successfully, later, in obstetrics and gynecology. Courtesy of the BMUS Historical Collection.

Unknown to Brown, as he worked on his bed-table contact scanner in 1957, a different form of contact scanner was being developed by Kikuchi and the Juntendo group. They called their technique 'One-point contact-sector scanning tomography' [83], since the transducer was constrained to rotate about its face, in contact with a fixed point on the patient's skin (Figure 30). Originally, the system was designed as a means of imaging the brain, for which a fixed point of contact on the skull had the advantage that artefacts due to variations in skull thickness were reduced, but the scanner later proved popular, in Japan, for scanning obstetric, gynecological and general abdominal cases.

B. Commercial B-scanners contemporary with the Diasonograph scanners.

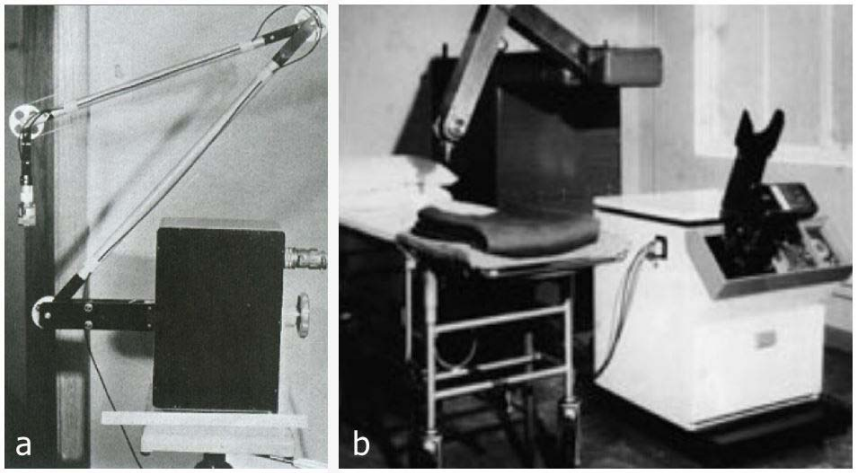

Fig. 31. a) The 'Porta-Scan' articulated-arm scanner designed in 1962, by engineers William Wright and Ralph Meyerdirk and commercially produced in 1963 by their own company, Physionic Engineering Inc., Colarado. b) A more sturdy articulated-arm scanner built by medical physicist Peter Wells, of Bristol General Hospital, UK around the same time as the Porta-scan. This was connected to an NE 4101 Diasonograph console, as shown. Photos courtesy of the BMUS Historical Collection.

In 1963, an articulated arm scanner, the 'Porta-Scan" (Fig. 31a) was produced by Physionic Engineering Inc., in Colarado. This company had been formed in 1962 by engineers William Wright and Ralph Meyerdirk, who had been working with Holmes and the Howry group. This type of probe support arm was much less bulky than the Diasonograph rectilinear rail-based system, and allowed easier probe movement, although it did not share its intrinsic rigidity and positional accuracy. At about the same time, medical physicist Peter Wells, at Bristol General Hospital, UK, developed a more sturdy example [84], which he interfaced to an NE 4101 Diasonograph console (Fig. 31b). Articulated arms became the norm in commercial compound contact B-scanners, other early examples being the Combison 1, from Kretztechnik in 1966, the SSD-10 from Aloka in 1967, the Picker Laminograph, a development of the Porta-Scan, in 1969.

Meanwhile, the era of water-bag scanners was far from over. In fact, the first ultrasound B-scanner to be commercially produced was the SSD-1 water-bag scanner (Figure 32), from the Japanese company, Aloka, in 1960. This company was founded by Uchida, previously with the Juntendo group, and the scanner was a development of their work.

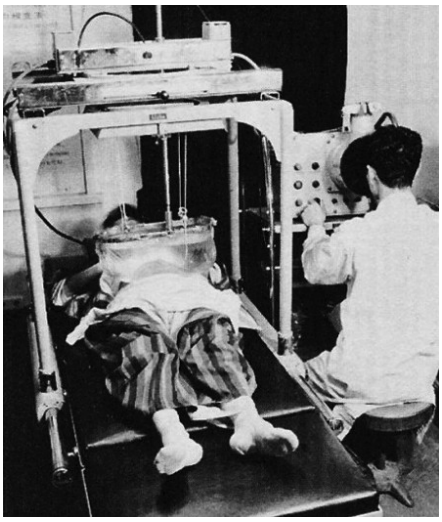

Fig. 32 The Aloka SSD-1 water-bag scanner, introduced in 1960. This was the commercial development of the system developed by the Juntendo University group (Section VIII *A*). Photo courtesy of the BMUS Historical Collection.

In 1965, a real-time water-bag-based mechanical scanner, the 'Vidoson', was commercially introduced by Siemens Medical Systems, Germany (Figure. 33). Developed in-house by engineer Richard Soldner, physicist Heinz Kresse and laboratory head Wolfgang Krause, this device involved the rotation of a wheel with three evenly spaced transducers set into its rim, rotating about the focal axis of a parabolic mirror. The beams reflected from the mirror remained parallel to the mirror's principal axis as they made repeated linear sweeps across the mirror's aperture, albeit at non-uniform speed. The Vidoson was the first commercial real-time ultrasound scanner; it achieved a frame rate of 15 images per second, each made up of 120 lines. The designers had initially planned to create a breast scanner, but the final device found its principal métier in obstetric and gynecological applications.

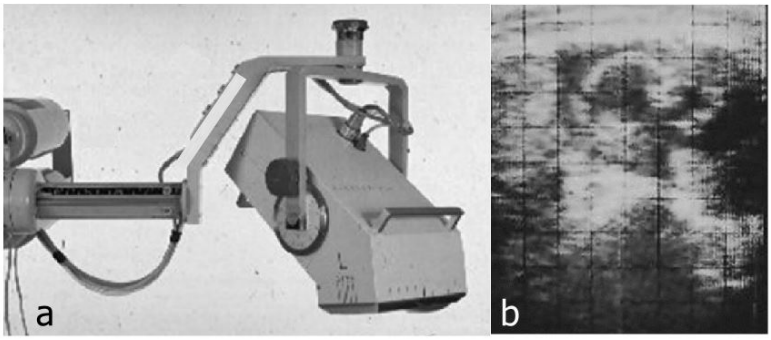

Fig. 33. a) The 'Vidoson' real-time water-bath scanner, was commercially introduced by Siemens Medical Systems, Germany, in 1965. As each of three transduces was rotated around the focal line of a parabolic mirror, its reflected beam was swept across the aperture, remaining parallel to the principal axis. b) An example of a single frame from an obstetric patient. Courtesy of the BMUS Historical Collection.

The mid-1970s witnessed the zenith of water-tank scanning. This took the form of the Australian 'Octoson', which involved the patient lying on a flexible waterproof sheet, forming the upper surface of a water-filled bath (Figure 34a). This scanner was the commercial version of a prototype annular array compound scanner developed by the team at CAL between 1973 and 1974 [85]. It was marketed from 1975 by Ausonics Pty., a company created by the laboratory (by then renamed the Ultrasonics Institute) for the purpose. Water tanks have a unique advantage in that they allow the use of large aperture annular array transducers, which can produce very narrow, dynamically focused, transmit-receive beams, and hence much better lateral resolution than is possible with a contact scanner. By using eight annular array transducers, arranged in an arc (Figure 34b), the time for each scan was kept to just 4 seconds. Each transducer was rocked through a range of angles such that eight sector-shaped scans were generated, each having 500 lines. These sector scans overlapped in the target region of the patient, producing a compound scan with precisely defined registration, as well as high line density, image uniformity and grey scale (Figure 34c). More than 200 Octosons, costing around $A100,000 each, were sold world-wide, before real-time, hand-held scanners became dominant.

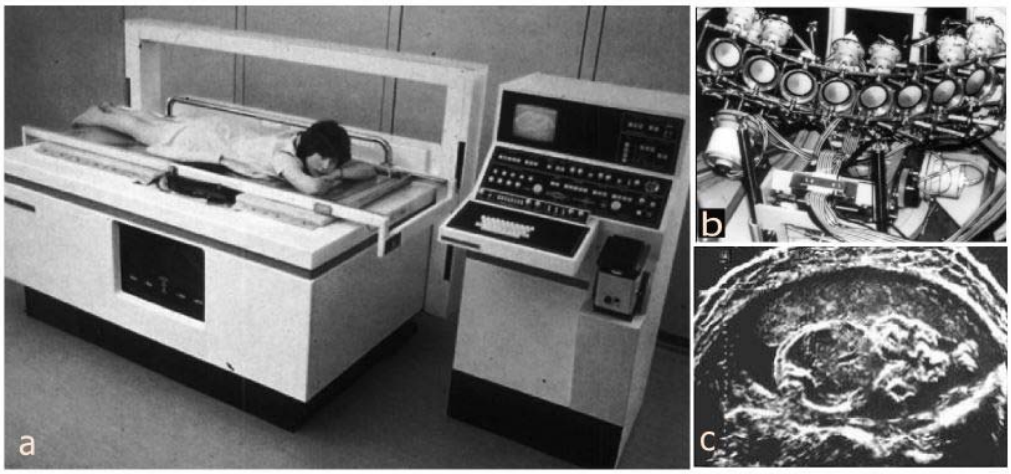

Fig.34 a) The 'Octoson' water-tank scanner, developed by the Ultrasonics Institute, Sydney and marketed, from 1975, by Ausonics Pty. b) Sector scan images from eight large aperture, dynamically focused annular array transducers, were combined to form a compound B-scan image. The image quality was unrivalled for its time, as illustrated by the obstetric example (c). Images from the Octoson sales brochure.

C. The real-time revolution.

A revolution began in the late 1960s and early 1970s that would ultimately lead to the near universal use of real-time, hand-held scanners. These can be categorized broadly as working either by: electronically stepping the beam along a linear array of transducer elements (linear array probes), electronically deflecting the beam to scan in a sector format (phased array probes), or mechanically sweeping the beam through a sector or in a linear fashion (mechanical scanners).

i. Early linear array probes.

In the west, it was long accepted that the first prototype diagnostic linear array system was one built by a group at Erasmus University, Rotterdam, led by Nicolaas Bom [86]. In fact, unknown to western researchers, at the same time as this group was building a scanner that gave an image of just 20 lines, a vastly more sophisticated prototype linear array scanner, with 181 channels (scan lines) and 200 narrow rectangular transducer elements, acting in groups of 20, was developed in Japan [87]. It was the work of Takasuke Irie and his supervisors Yoshio Hagiwara and Rokuro Uchida of the company 'Japan Radiation and Medical Electronics Inc' (later to become Aloka). Ignorance of this remarkable Japanese achievement continued through the 1970s and beyond, as western researchers and manufacturers produced their own linear array scanners. The first commercial example, in 1972, was the 20-channel Organon Teknika 'Multiscan', a direct implementation of Bom's 20-channel system. This was followed, in 1973, by the much superior and very popular 61-channel ADR scanner (Figure 35), made by the Advanced Diagnostic Research Corporation in the USA. A second ADR scanner, the 'ADR 2130', with electronic focusing, was marketed in 1975. Concurrent with, but unaware of, American developments, the author worked on ways of increasing the number and density of scan lines. The

most successful method was to use adjacent groups of narrow rectangular elements [88] [89]. A commercial spin-off from this work was 'RITA', a 40-channel add-on system from GEC Medical Ltd for enhancing an existing A-mode or M-mode instrument with a real-time B-scan facility. In 1976, a newly formed, Edinburgh-based company, Diagnostic Sonar, produced 'System 85', a 48-channel linear array scanner: This had a simultaneous A-scan display and prominent electronic caliper, reflecting the background of the company founder, Hans Gassert, who had hitherto worked for Nuclear Enterprises in marketing and sales.

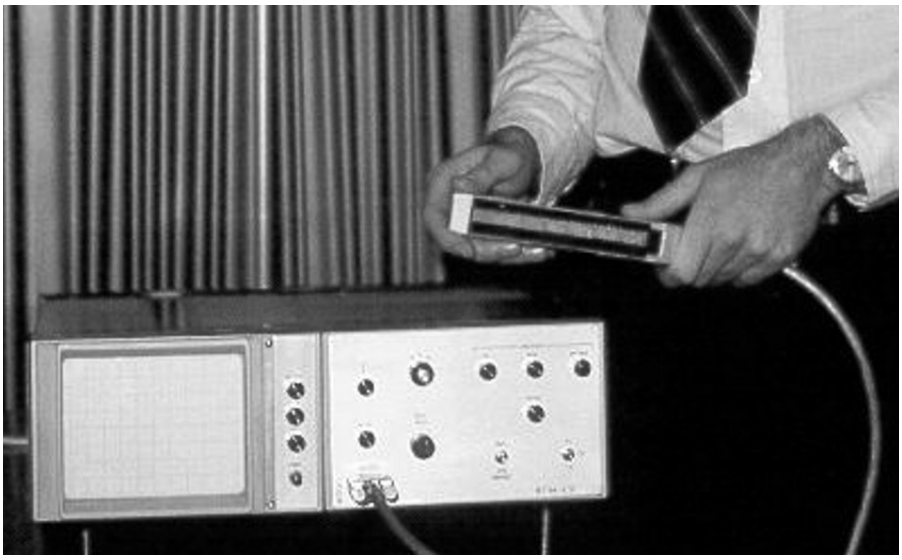

Fig. 35. The popular 61-channel ADR linear array scanner, produced by the Advanced Diagnostic Research Corporation in the USA, in 1973. Photo courtesy of the BMUS Historical Collection.

ii. Early phased array (electronic sector scanner) probes.

Phased array probes were first investigated as means of medical diagnostic imaging by Jan Somer at the Institute of Medical Physics-TNO, Utrecht, the Netherlands, in the mid-1960s. His array probe (Figure 36a) covered a 90° sector with only 21 elements, operating at a frequency of just 1.3 MHz, as he was primarily interested in imaging the brain through the intact skull, although some heart imaging was undertaken [90]. The

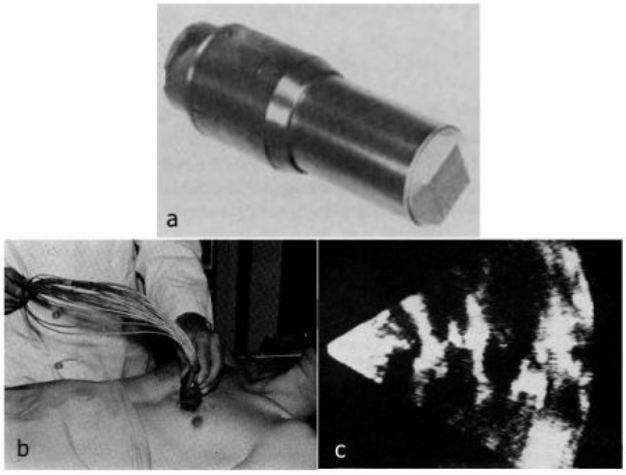

Fig. 36 a) Phased array probe made by Jan Somer of Utrecht University in 1968. Courtesy of Jan Somer. b) Prototype phased array from Frederick Thurstone's group at Duke University. c) Example of a heart scan from the Duke University. Photos b) and c) reproduced from Kisslo et al (1976) [93].

production version of this system was marketed as the 'Echostat' by the Diagnostic Electronic Corporation, USA in 1976, making it the first commercial diagnostic phased array scanner. However, it was too far ahead of its time and the limited technology available to Somer did not do the concept justice; consequently only a few were sold before it was overtaken by competition from commercial manufacturers [91]. The potential of the technique was further demonstrated by a group of biomedical engineers at Duke University, Durham, N Carolina, led by Frederick Thurstone, using digital technology (Figures 36b and 36c) [92][93]. By the end of the 1970s, technological advances meant that systems with much higher image quality from manufacturers such as Varian and Hewlett Packard entered the market.

iii. Early mechanical real-time probes.

Mechanical real-time probes involving reciprocating, linear scanning of transducers are subject to vibration problems when used with lower frequency, and therefore larger and heavier, transducers. Hence, for abdominal or cardiac use, sector scanners are the norm, and only this type will be mentioned here. The first mechanical real-time probe for scanning the heart was built in 1973 by James Griffith and Walter Henry at the National Institute of Health [94]. A 12mm diameter, 2.25 MHz transducer was driven by a crank attached to an electric motor such that it rocked back and forth continuously through a 30º sector at a rate of 15 Hz. The transmit pulses and echo processing were provided by a Smith Kline Ekoline-20 A-scanner, at a PRF of 2 kHz. Thirty images of clinically useful quality were produced every second, each consisting of 66 scan lines. In an effort to reduce the problem of vibration associated with rocking a transducer back and forth, in 1975, Hans Hendrik Holm and colleagues at the Gentofte Hospital, Copenhagen, Denmark built a system featuring a probe containing a 6 cm diameter wheel that spun continuously against an oil film on the patient's skin [95]. Four evenly spaced transducers were set around the rim of the wheel, each being connected, in turn, to the transmit/receive circuitry as it passed over the skin. This produced 16 sector images per second, each consisting of 69 scan lines in a sector width of 50 °.

Commercial mechanical real-time probes, of both 'rocker' and 'spinner' types were produced from the mid-1970's. In 1974, Toshiba, in Japan produced their first prototype real-time mechanical sector scanner, the SSL-51H. It produced 30 images per second, each consisting of 120 scan lines, it had a variable sector width from zero to 65° and the choice of interchangeable focused or unfocused transducers. In 1975, engineers Reginald Eggleton and Kenneth Johnston in Indiana, USA, described an 'add-on' system which could be connected to an existing M-mode heart scanner to enable it to provide real-time sector B-scans [96]. It comprised a compact, hand-held, cylindrical probe with a transducer at one end, rocking through ± 15° about the principal axis of the cylinder, and a plug-in module of associated electronics. Two transducers, at 2.25 MHz for adults and 3.5 MHz for children, could be interchanged, and the frame rate (0 to 60 Hz), PRF and depth of view could be varied. This system, with its 'rocker' type of probe achieved commercial success as the 'Cardioscan'.

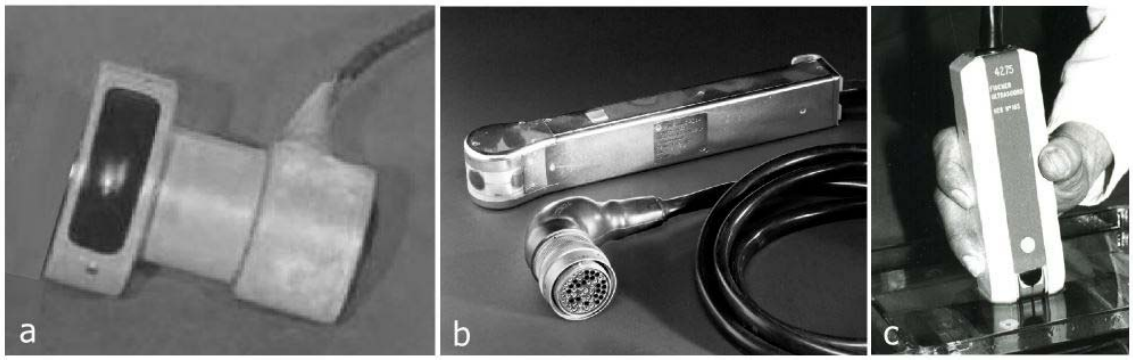

Fig. 37) a) The probe of the Combison 100 mechanical real-time scanner, produced by Kretztechnic in 1976. b) The EMISONIC 4264 'spinner' mechanical real-time sector scanner, offered as an option with the EMISONIC 4200 Diasonograph. This probe was developed by Norman McDicken's medical physics group at Edinburgh University. Photo courtesy of the Science Museum Group Collection © The Board of Trustees of the Science Museum. c) The Fischer 4275 further development of (b), available from around 1980 as an option with the Fischer 4200S. Photo courtesy of Norman McDicken.

In 1976, the Austrian company Kretztechnik produced the Combison 100. The probe of this scanner had a relatively large (approximately 10 cm) diameter continuously spinning wheel within a close-fitting oil bath, which had a roughly 5 cm long aperture, covered by soft plastic, in its rim (Figure 37a). Five 3.5 MHz transducers were evenly spaced around the rim of the spinning wheel, each of which became 'live' as it passed

across the aperture. The large wheel and wide aperture gave the advantage of a wide field of view close to the probe. In 1979, Norman McDicken and his medical physics group at the University of Edinburgh described their system [97] in which 4 transducers, set in the rim of a continuously spinning, 3 cm diameter, wheel at one end of an approximately 20 cm long probe could generate a sector of up to 180°. This system was more compact than previous 'spinner' designs and was produced commercially as a real-time plug-in option (Figure 37b) for the EMISONIC 4200 Diasonograph and later, in an even more compact form (Figure 37c), for the Fischer 4200S, as discussed in Sections VI and VII.

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APPENDIX

Reproductions of the sales brochures and leaflets of the Diasonographs and related commercial equipment referred to in this article, as well as a list of the locations of archived Diasonographs, are available in the appendix at the end of this special issue (pp 660-730).

AUTHOR BIOGRAPHY

Tony Whittingham ARCS BSc MSc PhD FInstP FIPEM graduated in Physics and Maths from Imperial College, London in 1965, before working on the dielectric properties of tissues in the Medical Physics Department at Aberdeen University. He became a Lecturer in Ultrasonics there in 1970, giving him the opportunity to become well acquainted with the NE 4101 Diasonograph.

From 1972 until retirement, he was Head of the Ultrasound Section of the NHS Regional Medical Physics Department in Newcastle upon Tyne, making pioneering contributions to linear array scanner design, developing CT and other ultrasound imaging systems, and designing instruments for measuring beam shapes, acoustic pressure, power and intensity. He has written and lectured extensively on beam-forming, ultrasound safety and modern developments in ultrasonic imaging technology. He has chaired the safety committees of BMUS and EFSUMB, co-authored Standards for the IEC and is a Past-President and an Honorary Member of the British Medical Ultrasound Society.

HEWLETT PACKARD - INNOVATIONS THAT TRANSFORMED DIAGNOSTIC ULTRASOUND IMAGING

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I. SETTING THE STAGE

This is the chronicle of how Hewlett Packard Medical Products in Andover, Massachusetts, USA (HP) became the world's leading echocardiography company starting as a concept at Hewlett Packard Research Laboratories in 1973 until its transfer to Agilent Technologies in November 1999. The author worked at HP in Andover from early 1981, just before the first system shipped, until the transfer of the medical products group in 1999. Before we start, what was the company's interest in medical ultrasound and what was the state of the art in ultrasound imaging?

Hewlett Packard, an instrument company, acquired the Sanborn Company in 1961 with the anticipation of entering the medical instrumentation market. At that time, Sanborn had a number of premier products including electrocardiographs and other medical test and measurement instruments. In that same year, HP's common stock first appeared on the New York Stock Exchange. The original Sanborn had about 950 employees and was located in Waltham, Massachusetts just outside Boston.

After the acquisition of Sanborn, HP explored new diagnostic markets that would add to and complement the recording systems and monitoring equipment that it currently sold. New diagnostic products were aimed at both clinical researchers in academic medical centers. The 7214A Ultrasound Diagnostic Sounder, a general-purpose instrument useful for cardiac, ophthalmological, neurological brain, fetal and kidney examinations was introduced in 1967. It could be used with 1.0, 2.5,5.0 and 10.0 MHz transducers. Interest in neurological ultrasound brain studies led HP to introduce products specifically for neurologists, the 7215A and later the 7215B Ultrasound Echoencephalograph. Their primary purpose was to characterize midline shifts and brain tumors. While the 7215A and 7215B enjoyed interest not only from neurologists but a number of other disciplines as well, the market was judged not to be large enough for more instruments at that time.

John Hart worked on both the 7214A Diagnostic Sounder and the 7215A/B Echoencephaloscope which was used to view the midline of the brain using ultrasound in either a pulse-echo or a through transmission mode. This A-mode (amplitude mode indicating echo amplitudes versus time) device was constructed as a plug-in to an existing HP 140T storage oscilloscope. It had a time gain compensation control, later to become a standard feature in ultrasound.

Just as this product derived from existing HP technology, future HP imaging systems would utilize features of the 78500 series Patient Information Center which centralized the monitoring of patients' vital signs at a nurses' station including ECG) [1] and was designed by the HP Waltham group in 1980. Once again, HP's strengths in waveform display and measurement were evident in the monitoring system as well a new multi-microprocessor architecture which included a newly developed MC5 microprocessor chip. The HP MC5, made with Hewlett Packard's CMOS exclusive silicon-on-sapphire (SOS) process, was one of the fastest 16 bit microprocessors available at 5MHz, and featured a high level instruction set to simplify the routing of data and later played a key role in HP's first ultrasound imaging system.

Meanwhile, in the 1970's, ultrasound imaging was diverging from single transducers on mechanical scanning arms and storage displays into arrays of transducers electronically scanned. The array development was split into linear and phased arrays. Whereas the A (amplitude)-line was one dimensional, the addition of an electronic geometric scanning method introduced the second dimension. In Figure 1, a simple linear array, introduced and popularized by N. Bom [2], in the early 1970's, is shown. Here a row of individual transducers are sequentially addressed by transmitting a pulse and receiving echoes along a line (like an A-line) and displaying each of these lines in their proper sequential geometry on a storage scope. Unbeknown to them, K. Irie at Aloka Japan, had demonstrated [3,4] linear arrays in which overlapping groups of elements were combined and sequenced to provide lines of much higher sensitivity. Whittingham independently had a similar idea of shifting elements in groups along the array and showed abdominal, obstetric and cardiac images using his scheme in 1975 [5]. Even earlier, by the early 1950's, J.J. Wild and J. Reid had taken advantage of World War II surplus radar units to demonstrate ultrasound imaging with handheld linear arrays [6].

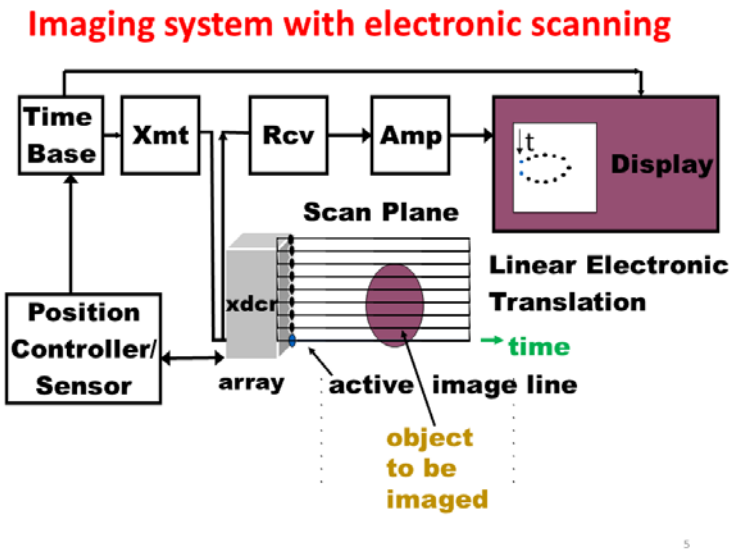

Fig. 1 Basic setup for B (Brightness)-mode imaging in which elements in an array are sequentially switched to produce image lines. (From [8])

Phased arrays, while superficially similar in physical appearance to a linear array, in that array elements are most often arranged in a row, differ in their scanning mechanism. Phased arrays in ultrasound were derived from sonar and radar arrays which swept a radial beam in a full circle or a sector of a circle for a plan position indicator (PPI) display that represented the radar antenna in the center of the display, with the range distance being the radial arm. The distinction between the scanning mechanisms is illustrated in Figure 2 [7]. Linear arrays incrementally scan laterally as Δx, whereas sector scanners change in angle line by line, as $\Delta\theta$.

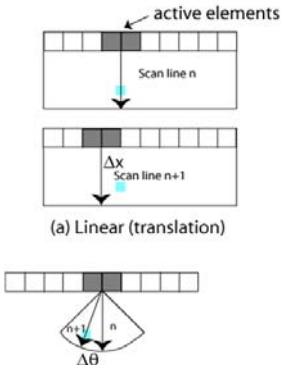

Fig. 2 Scanning mechanisms:A. Linear translation (rectangular image) B. Angular translation (sector image) (From [8]).

Historically, the term "phased array" came from radar where alterations in phase (within one cyclic period) were used to manipulate the beam. Somer [3,9] realized that for ultrasound, time delays were more appropriate for steering. He reached this conclusion from the need to send short pulses to achieve high axial resolution along the direction of propagation which, in turn, required wider bandwidths. His Electroscan 1 and 2 systems demonstrated [3,10] electronic steering of an ultrasound beam for imaging of the brain. To include wider bandwidths, he employed 30-32 meters of cables for finely tuning the delays needed for steering and three racks of equipment. At a Paris conference in May 1971 [3], Professor Frederick Thurstone asked to borrow a film made by Somer's system as an aid to obtain funding for his imaging work at Duke University.

Working with his student Olaf van Ramm, Thurstone designed a sector scan ultrasound imaging system with a major improvement: focusing. They used a PDP-11 computer to create rapidly the necessary time delays from wideband lumped constant analog delay lines to both steer and focus an ultrasound beam. Their initial system

employed a 1.8 MHz sixteen element array with a footprint (active area) of 14 mm x 24 mm. By compressing the echo signal envelopes logarithmically as brightness modulation on an HP 1311 scope display, they demonstrated a greater range of echo amplitudes in cardiac images than previously possible. They were able to enhance image resolution by introducing dynamic focusing on reception of the echoes. Their work, presented as demonstrations and video tapes at conferences and in two seminal papers [11,12] showed that real-time cardiac imaging was possible and attracted international interest. The present author was amazed to see their images at the IEEE Ultrasonics Symposium in 1976 [13].

II. Hewlett Packard Laboratories: Prototype Ultrasound Imaging System

Dave Wilson and John Larson, who worked together in Hewlett Packard's Research Laboratories in Palo Alto were inspired by Thurstone and van Ramm's talk on their new ultrasound imaging system employing two-dimensional electronic beam steering, given at an Acoustical Holography conference at a nearby hotel in 1973 [14]. Dave Wilson, a physicist who had been interested in medical imaging, was excited by the talk and became a champion for initiating an ultrasound imaging project at the labs. John Larson, who had just completed his PhD in Electrical Engineering at Stanford in 1971, specializing in piezoelectric ultrasound devices, was also on board. "Let's build one!" exclaimed Dave Wilson. With a four element array and crude imaging system, they helped convince Ed Karrer, who in turn obtained support from higher management for the project.

For Ed Karrer, it was the right idea at the right time. Hewlett Packard was uniquely poised for ultrasound imaging; it had scientists with the relevant expertise, experience in crystallographic quartz resonator standards, analog radar beamformers and chip design, state of the art fabrication facilities, many measurement and instrument products including the HP 2100 real-time minicomputer controller and HPIB interface bus for interconnecting instruments as well as a good relationship with the Stanford Medical School. The team was put together in late 1973 with Ed leading the group including John Larson, Dave Wilson and Rick Pering, a EE design engineer; they put together a prototype imaging system with the HP 2100. Within a few weeks, they had a working system with a 16 element array and the HP controller; however, they found that their array had grating lobe artifacts [15] and insufficient dynamic range. Still, they were able to see valves of the heart. They sought the advice of Dr. Richard Popp, a Stanford physician, who gave valuable feedback for improvements. He explained that they were seeing a heart valve but also other ghost valves which turned out to be artifacts of the imaging system (due to grating lobes). Scaling up the delay line approach to more channels that were needed to improve resolution and eliminate grating lobes would be prohibitively expensive. In 1975, Sam Maslak, a freshly graduated MIT PhD in EE joined the team to work on system architecture. A picture of the system in 1976 is shown in Figure 3. Amin Hanafy, who had some experience in designing phased arrays joined them in 1976 from the HP medical group in Massachusetts. Later Fleming Dias aided with transducer construction.

The overall goal was to develop a working proof-of-principle prototype system and transfer the technology to the HP Andover division for a design of a real-time cardiac ultrasound system. Several major challenges lay ahead for the team. The major ones were to improve image quality and real-time frame rate and to provide an economical solution to the time delay problem.

The key to improved resolution lay with the improving the array design. The team selected a 2.5 MHz array of 64 individually addressable elements. The number of active elements determined the lateral or azimuthal resolution and their spacing on half-wavelength centers eliminated grating lobe artifacts [15] seen in earlier designs. Packaging the array into a compact handheld unit suited for intercostal imaging proved challenging and necessitated the elimination of several artifacts [16,17]. A Mount Fuji artifact, so called because the top of the sector image was obscured by snow like reverberations that extended into the first several centimeters at the top of the image (as seen on a logarithmic image scale), proved especially challenging. Other issues such as inter-element coupling, element connections, directivity, new materials resulted in an impressive number of journal articles [18-27], significant design improvements, inventions [28-38] and frequent consultations with engineers Jim Fearnside, Jerry Leach, Dave Miller, and Gary Seavey from HP Andover. Their efforts [16] resulted in a short axial pulse with a resolution of about three wavelengths. To further improve out of plane resolution, Dave Wilson designed a rubber elevation lens to focus the beam in the elevation plane (perpendicular to the imaging or azimuth plane).

To be successful, phased arrays required precise time delay adjustment so that the waveforms arriving at N different element locations would arrive in synchronism, ideally resulting in a gain of N. Delay lines available at the time provided only approximate coherence because they came in standard available delay lengths.

Fig. 3 Rick Pering images a water balloon with HP labs system in 1976 (Courtesy of Ed Karrer)

In 1975 Hewlett Packard Labs team (Pering, Maslak and Wilson acting as system architects) [16] came up with a cost-effective solution for the beam former that would provide precise control and greatly reduce the number of delay lines needed; Rick Pering [39-41] invented a tapped summing delay line [39] which replaced many individual delay lines. As described in the patent Sam Maslak assigned to Hewlett Packard [42], it was disclosed that these switchable delays of the tapped delay line were combined with finely tuned delays provided by adjusting mixer phases in a heterodyne intermediate frequency scheme. The imaging system consisted of racks of equipment connected together as shown by Fig. 3. Also Dave Wilson developed a pulsed beam-former simulation program to predict the beam quality. By December 1978, Sam Maslak left the labs to pursue other ultrasound imaging interests at a more urgent pace [43]. In 1981, Amin Hanafy joined Maslak to found the startup ultrasound company, Acuson [44].

For their final design thirty-two central elements were used for transmit and all sixty-four could be receive elements. The lateral round trip (pulse echo) resolution for an equal number of transmitters and receivers is estimated by the equation for the best possible full width half maximum (FWHM) (-6 dB) beamwidth,

$$FWHM=0.886\lambda F/L = 1.36F/(Lf) \quad (1)$$

in which F is focal length (mm), λ is wavelength (λ =1.54(mm/μs)/f (MHz)) and L is the active aperture, L=np, n being the number of active elements and f the transducer center frequency in MHz. . For a well sampled array, L=np=nλ/2, the period p is a half wavelength and

$$FWHM=1.77*F/n \; . \quad (2)$$

For the well sampled array, resolution improves inversely by the number of elements. For the prototype system, the resolution at 12 cm can be estimated as midway between a 32 channel and a 64 channel system or 2.91mm. At 2.5 MHz, the prototype system experimentally achieved a -6 dB resolution at 7 cm in water of 2.5 mm in azimuth, 1 mm in range, and 3.5 mm in elevation [17].

John Larson also developed an idea for adding a linear array by a method he called “tractor treading,” an idea he invented from working on a farm. The scheme involved moving a group of active elements along a linear array by dropping and gaining an element on each end using a switching mechanism and shifting the focusing coefficients along the active elements; see Fig. 2a.

The prototype system block diagram [16,17] shown in Fig. 4 was described in papers co-authored by Karrer, Dias, Larson, Pering, Maslak and Wilson in 1980. A computer (the central processor was an HP 2100 minicomputer) orchestrated the different functions of the system. First, a command initiated a series of transmit pulses, delayed according to a selected focusing depth and steered for each image line, which were sent to the center transmit elements of the array. Second, pulse echoes arriving at array elements along the transmitted line direction were passed through a series of time gain compensation (TGC) amplifiers where they were sent to mixers. Third, depending on the transducer center frequency f_c, a local oscillator frequency f_0 was selected to produce the signals at an intermediate frequency (I.F.). Mixer local oscillators have a common frequency, but each channel had unique mixer phases which changed to support beam steering (line direction) and focusing by the phased array. Fourth, the delay lines needed to achieve the coarse delays were selected as needed to achieve focusing. Fifth, dynamic focusing was achieved by periodically updating the mixer local oscillator phases with depth. Sixth, the outputs of the delayed signals were summed together. Seventh, each beamformed line was then converted to a rectangular brightness modulated image on a cathode ray tube. The transfer of prototype 64 channel system to the HP group in Andover for conversion into a cart-based ultrasound imaging system began in 1977.

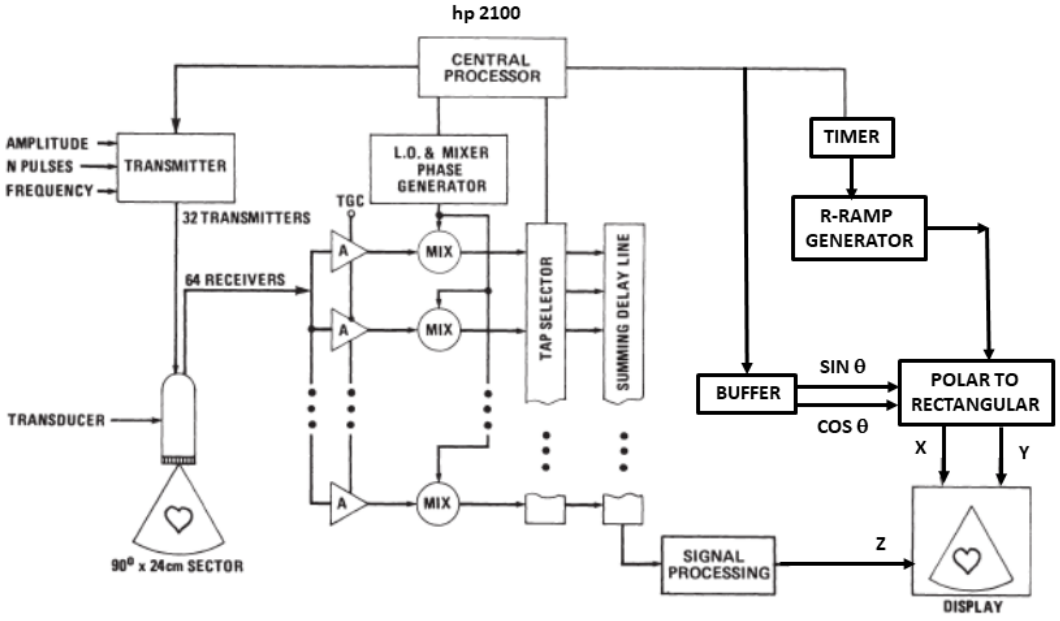

Fig. 4 HP Labs prototype ultrasound imaging system block diagram circa January,1977([16_17] and modified courtesy of Ed Karrer).

III. Hewlett Packard Andover: Designing and Shipping the First Modern Cart-Based Ultrasound System

A small core group led by John Hart, Director of Engineering, and Dave Perozek, the Division Manager, began working on a new ultrasound imaging system in 1976. In the late summer of 1977, Larry Banks, Arthur Dickey, Ray O'Connell, Ron Gatzke, and John Hart spent several weeks at HP Labs learning about the prototype ultrasound imaging system. On their return, they rattled around the new facility at Andover, Massachusetts, largely empty at the time, which was built on farmland on the shore of the Merrimac River. Their goal was not to replicate the prototype system with its racks of equipment, as shown in Fig. 3, but to design a cart-based, mobile, user friendly, real-time cardiac ultrasound imaging system with high image quality. By early 1978, with the first system block diagram, Ron Gatzke worked on analog and mechanical design, Ray O'Connell on scan conversion, Arthur Dickey on digital design, Larry Banks on the system controller, and Jim Fearnside on software. The transducer was supplied by HP labs. Fifteen boards were designed by three people to put together the first working system. The minicomputer was replaced by three fast MC5's, the same microprocessors that were used in the HP Patient Monitoring Center. The complicated routing of instructions was divided into three subsections: the front end (scanner), back end (display) and user interface(controller).

Fig. 5 Hewlett Packard 77020A phased array ultrasound cardiac imaging system with transducer shown on left side (Illustration by courtesy of Koninklijke Philips N.V).

The first imaging session in the late summer of 1978 was a disaster. The system did function as designed but revealed that some of the assumptions made in the original design fell short of what was necessary for a high quality image. "It was like searching for a polar bear in a snowstorm," recalled John Hart. The image was blocky, full of noise and lacked contrast. A deeper understanding of the significance of small quantization errors and better characterization of the transducer response were needed. The team set to work resolving issues and expanded with new hires. An innovation was that the beamformed data was signal processed, envelope detected, digitized and passed to the digital scan converter for eventual display in real time in an interlaced sixty Hz format. One of the major areas for improvement was the digital scan converter. A group consisting of Hugh Larsen, Steve Leavitt and Barry Hunt designed a new scan conversion algorithm, the r-theta converter, which translated the radial geometry of the sector scan (Figure 2B) into the rectangular world of pixels on the display and eliminated the blockiness (quantization) and Moire patterns prevalent on previous ultrasound systems [45]. Ron Gatzke and six others achieved the highest part density in the Medical Products Group on their scanner boards each of which had twelve channels [46]. Jim Fearnside led a group on transducers which initiated investigations on transducer properties and their measurement and others examined ways of improving beamforming and system performance.

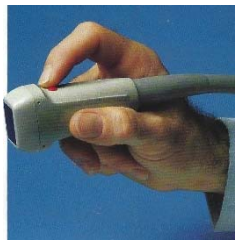

Fig. 6 21200C 2.5 MHz 64 element array (Illustration by courtesy of Koninklijke Philips N.V).

It is not the purpose of this study to document the design of the first Hewlett Packard diagnostic ultrasound imaging system, the HP 77020A, because it has been already described in incredible detail in a series of articles appearing in three issues of the Hewlett Packard Journal [47-49]. Instead, some of the revolutionary features of the system will be emphasized since the HP 77020A became the forerunner of the modern ultrasound imaging systems that were to follow.

As seen in Figure 5, the HP 77020A was mobile, on a cart with wheels unlike its

predecessors which had racks of equipment, and were fixed or sometimes, on wheels. While not the first commercially available phased array system (a 32 channel Varian system was selling at that time); the HP imaging system struck a death knell to most previous ultrasound imaging systems based on mechanical and rotating single transducers. The future for ultrasound imaging became electronically scanned phased and linear arrays. In Figure 6 is a picture of the first HP phased array, the 21200C with sixty-four 2.5 MHz elements which came with the HP 7702A system. Because of its unique heterodyne beamformer architecture, sixty-four channels were able to fit in a cart-based system for the first time. A lightweight, easily maneuverable, handheld 64 element phased array transducer designed for intercostal spacing provided a small footprint (acoustic window) yet a large interior field of view with a sector scan format. The display offered a large smoothly interpolated, highly detailed, high resolution real-time image unlike previous ultrasound imagers, which often offered repurposed oscilloscope monitors or small displays. For transmitting, the 77020A had thirty two newly designed fast transmitters and for receiving, sixty four receive elements with dynamic focusing. Eight time gain compensation controls helped offset absorption and beam losses.

A single central MC5 microprocessor orchestrated the entire system consisting of three subsystems, two of which had their own microprocessor as illustrated by Figure 7, and juggled peripheral devices and the user interface as well. The user interface provided not only acoustic control of transmit focus and scan depth but also the capabilities to annotate and enter patient data by a keyboard, make measurements on the image via a provided joystick, and control video recording of the display in a centralized location. Another innovation was to include M-mode with B-mode in the first multimode display. A key design strategy for Hewlett Packard to grow their customer base was to provide an upgrade path: HP customers were assured that they could always upgrade their system to the latest version.

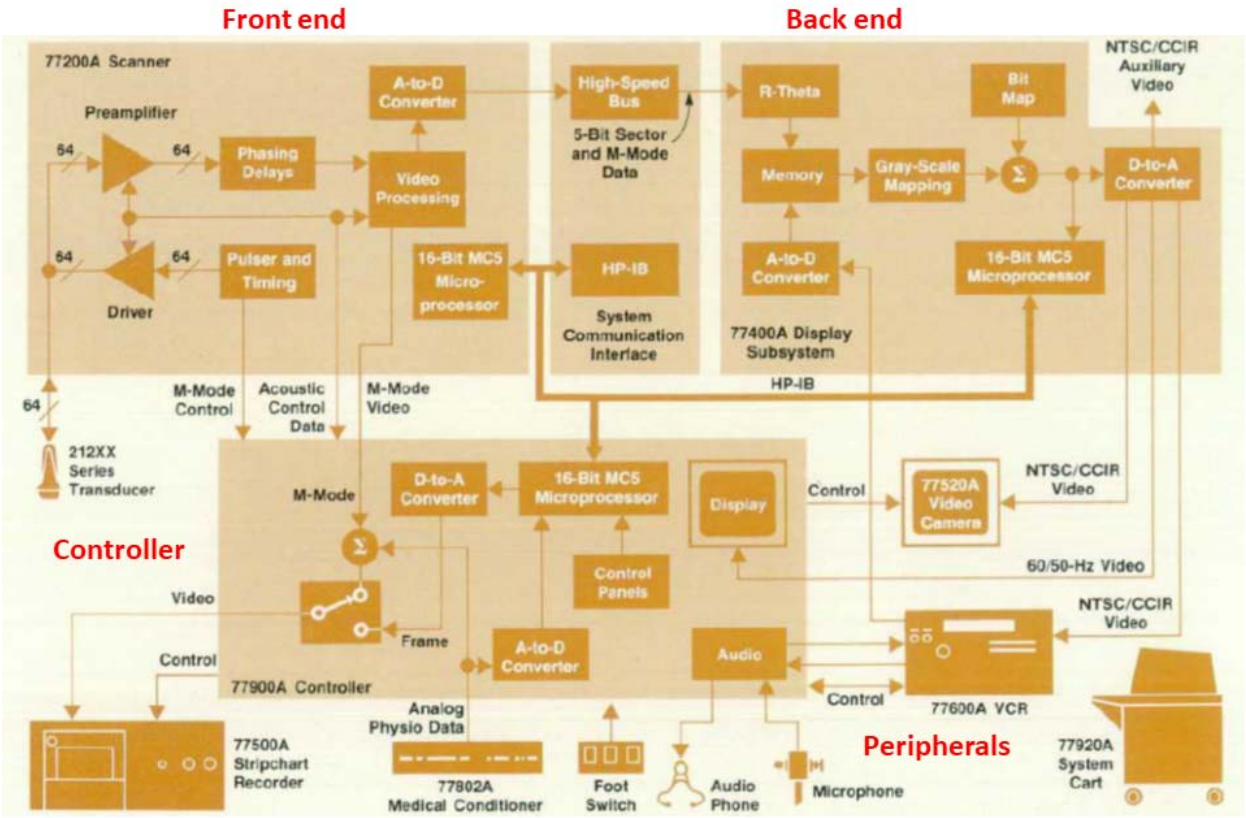

Fig. 7 HP 77020A block diagram showing the front end, back end and controller subsystems (Illustration by courtesy of Koninklijke Philips N.V).

Fig. 7 also illustrates some of the other details of the system architecture. The subsystems were linked by the HP-IB interface bus and software which allowed communication and control throughout the HP 77020A system. The stripchart recorder, a holdover from the Sanborn era, was in a separate cart and eventually deleted.

IV. TRANSDUCERS

Each element in an array can be considered to be an individual transducer. Figure 8 shows the construction of an array. Transducers are piezoelectric devices which send pressure waves into the body on the application of voltage impulses and reciprocally convert returning ultrasound pulse echoes back into electrical signals which eventually are assembled into an image. Ideally, each element should be identical to control beam properties which directly affect image quality. Higher frequency arrays required the fabrication of increasingly small elements repeatably to achieve a period, p, of half a wavelength (recall λ=c/f). Fig. 8 shows array construction.

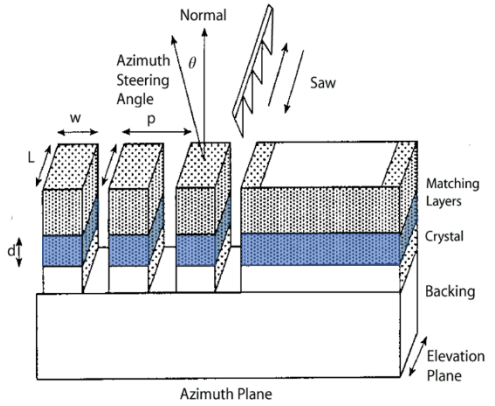

Fig. 8 Array construction showing elements of period p and width w.

Early recognition of the importance of transducers resulted in investments in advancing design through simulation, new materials and extensive measurement and fabrication capabilities. Nearly all arrays were fabricated in house and underwent rigorous measurements and quality checks. Improvements in measurement capabilities both in the engineering and manufacturing laboratories took advantage of the latest Hewlett Packard instrumentation in oscilloscopes, spectrum analyzers, network analyzers, computer and digital memories. Realistic internally developed one dimensional transducer design programs considerably sped up the transducer product cycle [50-52]. Later, commercially available programs were used. Finite element programs such as PZFlex modeled more complicated array effects due to unwanted array construction interactions [53] and were validated by an advanced laser probe measurement system [54]. Improvements in array construction reduced these second order effects and unwanted image artifacts. The key to accurate modeling lay in the development and careful characterization of new materials [55] and piezoelectrics [53] used in arrays. Higher performance design depended on finding materials with specific acoustic and mechanical properties. Experts at the forefront of transducer science were added to an already experienced transducer team: T.R. "Raj" Gururaja, inventor of piezoelectric composites, Jie Chen, an expert on piezoelectric materials and Martha Grewe, an expert on designing matching layers.

The steady cutting edge progression in transducer performance tracked the expansion of the bandwidth and sensitivity of arrays as shown in Figure 9. High sensitivity is related to the transducer's ability to receive weak pulse echoes from deep within the body (penetration depth). Bandwidth is highly valued because the greater the fractional bandwidth (usually expressed as -6dB value)FBW =100*($f_{-6dBhigh}$- $f_{-6dBlow}$)/f_{center}), the more frequencies of operation can be fit into it. The first generation of transducers had a bandwith appropriate for a single imaging frequency (Figure 9A) 30%. The next improvement, wideband arrays expanded the frequency range to include a Doppler frequency and an imaging frequency (Figure 9B), FBW >40%. In the next "dual" generation of transducers, bandwidth accommodated two imaging frequencies FBW>60%. The following transducer family, the "ultraband" group was wide enough to hold an imaging frequency, f_c, as well as its second harmonic, 2 f_c (Figure 9C), FBW>85%. A single crystal transducer (described later) could replace several narrower band transducers (Figure 9D), FBW>100%.

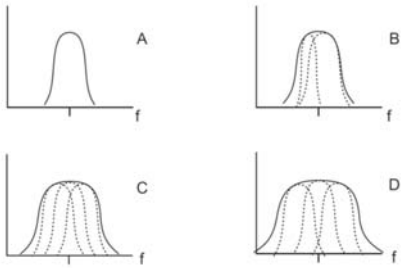

Fig. 9 Progression of increasing bandwidth for HP transducers from 1981 to 1999 (from [8]).

Until June 1990, most of the HP transducer arrays were of the sector type (Figure 2b). At that time, HP introduced the world's first steerable focusing linear array. With a center frequency of 7.5 MHz, the array was guided by a tractor-treading principle in which the 128 channels of the system were connected to the 288 elements of the array through electronic switches. Unlike previous arrays, the image could be steered in the shape of a

parallelogram, resulting in high resolution images of detailed vessel structures and blood flow in unparalleled clarity. In addition, the formats of Figures 2a and 2b could be combined in a new trapezoidal image format: a rectangular format plus two half sectors on each end. As an added bonus, the 21258B operated at 5.5 and 7.5 MHz, and its sister array, the 21255B, at 3.7 and 4.5 MHz. These new arrays were championed by Rick Snyder and a group led by Ray O'Connell, with Matt Mooney and Martha Grewe Wilson designing the transducer [56].

HP expanded its transducer family by designing arrays specialized for different clinical applications. An assortment of different types available in 1999 is illustrated by Fig. 10.

One specialized type of array is the transesophageal probe, a small high frequency phased array which when placed in the esophagus, provides an unobstructed acoustic window in close proximity to the heart, thus bypassing intervening layers of fat and muscle which could degrade transthoracic cardiac ultrasound imaging. Jim Fearnside was an early advocate for transesophageal probes; Linda Carlson took on the challenge of facilitating clinical evaluation of and training for them. The first Hewlett Packard Transesophageal probe (TEE), introduced in 1986, a 5 MHz 64 element array mounted on the end of a gastroscope, allowed high definition views of the heart and blood flow. The gastroscope was necessary to manipulate the probe into a desired position to see selected features of the heart. An improvement, in terms of viewing capability was the TEE biplane which consisted of two orthogonal switchable 5 MHz 48 element arrays became available in 1990.

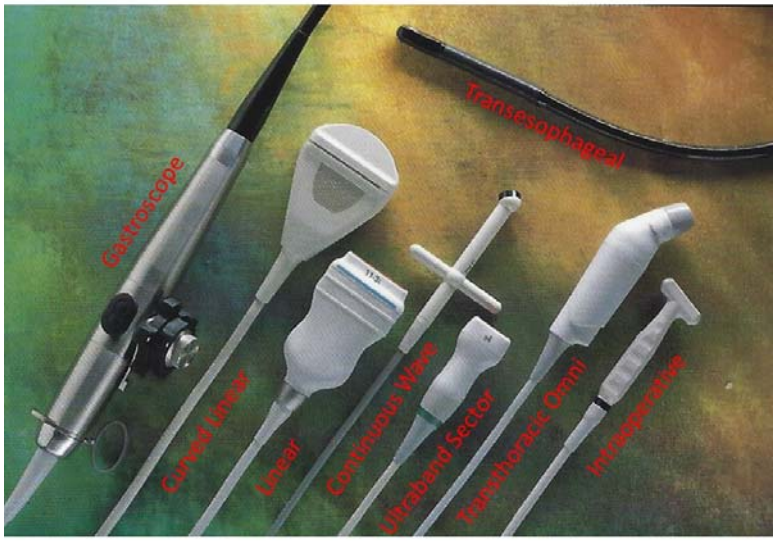

Fig. 10 A sampling of HP transducers circa 1999 (Illustration by courtesy of Koninklijke Philips N.V).

Even though others had thought of the concept of a rotatable TEE [57], it wasn't until Jim Fearnside's team, including Mike Peszynski, designed a multiplane or Omniplane transducer as a practical reality [58-60] at HP. The Omniplane TEE, a 5 MHz 64 element array mounted on the distal end of a gastroscope released in 1992, could be rotated precisely through a motorized drive and new gastroscope. This control capability overcame many of the previous positioning difficulties, provided repeatability and produced spectacular images. Because of the clarity of images, cardiac surgeons and anesthesiologists as well as cardiologists wanted the Omni; it was in great demand [61]. The next version, the Omni 2, which could acquire a sequence of images placed at precise angles, made possible the first three dimensional images of a live beating heart. See Fig. 10. A transthoracic version of the Omni for 3D and small pediatric transesophageal probes followed. Mike Peszynski, inventor on four TEE patents assigned to Hewlett Packard continues to innovate these probes for Philips and holds several dozen patents at last count (see matrix arrays).

What was the ultimate performance possible in a transducer? Had the end of the road been reached? Piezoelectric materials used for ultrasound arrays were polycrystalline ceramics whose initially randomly oriented domains could be mainly but imperfectly aligned in the same direction by a process called "poling," the application of a high voltage. Another type of piezoelectrics, initially investigated by Toshiba [62] for medical ultrasound, are single crystalline materials which offer the potential of nearly perfect domain alignment, resulting in extremely high electromechanical coupling and high applied strains.

In the spirit of finding an ultimate transducer material, T.R. Raj Gururaja and Jie Chen began exploring options. Through Jie Chen's connections with the Shanghai Institute of Ceramics, Chinese Academy of Sciences (SICCA),

which had the capability to grow these types of crystals under extremely exacting conditions, the first sample, the size of a mustard seed, was obtained in 1996. Realizing that there was no program in the United States to systematically investigate single crystalline materials which would be best suited for transducer arrays, Raj led workshops and helped organize a team suitable for a project of this scope. In January, 1998, we submitted, along with Michael Greenstein and Paul Lum, of HP Research Labs, a single crystal research proposal to Wallace Smith, Office of Naval Research [63] which was ultimately funded. A two year program to develop domestic sources to grow large enough samples suitable for arrays and to characterize the basic anisotropic piezoelectric and dielectric properties of the materials was underway. Meanwhile, Jie Chen received a sample from SICCA from which they were able to make the first array. Even though only twenty elements survived, a fractional bandwidth greater than 100% was achieved in 1998. The transducer group opened a commanding competitive lead in finding sources for larger single crystals and the crystal orientations optimal for arrays. Even though this group [64,65] persisted through a reorganization (Hewlett Packard split into Agilent Technologies and Hewlett Packard in 1999) and an acquisition by Philips, their work led to a new generation of transducer arrays with properties that far surpassed all previously made transducer arrays. After seven years more of development by Philips, this new technology became the highly successful generation of transducers branded "PureWave,"[66] and has led to even greater innovations (see matrix arrays).

V Color flow/doppler

It is impossible to capture the many innovations over the twenty year period at HP Andover in a short article, so you will find highlights here rather than a comprehensive history. Eventually, the engineering team grew from 50 in 1983 to 200 in 1999. Leaders of the HP Imaging Systems following John Hart included Paul Magnin, Al Kyle and Cynthia Danhier. Each ultrasound imaging system came out as an upgrade usually identified by a letter. This meant that whatever system a customer had, it was "upgradeable" to the latest version, which required technical and software forethought. Shown in Figure 11 are the upgraded HP imaging systems for the first few years.

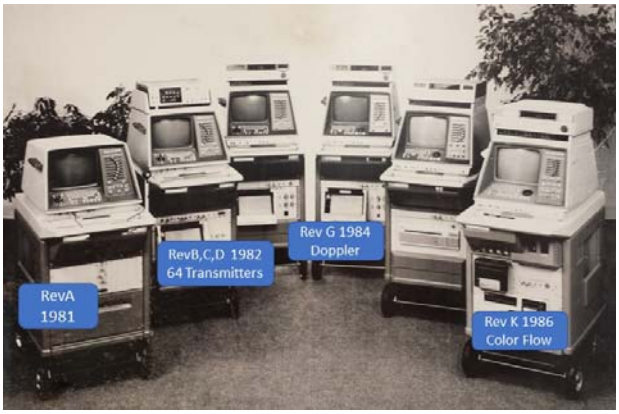

Fig. 11 HP imaging system upgrade path from 1981 to 1986 (Illustration by courtesy of Koninklijke Philips N.V and annotated by Karl Thiele).

One of the most popular upgrades was Rev K, color flow imaging. Barry Hunt related the back stories behind the addition of Doppler and Doppler color flow imaging to the system. In the early nineteen-eighties, the engineering team working on Doppler was falling behind and an outside word class expert was called in to assess the situation. He identified six major problems with their present approach. For a large consulting fee, he offered his services to fix the problems in a short amount of time. John Hart met with the team behind closed doors. He asked them "Should we hire this expert or solve the problems ourselves? You decide." The team unanimously chose to solve the Doppler issues themselves. The learning process was painful, but they redesigned, added synchronization and routed pulsed wave Doppler through the array and separately isolated continuous wave Doppler. The resulting Doppler upgrades, Rev G in 1984 and H and I in 1985 were successful. Once again, detailed descriptions of the Doppler systems appeared in the HP Journal [49].

One aspect of the HP culture (discussed later) was that engineers were encouraged to innovate. While the scanner engineering group was busy with the next step in the evolution of the system (system 2), a new idea emerged. Paul Magnin did a study of color flow algorithms. Based in part on their experience of the Doppler improvements, Barry

Hunt and Dave Hempstead realized that it may be possible to add a color flow capability to the present system architecture. They submitted a proof of principle proposal, System 1 Flow Mapping Proposal, in April 1984. With Ray O'Connell's help , these ideas were posed to management. Steve Leavitt was tasked to manage a small group including Barry Hunt, Dave Lipshutz and Dave Hempstead to do a paper study on implementing Color Flow Mapping. Dave Lipshutz came up with an innovative scheme. Instead, by the summer, the group decided to build a prototype system using separate scan converters synchronized together to represent the red, blue and green colors and they took advantage of the new pulsed Doppler architecture and used an HP 150 controller. By November 1984, they succeeded in making the first color flow image. This skunkworks approach is shown in Figure 12. Afterwards, serious incorporation of color flow mapping into the existing imaging system began. Management gave a green light to the color flow mapping project and postponed work on system 2.Taking advantage of the inherent flexibility of the system architecture and the Doppler improvements, a larger team including Paul Magnin, Ron Gatzke, Karl Thiele, Tomo Hasegawa and Les Halberg among others, was able to fit color flow into the existing cart. A clinical trial with Dr. J. Kisslo at Duke went well. The introduction of the first color flow mapping in North America as the Rev K upgrade in the fall of 1986 was one of the most successful HP ultrasound innovations. These efforts were described in articles and a series of patents [67-70]

Fig. 12 First HP prototype color flow mapping system, Nov. 1984, Insert: color flow image. (Courtesy, Barry Hunt).

VI. Matrix array

The race to obtain better image quality and resolution, was often played out by increasing the number of channels in the system as predicted by Eq. 2. Another fundamental limitation was that this improvement only applied to the azimuth plane which was controlled by electronic control of the focusing and steering of the beam through array elements. The other dimension, orthogonal to the imaging plane, or elevation plane (Figure 8) was typically focused at only one depth by a mechanical lens. The unfortunate consequence of this approach was that strong scatterers lying outside the imaging plane were mapped into the imaging plane. The method needed for perfect three dimensional electronic focusing in both dimensions was well known: a two-dimensional array. The barrier to this approach was that even more elements and system channels were needed: on the order of n-squared. For example, a typical 1D array would have 64 to 128 elements; scaled to a 2D array, this may involve 4000 or 16000 elements!

For many years, the problem of the 2D array seemed insurmountable and compromise solutions, such as sparse arrays and cross arrays, had very high clutter levels and therefore poor image quality [71]. Marty Mason, transducer manager, challenged Bernie Savord to implement 3D imaging electronically. Bernie Savord begrudgingly complied, not thinking it was possible. To his amazement, he found a solution and that it could have been done with technology fifteen years earlier. The breakthrough was to organize the 2D elements in groups, each of which had a set of "microbeamformers" or subarrays for fine delay adjustments and the outputs of which were routed to a summing node [72]. Each summing node corresponded to the usual number of beamformer channels in the system which performed the coarse delay functions. Then, the imaging planes could be stepped across the array. He found the grouping concept in John Larson's earlier patent useful [73]. In 1996, Bernie Savord started on implementations for miniaturizing the microbeamformers to fit within the probe handle [74,75]. One of the challenges of the integration of the large number of microbeamformers was to minimize heat consumption in the handle.

Another major challenge was how to construct the array and connect to thousands of elements, each of which was spaced at about half a wavelength apart. Fortunately, this was a problem that had been worked on for years both at

HP Andover by Rod Solomon [76] and at HP research labs by Michael Greenstein [77-79], so a fitting and practical solution was developed.

In parallel, another effort was needed to transform the acquired array data into three-dimensional visualizations. Arthur Dickey's forethought of this challenge put Karl Thiele in search of a solution. Working together, Karl Thiele and Bernie Savord put together the first working prototype shown in Figure 13. One of the two connectors in Figure 13A was for the analog signals that plugged into the imaging system, and the other was used for the signals needed to program the probe. By 2001, at the American Society of Echocardiography annual meeting, they put on a private demonstration of the matrix array prototype shown in Figure 13.B and real-time 3D imaging to enthusiastic responses. After considerable work by a Philips team, the 2.5 MHz X4 matrix array was shipped as a product in 2002, along with a new system to run it, the SONOS 7500. Not only did this transducer have 2880 elements, but all the microbeamformer electronics also fit in a handle not much bigger than the first 64 element HP phased array (Figure 6) and they served the function of hundreds of front end boards. Later the single crystal transducer design (Pure Wave) was incorporated into a matrix array in the X7-2 with 2500 elements. Similarly, matrix array transesophageal probes with thousands of elements became reality [80]. At the latest count as of this writing, the X-6 matrix array has over 9000 elements [81].

Fig. 13 Matrix arrays with microbeamformers in the handle: A.(left) first working prototype B. (right) close to final product (Courtesy of Karl Thiele).

VII. OPTIGO

In 1998, a small group began working on the SPUD (Small Portable Utrasound Device) imaging system based on novel concepts. The team included Jim Fearnside, Rachel Kinicki, Joe Fallon, Mike Anthony, Charles Dowdell, Ted Fazioli, Matt Mooney and Steve Leavitt. The goal was to cram an entire imaging system in a size smaller than a laptop and it would be battery-operated, low cost and light weight. The design centered around application specific integrated circuits(ASIC's). Steve Leavitt developed one which included both scan conversion and color flow mapping. The team also created beamformer ASIC's in which there was a processing channel for each element in a phased array transducer . Each beamformer ASIC replaced several scanner boards of earlier full sized imaging systems. Also, other formats such as linear, curved linear and combination formats could be accommodated. The TGC controls were automated and the user-interface consisted of intuitive simple controls. The result was a reconfigurable and scalable ultrasonic imaging system which could adapt to different types of transducers. Output could be stored on a memory card and the display was a 6.5" LCD screen. In addition, different operating modes, processing algorithms and display options were under software control [82]. Even greater miniaturization was achieved by using the microbeamformer principles used for the matrix arrays. Here, for these one dimensional arrays, sub-beamformers, as shown in Figure14, could be put right in the probe handle, aiding the further miniaturization of the overall device. The device appeared as a Philips product, the OptiGo in 2002 aimed initially at the cardiac screening system. [83], (see Figure15). The challenge of how to mass-market a low cost ultrasound product. (about ten thousand dollars) delayed the release of the OptiGo. The sales problem was that incentive was always greater for selling a regular sized system with a larger profit margin than that for an OptiGo . A second problem was the lack of a hard copy output needed for medical billing. The beamformer in the handle foreshadowed the Philips' Lumify in which, after considerably more development and miniaturization, the front end and entire

beamformer and processing were fit into the transducer handle by November 2015 and connected to a smartphone for display and communication to the web.[84].

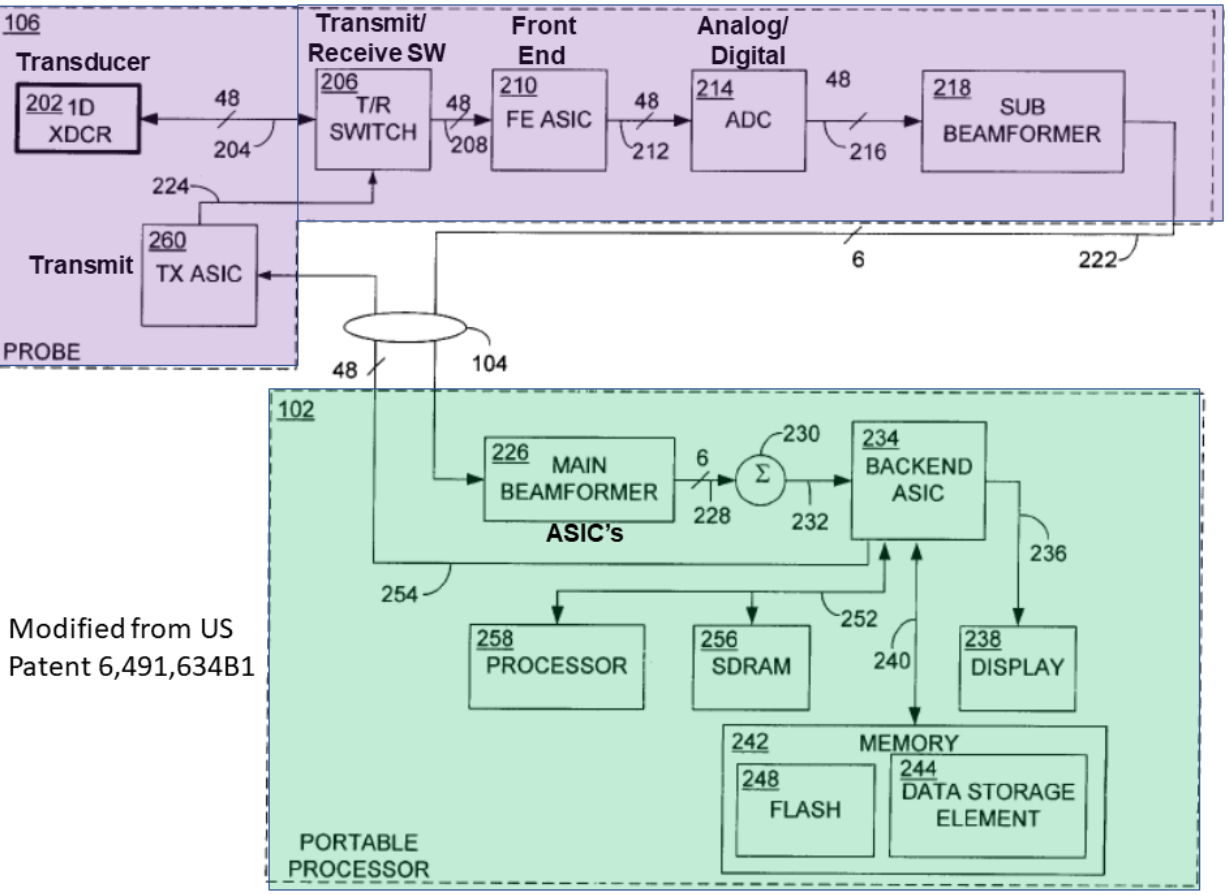

Fig. 14 Block diagram for portable ultrasound imaging system with sub-beamformers in probe handle (From US Patent 6,491,634B1).

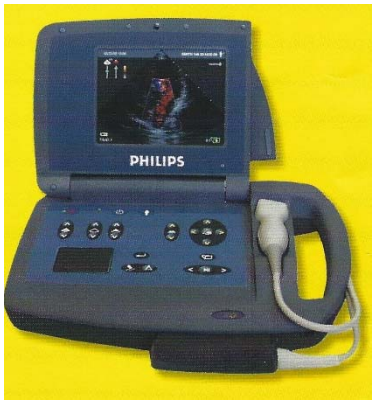

Fig. 15 OptiGo portable low cost ultrasound imaging system released by Philips in 2002 (Illustration by courtesy of Koninklijke Philips N.V).

VIII. RESEARCH SYSTEM (PAWS)

In 1992, Steve Leavitt floated the idea of a research ultrasound imaging system to management. His concept was to digitize the radiofrequency inputs of each of the transducer elements and store them in a digital memory for prototyping future systems, new beamforming and signal processing schemes. Management agreed and Steve Leavitt led a group including Dave Lipshutz, Ron Gatzke, Ban Dinh and Bernie Savord to design the system. As shown in Figure 16, the main processor of the Phased Array Work Station (PAWS) was an expensive Convex-240 (240Megaflops) supercomputer which required massive electrical cabling which Ban Dinh installed. A software acquisition controller managed the setting of a SONOS1500 front end the analog output of which was routed into the PAWS cart which included 10 bit 20 MHz analog to digital conversion boards for digitizing the signals from each of the 128 transducer elements. The output of these boards was fed in parallel simultaneously into a bank of random access memory units which in turn streamed their output into the Convex supercomputer which then organized the data into long term digital storage. The Convex then ran algorithms which processed the data and formed images.

The PAWS system was instrumental for trying out different approaches for the future digital system design, nicknamed "Titan," which eventually became the SONOS 5500, released in 1997. PAWS was modified to try out new arrays such as electrostrictors. Steve Leavitt led the Advanced Projects Group for 1992 to 1995 after which it was led by Jim Fearnside. Don Orfino wrote software in MATLAB which emulated the standard delay and sum beamformer, filtering, back end and display processing and controls. He later left and joined Mathworks which produces MATLAB. The present author explored 1.5D electrostrictive arrays (intermediate between 1D and 2D arrays [85]) on the system and the effects of aberration, based on a joint project with Prof. R. Waag at the University of Rochester for obtaining acoustic chest wall data [86]. He also investigated applying seismic algorithms with Dan Burns, a consultant. The seismic processing eliminated much of the clutter in cardiac images but, took too long to run, even on a supercomputer [87].

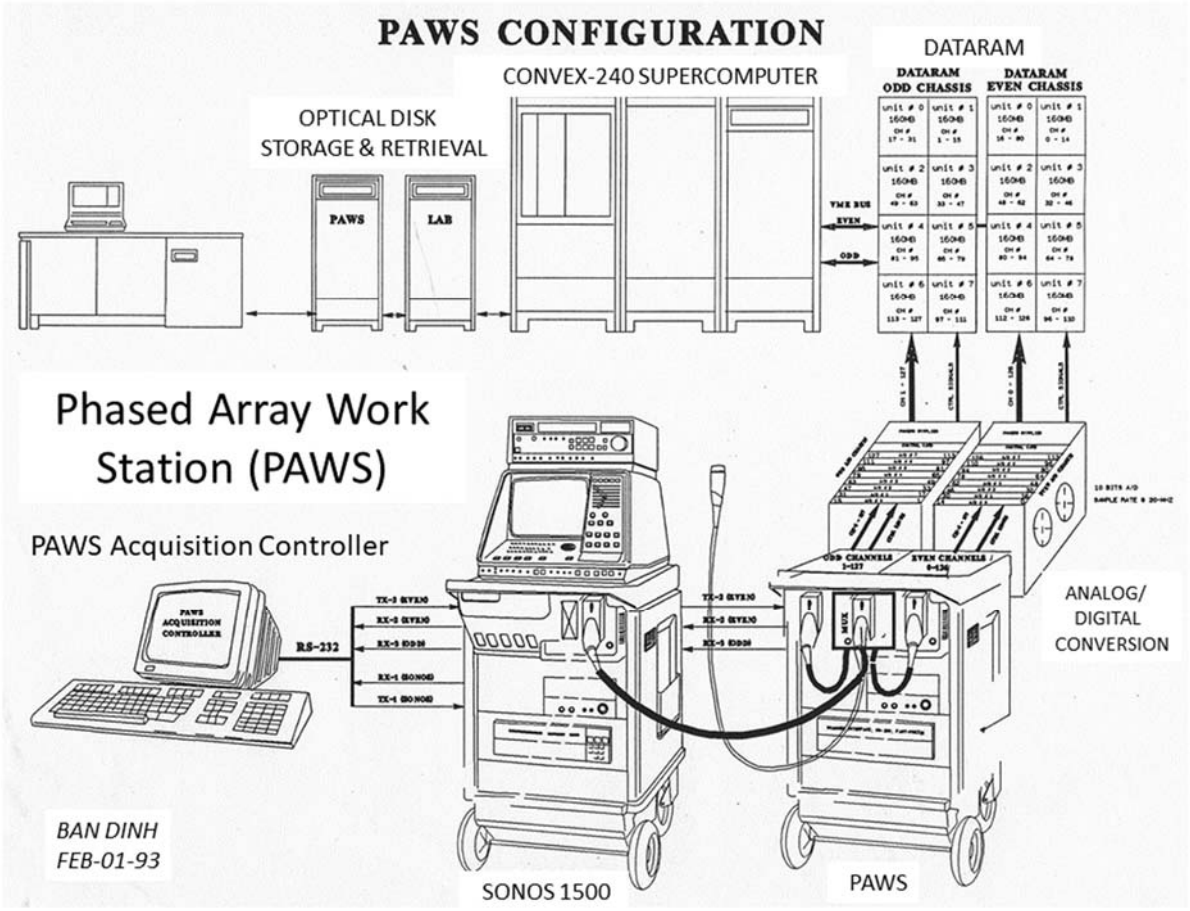

Fig. 16 PAWS system (Courtesy of Ban Dinh).

After about five years, Moore's law caught up with the Convex supercomputer. Hewlett Packard's state of the art workstations had evolved to be as powerful as a 1992 Convex-240. By then, too, the SONOS 5500 digital imaging system had become both a product and an experimental base for testing new imaging ideas. In an ironic twist, the next generation of Convex computers were based on HP RISC (Reduced Instruction Set Computer) microprocessors (used in workstations) running in parallel. In 1995, Hewlett Packard acquired Convex. PAWS was dismantled and donated to charity. The concept of PAWS, then protected by HP proprietary restrictions, reemerged in present day ultrasound research systems which consist of a software controlled front end, high speed digitization and storage and imaging system simulation in software.

IX. HP vs. ACUSON

Sam Maslak left HP labs in November, 1978 after assigning the mixer imaging system architecture patent [42] to HP. By 1981, Maslak and Robert Younge, a colleague at Hewlett-Packard, joined with Amin Hanafy, a key transducer expert at HP medical products and HP labs, to form Acuson in 1981. They were able to raise sufficient venture capital based on their business plan [88,89] by January 1982. Based on the assumption that the plan was executed, a primary emphasis was the radiology market, including obstetrics. The obstetrics segment, which propelled ultrasound imaging into existence [90], continues to be one of the largest ultrasound imaging market segments. Acuson's strategy was to provide a premium image, called the "gold standard" at a premium price. In ten months, a prototype system capable of producing either sector or linear format images passed its first water tank trial [91]. In 1983, Hugh Larsen left HP Medical Products to join the Acuson team. By the end of 1983, Acuson shipped its first thirty systems [92]. How was it possible for Acuson to produce a quality system so quickly? Sam Maslak offered advice for startups: address (and eliminate) technical risk [92] first. From the outset, he was keenly aware of beating the competition [93]. To compare major HP and Acuson events, a timeline is provided by Figure 17.

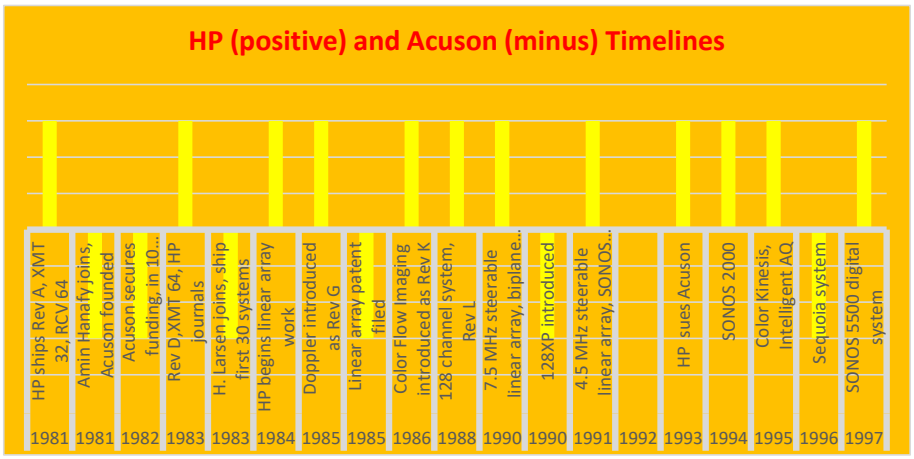

Fig. 17 Timelines for HP (positive bars) and Acuson (negative bars) product and other events.

Sam Maslak explained how Acuson was able to produce high quality imaging: "Computed Sonography [94]." He emphasized that a hybrid analog/digital computer in the Acuson system managed the imaging parameters. By comparison, HP had a scanner microcomputer for controlling beam characteristics as well as an overall system controller microcomputer (Figure 7). Resolution is mainly controlled by the number of elements, according to Eq. 2. Acuson had 128 elements at its first launch compared to HP's 64. HP did not catch up to 128 elements until 1988 with Rev L. Contrast was another factor Maslak identified which meant low sidelobes obtained through apodizing or weighting the output of the elements [95]. HP was aware of this as well; see Figure 2 in [16], a 1980 paper describing the HP system to which Maslak was a coauthor. The last factor identified in [94] is image uniformity obtained by using a "tracking lens," in the Acuson system as well as by keeping the F number (=F/L) constant (see Eq. 1) in the "tracking lens" with depth. In the HP systems as well as the HP labs prototype an equivalent process was used called "dynamic focusing."Acuson advertised "upgradeability" as a key feature of their systems. By the time Acuson introduced its first system, HP was on its third upgrade, Rev. D.

Relations between HP and Acuson remained cordial until Acuson, which had spectacular growth and success in the radiology and obstetrics markets, decided to enter the echocardiology market, in which HP was the dominant leader. Once this entry happened, Acuson and HP were on a collision course. To recap some of the highlights of HP

development on the timeline, previously described, 1986 saw the introduction of color flow, and in 1990 and 1991, the entry of the first steerable linear arrays. Meanwhile, Hugh Larsen and Sam Maslak filed a strategic patent on behalf of Acuson in November 1985 on linear arrays which was granted in October 1987 [96]. The curious aspect of this patent was that it was entirely dependent on Maslak's mixer patent assigned to HP during his HP labs days; .ie. it could not be implemented unless the company had a license to use the earlier patent [42]. In 1993, HP had strong evidence that Acuson was infringing on their mixer patent [42] and filed a law suit against Acuson [97]. Acuson countersued, claiming that HP had infringed on their linear array patent [96]. By then, HP had two linear array products on the market; it had been using the tractor treading concept that traced back to John Larson, who did not file a patent based on HP interests at the time. Even though the results of negotiations between the two parties were not made public, the fact that HP continued selling linear arrays and Acuson still had their systems indicated a truce was reached.

By 1993, both companies were more interested in their futures. Acuson invested heavily in their new Sequoia platform [88], so much so, that their profits flattened and they began to lose their market share. When Sequoia debuted in 1996, it followed a previous Acuson trajectory of super premium systems that outperformed others and was also, by far, the world's most expensive ultrasound system. Acuson's premise that premium systems create their own market no longer held in changing economic times [88]. Competitors had also produced systems with excellent images and cost consciousness had set in. Meanwhile HP launched their premium digital system, the SONOS 5500 in 1997. In the echocardiology market segment, by 1983, HP quickly rose to the top and stayed there as Figure 18 shows.

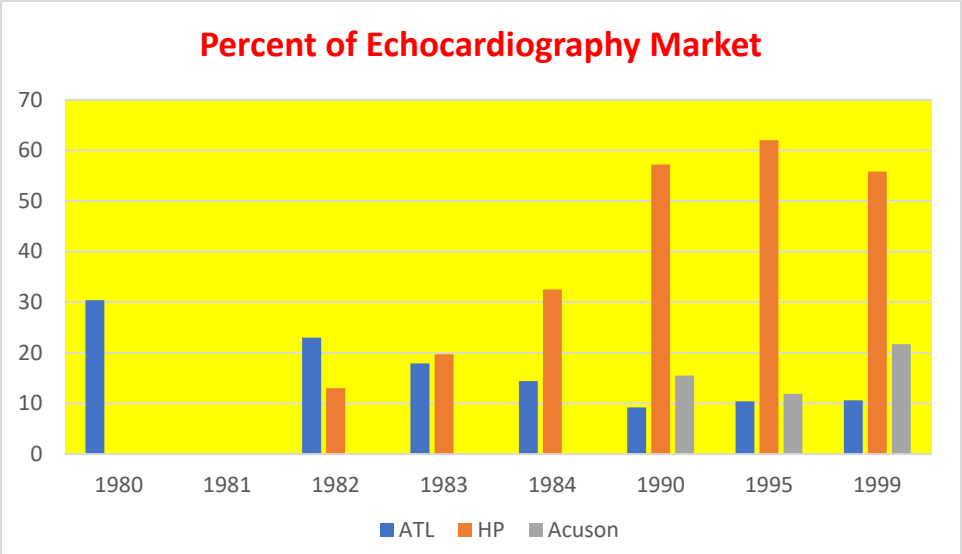

Fig. 18 Percentage of total echocardiography market for ATL , HP and Acuson by year (Courtesy of Harvey Klein).

Even greater changes were in store for diagnostic ultrasound. In 1980, before HP's entry, twelve companies shared most of a total ultrasound imaging market of two hundred and seven million dollars [98]. By 1998, three companies held 48% of the overall diagnostic ultrasound two and a half billion dollar market: ATL 18%, HP 16% and Acuson14% [98]. In a surprise move, Philips acquired ATL in late 1998. Due to internal squabbles, Hewlett Packard split into Hewlett Packard (computers, printers, etc.) and Agilent Technologies (instrumentation, electronics) in November 1999. At the time of the split, Hewlett Packard had grown to a forty-two billion dollar company employing eighty-four thousand people worldwide. Recall that Hewlett Packard entered the medical business by acquiring Sanborn and it joined the New York stock exchange in 1961. Siemens acquired Acuson in September 2000 for seven hundred million dollars and Sam Maslak stepped down as CEO. In November 2000, Philips, which had a previous relation with HP, acquired the health care group (including ultrasound) from Agilent Technologies. The CEO of Agilent remarked "We concluded we had higher priorities to invest in and health care came in the bottom of the list."[99]. Today through acquisition of the top ultrasound companies, Philips Health care dominates the health care market and ultrasound imaging and benefitted from Hewlett Packard's legacy of innovation. Now only three large companies lead the medical imaging market: Philips, General Electric and Siemens.

X. HP WAY

The work environment at HP Andover inspired innovation and teamwork. The HP Way evolved from the combined experiences of Bill Hewlett and Dave Packard from decades of company growth [100]. An open, collegiate atmosphere encouraged individual motivation, initiative, creativity, freedom and trust. Consider that this whole ultrasound imaging enterprise began because of an enthusiastic engineer and a physicist at Hewlett Packard Laboratories, not from a top-down directive.

An informal, decentralized management style prevailed based on "management by objective" which allowed some flexibility in how team players could best achieve common purposes. "Management By Walking Around," encouraged open communication between managers and their team. Because Hewlett and Packard realized that company growth depended on a continual influx of new product ideas, engineers were encouraged to innovate and supply stocks remained open so they could try out new ideas. In the beginning, there were two coffee breaks with free doughnuts which led to learning about other projects and forging new collaborations. There were also beer busts, picnics, and community contribution activities. HP tended to hire well qualified people and keep them which led to company loyalty. During downturns in the economy, everyone took a 10% cut (working 9 days out of 10) until the economy recovered. Employees were eligible to participate in profit sharing in the form of buying Hewlett Packard stock at subsidized rates. Also there was a commitment to produce quality high performance products which resulted in greater customer satisfaction.

At its best, the HP Way worked well. In this article are examples of trusting in individuals to make decisions and to recover from failure, ideas arising from within, teamwork required for large complex projects, managers challenging others to do the seemingly impossible, and an overall commitment to excellence.

ACKNOWLEDGEMENTS

In an article of this length, regrettably, I was not able to include many of the people and achievements that occurred at HP Andover. Even though technology was highlighted, other parts of the team such as marketing and manufacturing also played important roles. In particular, marketing's enthusiastic engagement propelled the success of the ultrasound imaging enterprise. Manufacturing's partnership with engineering assured that high quality, reliable and easily reproduceable products found their way to customers worldwide. I am grateful to my former colleagues at Andover and at the former research labs who spent many hours explaining and recalling events for this article. In particular, I would like to thank Linda Bogdanoff, Linda Carlson, Jie Chen, Ron Gatzke, Martha Grewe Wilson, Kurt Guggenberger, Raj Gururaja, Ed Karrer, Barry Hunt, John Larson, Al Kyle, Steve Leavitt, Paul Magnin, Jim Mniece, Mike Peszynski, Bernie Savord, Rick Snyder, and Karl Thiele. Special thanks to Harvey Klein for digging up archival market data. The author takes responsibility for any errors and omissions.

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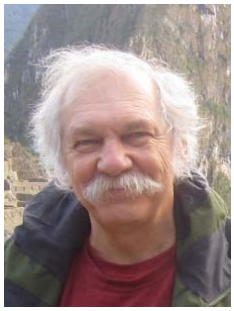

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HISTORY OF DOPPLER ULTRASOUND

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I. HISTORICAL BACKGROUND

Doppler ultrasound involves measurement of the velocities of blood or tissues by the Doppler principle. The Doppler effect concerns the change in perceived frequency by an observer of a wave as a result of motion of either the source or observer or both. This effect, which was subsequently named after him, was proposed by the Austrian physicist Johann Christian Doppler in 1842 [1], to explain the colour of binary stars. The now classic Doppler equation was deduced:

$$f_o = f_s \left(\frac{c-v_o}{c-v_s}\right) \quad (1)$$

where f_o is the frequency perceived by the observer, f_s is the frequency emitted by the source, c is the wave speed in a stationary medium, v_o is the vector velocity of the observer and v_s the vector velocity of the source.

In fact the effect proposed by Christian Doppler was too small to explain the colour of stars. However the effect was shown to be valid for sound by the Dutch physicist Christophorus Buys-Ballot in 1845. His experiment involved a group of musicians playing on a moving train, with stationary observers on the ground as the train passed. Subsequent work has confirmed that the Doppler effect is valid for any wave, including sound, electromagnetic (radio, microwave, light etc.) and gravitational waves. Figure 1 illustrates the Doppler effect. Further reading of Christian Doppler and early work on the Doppler effect may be found in review articles [2-3].

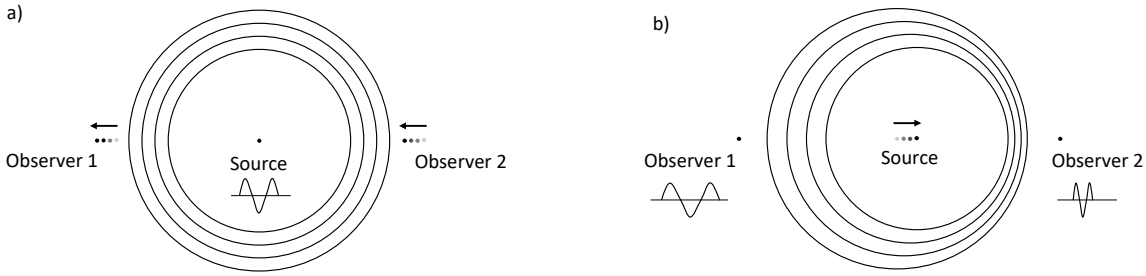

Fig. 1. Doppler effect; there is relative movement between the source and the 2 observers. a) Wave motion with respect to the source. The wave propagates symmetrically in all directions at a frequency *f*. b) Wave motion with respect to the observers. Wave propagation is asymmetric; there is contraction of the wave as perceived by Observer 2 who perceives a higher frequency (f_2>f); for Observer 1 there is dilation of the wave who perceives a lower frequency (f_1<f).

II. EARLY DOPPLER ULTRASOUND

The first medical investigations using the Doppler effect were undertaken by Shigeo Satomura from the Institute of Scientific and Industrial Research of Osaka University in Japan. Working in the area of industrial radar and ultrasound he was encouraged by his supervisor Kinjiro Okabe to investigate medical applications. Satomura's first studies were actually not on blood flow, but on cardiac motion, and therefore the first Doppler ultrasound paper is also the first Tissue Doppler paper. The paper published in 1956 in Japanese was titled 'A new method of the mechanical vibration measurement and its application' [4] A follow-up paper was published in English in 1957 [5]. The Doppler device used a 3 MHz continuous wave probe with a central transmit element surrounded by a ring shaped receive element. The receive signal was demodulated and band-pass filtered from 500-1500 Hz. The Doppler signal was displayed as an amplitude signal along with the ECG and cardiac sounds (Fig. 2). These 2 papers show for the first time the now widely used Doppler equation:

$$f_d = 2\frac{v_o}{\lambda} \quad (2)$$

where f_d is the Doppler frequency, v_o is the velocity component along the ultrasound beam and λ is the wavelength.

Noting that:

$$c = f\lambda \quad (3)$$

where f is the transmit frequency and c is the speed of sound, equation (2) can be rearranged to give the modern version of the Doppler equation as applied to medical ultrasound, assuming that transmit and receive beams are aligned:

$$f_d = \frac{2fv\cos\theta}{c} \quad (4)$$

where θ is the angle between the beam and the direction of motion.

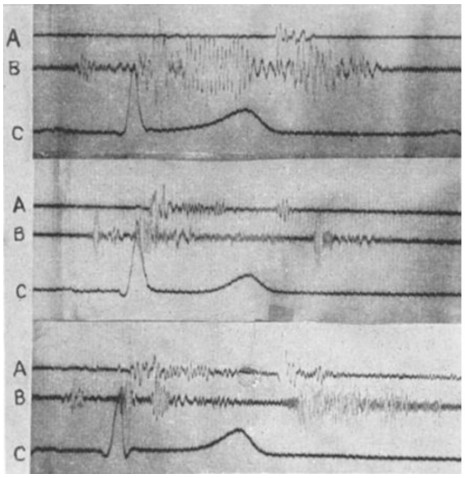

Fig. 2. Oscillograms obtained by Satomura (1957) from the heart. A: Heart sounds, B: Doppler signal, C: ECG. Reproduced from ; Satomura S. Ultrasonic Doppler method for the inspection of cardiac functions. J Acoust Soc Am1957;29:1181-1185, with permission of the Acoustical Society of America.

In a review paper [7] on early Doppler development it was noted that Satomura and his colleagues identified 2 types of Doppler signal from the heart; those with frequencies below 500 Hz which were thought to arise from heart wall motion and one at 1000 Hz which was thought to arise from valve motion. The first paper reporting detection of blood flow was published by Satomura in a 1959 paper titled 'Study of the flow patterns in peripheral arteries by ultrasonics' [6]. Doppler signals were obtained from water flowing in a tube, and also for flow in the brachial artery and vein (Fig. 3).

The 2 papers by Satomura in 1956 and 1959 represent the first studies of Doppler ultrasound in humans. Satomura went on to develop the 'Ultrasonic Blood Rheograph'. The Rheograph was the first commercial ultrasonic Doppler flowmeter, manufactured by the Nippon Electric Company (NEC) and available from 1959 (Fig. 4). Tragically Shigeo Satomura died of a subarachnoid haemorrhage in April 1960. The work on the Rheograph system was presented by his colleague Ziro Kaneko at the Third International Conference on Medical Electronics in the same year [8].

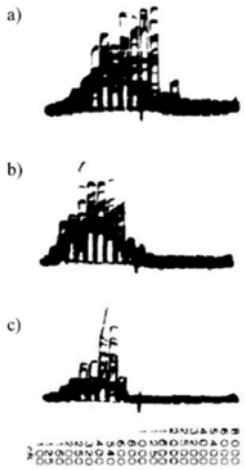

Fig. 3. Frequency spectra obtained from: a) brachial artery systole, b) brachial artery diastole, c) brachial vein. Reproduced from Ultrasound Med Biol, Vol. no. 12, Kaneko Z, First steps in the development of the Doppler flowmeter, pp. 187-195, copyright Elsevier (1986).

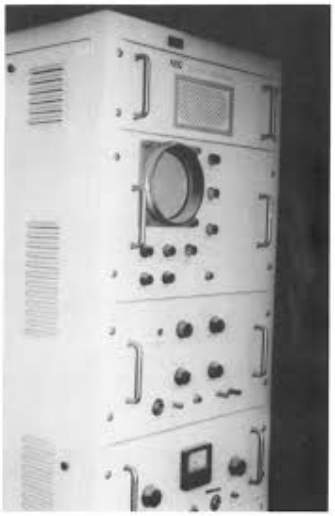

Fig. 4. First commercial Doppler ultrasound system; the 'Blood Rheograph' available in 1959 and developed by the Nippon Electric Company (Japan). Reproduced from Ultrasound Med Biol, Vol. no. 12, Kaneko Z, First steps in the development of the Doppler flowmeter, pp. 187-195, copyright Elsevier (1986).

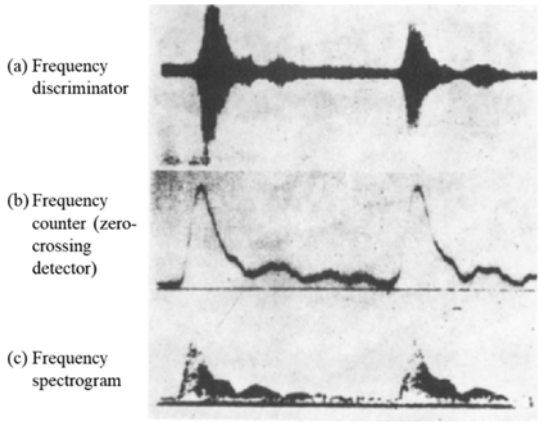

Fig. 5. Blood flow patterns in the brachial artery registered by 3 methods. Reproduced from Ultrasound Med Biol, Vol. no. 12, Kaneko Z, First steps in the development of the Doppler flowmeter, pp. 187-195, copyright Elsevier (1986).

Satomura's work was carried on by colleagues and a number of developments were made. There was the realisation that the Doppler signal from blood did not arise from turbulence but from red cells, and that the Doppler frequency was correlated with blood velocity [9]. It was recognised that the recording system was a critical component and several different methods were developed (Fig. 5). A frequency discriminator recorded voltages below a set value of 1000 or 2000 Hz. Use of a zero-crossing detector was also attempted [10]. This method, which became popular in early Doppler, was dismissed by the Osaka group as it suffered from interference from low frequency noises present in clinical studies. The final and preferred method was spectral display consisting of a Doppler frequency - time trace [11]. A method of Doppler detection was developed which allowed the Doppler signals arising from forward and reverse flow to be separated [12].

The paper by Kaneko [7] provides more details of the early development of the Doppler flowmeter.

III. Continuous wave (CW) doppler

The first CW Doppler systems were 'blind' in that there was no accompanying B-mode image. The transducers were typically designed in a split-D format with adjacent transmit and receive elements or followed the Satomura approach of a central circular transmit element with a surrounding receive ring element.

Compact non-directional CW Doppler systems were described by the group from the University of Washington in Seattle [13-16]. This followed from work by the same group on the development of invasive probes for measurement of flow in arteries [17-18]. The Doppler output was in the form of an audible signal generated from a zero-crossing detector where hardcopy recordings of the Doppler trace could be made through connection to a chart recorder. Recordings were made of flow in arteries in the upper and lower limbs, carotid arteries and aorta.

In addition the first waveforms during pregnancy were recorded, from the uterine artery and vein, and from the fetus; figure 8 in the 1966 paper by Rushmer [15] identifies 'fetal flow' which looks to be from the fetal aorta.

The compact CW Doppler described in 1966 by Rushmer [15] was the forerunner of the 'pocket Doppler'. It soon became apparent that there was a wealth of information in the audible signal which the operator could use to identify normal from abnormal flow in disease. The small size, low cost and clinical utility helped to spread pocket Doppler in hospitals. Over 50 years later the design of pocket Doppler is virtually unchanged from that described by Rushmer et al; continuous-wave non-directional with audio output and can easily fit in a coat pocket.

Fig. 6. Doptone CW Doppler system manufactured by the Smith-Kline Instrument Company (Philadelphia, USA). The system was the first fetal heart monitor. The version illustrated was used in peripheral vascular applications. Reproduced with permission from the British Medical Ultrasound Society.

Recordings of Doppler from the fetal heart were reported by Callaghan in 1964 [19] and Johnson et al. in 1965 [14]. The CW Doppler system developed by the Seattle group was commercialised by Smith-Kline Instrument Company (Philadelphia, USA) as the 'Doptone' in 1965 (Fig. 6), which was used for fetal heart detection [20] and for applications in the peripheral vascular system [21].

Subsequent developments paralleled work done in Japan. Directional detection was developed by McLeod in 1967 [22] where audio signals from forward and reverse flow were available as separate audio channels, e.g. using stereo headphones. Single-line display was developed using a zero-crossing detector, and later spectral Doppler display was implemented [23]. The latter involved recording the Doppler signals on magnetic tapes and sending them to Northrop Nortronics (Needham Heights, USA) and waiting 6 weeks for the results [24].

The basis of the original quadrature detector developed by McLeod involved splitting the Doppler signal into 2 paths, phase shifting one channel by 90º to create a direct (D) and a quadrature (Q) signals, and comparing the phase lag between the D and Q signals. Depending on the lag the signal could be switched to either the forward or reverse channel. This system suffered from switching artefacts and for flow in which there was simultaneous forward and reverse flow the flow direction could not be resolved. Later developments overcame these limitations and are described by Coghlan and Taylor in 1976 [25].

The zero crossing detector was widely used in early Doppler systems. The detector in its simplest form produces a signal every time the display goes from negative to positive. With no noise and no offset the output from the zero-crossing detector should be equal to the mean (RMS) Doppler frequency [26]. Noise produces a large number of false crossings, and in practice an offset threshold is implemented using a set-reset procedure; that is a trigger is set if the amplitude exceeds a positive threshold value, and the trigger is reset when the amplitude exceeds a negative threshold value. In this way noise has limited effect and the system is able to provide an output proportional to frequency [27]. Single-line Doppler displays were phased out following the introduction of real-time spectral display as described in the next paragraph.

Early Doppler spectral analysis was performed off-line, commonly using a swept-filter system. Sound spectrographs were developed during World War 2 to help analyse enemy messages [28]. Subsequently sound spectrographs were used in speech therapy and in the recording of bird songs. In the Kay Spectrograph (Kay Electric Company, Pine Brook, USA), charge sensitive paper was attached to a rotating drum (Fig. 7). As the drum rotated the Doppler signal was filtered by a narrow band filter which scanned the frequency range increasing from negative to positive frequencies. The pen touched the paper when there was signal. A two second spectrograph was produced in around 2 minutes. The advent of real-time spectrum analysis [29-33] allowed this feature to be incorporated into commercial systems. The availability of Doppler spectral data opened up the field of Doppler waveform analysis which is discussed in the next section.

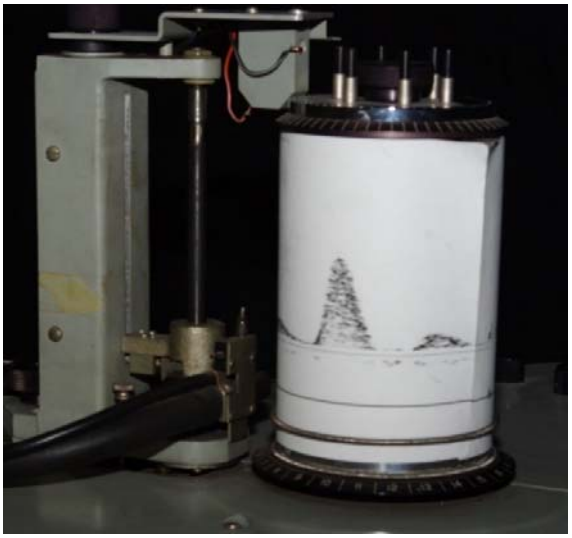

Fig. 7. Spectral Doppler tracing made by the Kay Spectrograph (Kay Electric Company, Pine Brook, USA). The tracing is in-place on the paper attached to the recording drum of the machine. Reproduced with permission from the British Medical Ultrasound Society.

In peripheral arteries CW Doppler was used to evaluate the extent and location of arterial disease. Local increases in Doppler frequency occurring as result of stenosis could easily be observed by tracking the transducer along the artery. Peripheral arteries are approximately parallel to the surface, so even though the exact beam-vessel angle is unknown provided that a similar beam-skin angle is adopted the Doppler frequency data could be compared between patients. Spencer and Reid in 1972 [34] demonstrated the increase in maximum frequency with degree of stenosis in carotid arteries. Most quantification involving flow waveforms from CW Doppler has involved quantities related to waveform shape as discussed in section IV.

In cardiology stand-alone CW Doppler was used to measure the flow waveform from the ascending aorta, as an indicator of cardiac output [35]. The transducer was placed on the suprasternal notch and angled down and to the left; the blood flow in the aorta was approximately parallel to the beam axis at this position. The device was subsequently commercially marketed as the 'Transcutaneous Aortovelograph' (Muirhead Medical Ltd, Beckenham, Kent, UK).

CW Doppler had limited clinical impact in cardiology [36], however Feigenbaum [37] notes 'The breakthrough came when Holen then Hatle demonstrated that haemodynamic data could be accurately determined with Doppler ultrasound'. Both groups used CW Doppler to acquire velocity data from which pressure gradient was calculated [38-41], further discussed in section IX. Gradually stand- alone CW Doppler systems became redundant with the widespread availability of duplex then colour flow Doppler systems.

Further reading on the history of CW Doppler may be found in review articles [42-45].

IV. Waveform analysis

The absence of B-mode imaging meant that stand-alone CW Doppler systems were mainly used in arteries with defined locations and/or defined waveform shapes. Early applications were therefore in the lower and upper limbs and in extra-cranial arteries. The abdomen and fetus were difficult as there are many arteries with similar waveform shapes and there was no easy way of distinguishing from which vessel the Doppler signal arose.

The other limiting feature of stand-alone CW Doppler is the lack of knowledge of the angle θ between the beam and direction of motion. Conversion from Doppler frequency shift to velocity, commonly practiced using duplex Doppler (see below), requires knowledge of the angle θ so is not possible using stand-alone CW Doppler. Quantification of the Doppler waveforms from stand-alone CW Doppler therefore relied on indices related to the Doppler frequency (e.g. maximum Doppler frequency) or indices of waveform shape. As the Doppler signal varies linearly with velocity, indices of waveform shape should (at least in principle) be independent of the angle θ. Some indices of waveform shape are described below and illustrated in Fig. 8.

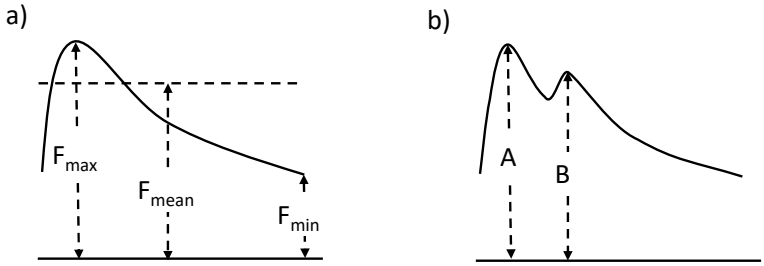

Fig. 8. Waveform indices. a) Measurements relevant to RI and PI. b) Measurements relevant to the A/B ratio.

Pulsatility index (PI). Early studies using CW Doppler observed that there was considerable variation in the degree of diastolic flow in arteries. Arteries supplying muscle at rest (eg. femoral, brachial) were highly pulsatile with little reverse flow while others (carotid arteries) exhibited a high degree of diastolic flow. Doppler waveforms distal to an arterial stenosis lost pulsatility. In its original formulation by Gosling and King in 1969 [46] PI was defined in terms of Fourier components but was later simplified [47] (Fig 8a) to:

$$PI = \frac{F_{max} - F_{min}}{F_{mean}} \tag{5}$$

Resistance index (RI). The RI also quantifies the degree of diastolic flow. This was developed by Pourcelot in 1974 [48] and is defined (Fig. 8a) as:

$$RI = \frac{F_{max} - F_{min}}{F_{max}} \tag{6}$$

A/B ratio. It was noted that waveforms from the carotid arteries and supraorbital arteries have a second peak whose height increases when there is major carotid atherosclerosis. The A/B ratio was therefore defined as the height of the major systolic peak divided by the height of the secondary peak [47], Fig. 8b. In practice not all waveforms have a clearly distinguishable second peak making this index impossible to calculate in all patients.

The above indices are measured from a single trace; so from the zero-crossing output when there is no spectral display, or from either the mean or maximum Doppler frequency when there is spectral display.

The availability of spectral display offered new possibilities for quantifying waveforms. In particular it was noticed that there was a difference between the waveforms from normal and diseased arteries. Normal arteries have a 'window' beneath the outer maximum Doppler shift where there is little data, as most of the velocities in the Doppler sample volume are travelling at similar velocities. In diseased arteries there is loss of this window as a result of turbulence. Several studies have developed indices which quantify the degree of broadening, reviewed in the book by Evans et al. on page 173 [49]. All the indices have the same intent so only a few are shown below:

$$SB = \frac{F_{max} - F_{min}}{F_{max}} \quad \text{(Johnston et al. 1981, [50])} \tag{7}$$

$$SB = \frac{F_{min}}{F_{max}} \quad \text{(Rittgers et al. 1983, [51])} \tag{8}$$

$$SB = \frac{F_{max}}{F_{mean}} \quad \text{(Sheldon et al. 1983, [52])} \tag{9}$$

In fact there are many sources of spectral broadening other than disease which makes it difficult to compare studies.

Other more complex forms of waveform analysis were explored. Laplace Transform Damping developed by Skidmore et al [53-55] was applied to the maximum Doppler frequency waveform from arteries of the lower limb. The method modelled the artery as a simple equivalent circuit and extracted parameters related to stiffness, distal impedance and diameter. However the model was unrealistic in that it constrained the waveform to start at zero flow which was unrealistic, and it did not account for the large components of reflected waves seen in disease. Principal component analysis is a generic method which breaks down the data into a number of base components, similar to Fourier analysis. This was applied to the whole Doppler spectrum from arteries in the lower limb, treating the Doppler spectrum as an image [56,57]. These more complex methods have not passed into clinical practice.

Research into waveform analysis has continued up to 2020. However the main work in this area was undertaken in the 1960s-1980s and there has been little progress since that time. Waveform analysis was superseded by the area of velocity measurement (section IX), enabled by the advent of duplex Doppler (section VII). Further reading on waveform analysis is provided in reviews [49,58].

V. Pulsed wave (PW) doppler

For CW Doppler the sensitive area arises from the cross-over of the transmit and receive beams. There may be 2 or more vessels within the sensitive region, and the exact depth from which signals arises is not known. The impetus for pulsed wave Doppler came from the need to control the depth from which Doppler signals arose. Pulsed wave Doppler systems were developed contemporaneously by Baker et al (1967, 1970), Peronneau and Leger (1969), Wells (1969) and Flaherty and Strauts (1969) [59-63]. Later Angelsen (1975, 1976) [64,65] developed a combined PW/CW Doppler unit PEDOF (Pulsed Echo Doppler Flowmeter), marketed by Vingmed (Horten, Norway). A later version of the device was marketed in 1981 as ALFRED (All Frequency Doppler).

Stand-alone PW Doppler systems did not gain the same clinical acceptance as stand-alone CW Doppler. The operator needs to adjust both probe position and depth to obtain a Doppler signal, noting that these are blind systems where the exact depth and location of the vessel is unknown. Pulsed wave Doppler suffers from aliasing, so could not accurately measure high velocity jets in disease. There was limited impact of PW Doppler in cardiac studies; that is until the work by Holen and Hatle using both CW and PW Doppler (see above) [36, 37]. Stand-alone PW Doppler was largely replaced by duplex Doppler in the 1980s. The one area where stand-alone PW Doppler gained clinical acceptance is in transcranial applications; i.e. use in the intra-cerebral circulation. Transcranial Doppler (TCD) was introduced by Aaslid et al. in1982 [66]. TCD has been used for a diagnosis of cerebrovascular disorders such as stroke, vasospasm and subarachnoid haemorrhage and monitoring of cerebral emboli [67,68].

Multi-gate PW Doppler systems were developed in 1974-75 for simultaneous measurement of the Doppler frequencies from several positions across the vessel. This allowed the first ultrasound measurement of the velocity profile in arteries [69-73]. In terms of clinical application the measurement of velocity profile has not, to date, proven to be useful in its own right. However the measurement of velocity profile has contributed to an understanding of haemodynamics in arteries. At the time established techniques for measurement of velocity profile were invasive and typically involved a hot-wire probe, inserted through arterial puncture, where the probe cooling was related to local blood velocity [74]. This method was unsatisfactory in that, apart from its invasive nature, introduction of the probe affected the flow field. The availability of a non-invasive technique which could measure velocity profile was a major advance in haemodynamic measurement.

VI. Ultrasound angiography

When a CW or multi-gate PW Doppler system was combined with a positioning arm it was possible to build up images of blood flow similar to X-ray angiograms. Ultrasound angiography systems were described for CW Doppler by Reid and Spencer in 1972 [75], for PW Doppler by Hokanson et al. in 1971 [76], and for multi-gate PW Doppler by Mozersky et al. in 1971 [77] and Fish in 1972 [78]. These provide bistable images related to the presence or absence of flow. Curry and White in 1978 [80] developed an ultrasound angiography system in which the image is colour coded dependent on Doppler frequency shift (Fig. 9).

The system developed by Fish [78,79] was further developed by GEC Medical Equipment Ltd (London, UK), later part of Picker International, as the ‘Mobile Artery Visualisation and Imaging System’ or ‘MAVIS’. The device had 30 range gates with a minimum gate separation of 0.64 mm. The device could display 2D images of flow along with the velocity profiles and volumetric flow waveforms (obtained by integration of velocities). Clinical studies were conducted using MAVIS into the 1990s [81-87]. However MAVIS was somewhat ahead of its time; and GEC concluded that ‘the complexity of the equipment and its relatively high cost made it uncompetitive in the ultrasound market’ [88].

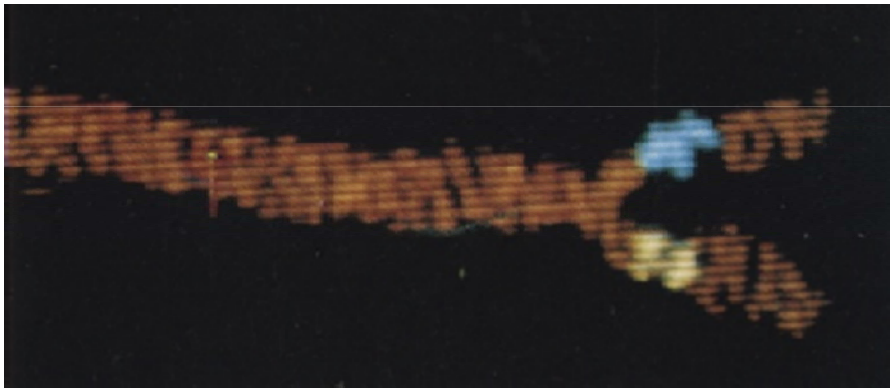

Fig. 9. Scan of the carotid artery using the Echoflow system. The display is colour-coded with higher Doppler frequencies from diseased regions shown in yellow and blue. Reproduced from Ultrasound Med Biol, Vol. no. 4, Curry GR and White DN, Color coded ultrasonic differential velocity arterial scanner (Echoflow), pp. 27-35, copyright Elsevier (1978).

VII. DUPLEX DOPPLER

'Duplex Doppler' refers to the combination of B-mode imaging and Doppler (either CW or PW). Most of the literature below refers to PW duplex Doppler, however CW duplex Doppler has also been reported [89]. Continuous wave duplex continues to be used clinically, particularly in cardiac Doppler to measure high velocities which are subject to aliasing when PW Doppler is used.

The term 'duplex Doppler' was initially introduced by Barber et al in 1974 [90,91] who combined a mechanical sector scanner with an adjacent off-set PW Doppler transducer. This allowed acquisition of real-time B-mode images and real-time Doppler, but not simultaneously. The duplex scanner allowed the operator to identify the vessel of interest, position the sample volume at an exact position within the vessel, freeze the B-mode image and obtain the Doppler waveforms.

The Seattle group continued to develop their duplex system. The review by Beach in 2005 [43] notes that the initial Barber duplex system 'proved too cumbersome to operate', and that the third iteration of the duplex system developed in 1977 had a 'scanhead that could be easily handled'[92,93] (Fig. 10). The Duplex Scanner 3 incorporated a prototype real-time spectrum analyser (Honeywell). Thus the modern duplex system had arrived fully formed, almost.

The company Advanced Technologies Limited (ATL) was established by Baker, as a spin-out from the Seattle group, in 1969. The first commercial duplex system was available in 1974-75. The Mark V Duplex scanner produced by ATL was released in 1980 and used 3 fixed-focus 5 MHz transducers within a rotating wheel (a configuration also called a 'spinner'), described by Breslau in 1982 [94]. During Doppler acquisition the B-mode image was frozen, and one of the transducers was used to generate the Doppler beam (rather than the offset Doppler of previous versions of the system).

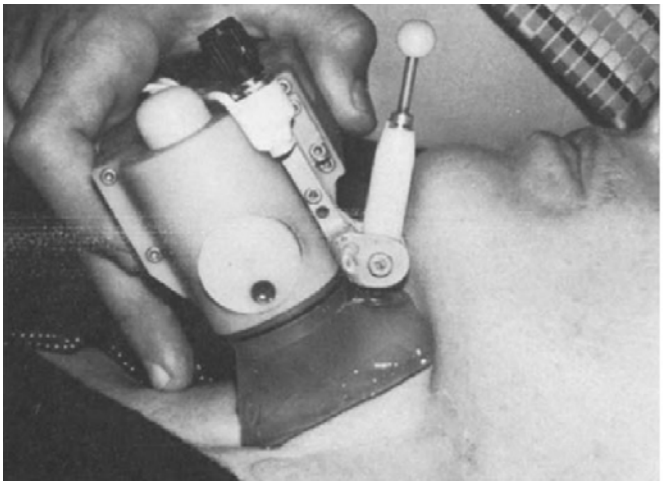

Fig. 10. 'Duplex III' transducer developed by Philips et al (1980). The offset Doppler probe is seen on the right of the transducer. Reproduced from Ultrasound Med Biol, Vol. no. 6; Phillips DJ, Powers JE, Eyer MK, Blackshear Jr WM, Bodily KC, Strandness Jr DE, Baker DW; Detection of peripheral vascular disease using the Duplex scanner III, pp. 205-218, copyright Elsevier (1980).

Contemporaneously SRI International (Menlo Park, California, USA) in partnership with the Mayo Foundation in Rochester (USA) developed their 'B-scan / Doppler' device [73,95]. This consisted of a mechanical sector scanner (reciprocating transducer) with an offset PW Doppler with 20 gates. An improved device consisted of a multi-element annular array with an offset PW Doppler [96].

Using early duplex Doppler studies were undertaken on carotid disease. Diagnostic criteria were established based on the spectral Doppler waveforms [97-101].

In the duplex devices described above, simultaneous real-time display of B-mode and PW Doppler is not possible due to the noise generated from the moving transducers. Typical operation involved a Doppler spectral trace with periodic gaps during which time the B-mode image was updated. Real-time B-mode and PW Doppler requires linear or phased array technology. The group from Osaka (Japan) reported cardiac use of a phased array duplex device consisting of an Aloka SSD-120 B-mode imaging system with an offset Hitachi EUD-4Z PW Doppler [102,103]. The same group reported use of one of the first commercial array duplex systems (Toshiba SSH 11A B-mode incorporating SDS 10A PW Doppler) for which the Doppler beam originated from the array [104-106].

Linear array duplex systems for use in obstetrics were described by Eik-Nes et al. in 1982 [107] and Teague et al in 1985 [108]. These were hybrid systems consisting of a real-time B-mode imaging system and an offset PW Doppler system. Toshiba in 1982 produced one of the first duplex linear arrays incorporating PW Doppler into the array (SSL-53M linear array with the SD-10 pulsed Doppler module). Berson et al. in 1987 [109] described a duplex system which incorporated Doppler into the linear array.

The essential features of the duplex system, incorporating real time B-mode, real time PW Doppler and real-time display of the Doppler spectrum have remained unchanged to the present day (2020).

Applications in areas of the body consisting of multiple vessels with similar waveforms were impossible for stand-alone Doppler. These areas were now accessible using duplex scanning, and the first investigations in the fetus [110] and abdomen [111] followed.

It is worth noting that there were only a few years (4-6) between the development of PW Doppler and the development of duplex ultrasound. Very little of the PW Doppler technology from the intervening period survived into clinical practice, emphasising the importance of the development of duplex ultrasound.

VIII. Colour imaging of blood flow and tissue motion

Despite the very considerable developments and technical adventures in Doppler ultrasound since its introduction in 1959 by Satomura, there was actually little clinical penetration by the early 1980s. The introduction of colour flow, initially by Aloka in 1982, moved Doppler ultrasound into mainstream clinical usage. Colour-flow for the first time provided a real-time view of blood flow which could compliment real time B-mode imaging.

The Seattle group continuing the development of the Duplex system produced the 'Duplex scanner IV', with 3 rotating transducers for formation of the B-mode image and an offset PW Doppler. Brandestini et al in 1978 [112] had developed multi-gate PW Doppler, which was incorporated into the Duplex scanner IV by Eyer et al. in 1981 [113], Fig. 11. By scanning the PW Doppler through the field of view the first colour flow images were obtained. Each image involved manual movement of the PW Doppler transducer over about 20 seconds. The system could also be used in M-mode.

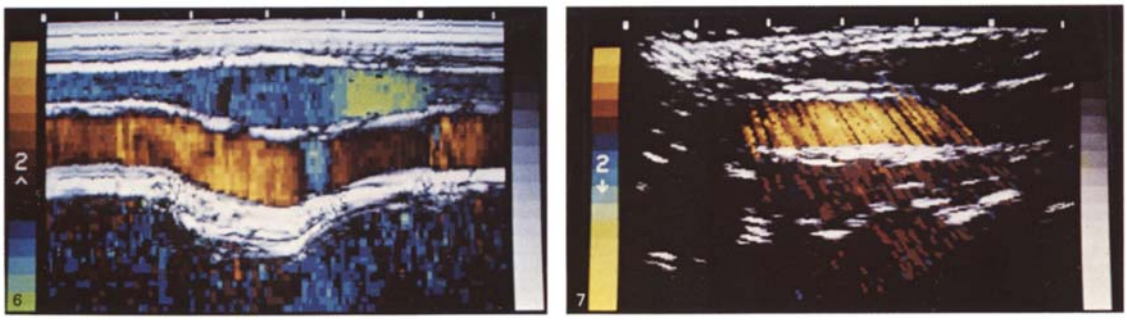

Fig. 11. Colour flow images from Eyer et al. (1981). a) Composite echo/flow M-mode of the jugular vein and common carotid artery during a Valsalva manoeuvre. Increasing time is defined to be from left to right with the entire horizontal axis covering 3.0 seconds. b) B-mode flow map of the common carotid artery obtained in the mid-neck region. Reproduced from Ultrasound Med Biol, Vol. 7; Eyer MK, Brandestini MA, Philips DJ, Baker DW; Color digital echo/Doppler image presentation, pp. 21-31, copyright Elsevier (1981).

The move to real-time colour flow was made possible by the development of the autocorrelation technique for direct measurement of the mean Doppler frequency by Kasai, Namekawa and colleagues from Aloka Company Ltd., Japan [114-116]. In 1982 the first commercial colour flow system was based on the autocorrelator and was produced by Aloka. The autocorrelator requires a minimum of 3 ultrasound pulses to produce a value for estimated mean Doppler frequency (compared to 50-100 for spectral Doppler) and was the breakthrough which made possible real-time colour flow imaging. Colour flow was quickly adopted for cardiac use [117-119] with early studies in arteries [120,121].

Developments in signal processing of colour flow systems are covered by reviews by Evans [49,122,123]. This article will discuss only a small number of relevant developments. The original autocorrelator technique described by Kasai was extended by Loupas et al. in 1995 [124,125], who developed '2D autocorrelation' which has been widely adopted in the commercial sector. The clutter filter is a key component of the processing chain. Early colour flow systems had poor ability to visualise low velocities as a result of the simple design of the clutter filter, and the main use of colour flow was in cardiology where jet velocity is high. Improvements in clutter filter design led to an improvement in the ability of colour flow to visualise lower velocities, and this was followed by widespread clinical adoption of colour flow in radiology.

Three quantities are calculated in colour flow; mean Doppler frequency, Doppler signal power and 'variance'. The variance is a measure of the spread of Doppler frequencies within the received signal. Variance increases in turbulence and may be shown together with the mean-frequency in a composite display.

Display of the Doppler power was a feature of early colour flow systems, but the same settings were used as for display of mean frequency [126-128]. Optimisation of the colour flow settings by Rubin et al. [129,130] enabled improved visualisation of small vessels, and 'power Doppler' became of clinical interest. Power Doppler has been widely used to provide qualitative and quantitative data on vascularity.

Optimisation of the colour system also allows visualisation of tissue motion. The technique of 'Tissue Doppler Imaging' or TDI was introduced by McDicken et al. in 1992 [131]. The signals from tissue are some 40dB higher than from blood so the Doppler gain is reduced. The signal from blood is of very low magnitude so is not displayed. The clutter filter and blood tissue discriminator are redundant. The signal strength is high so fewer pulses are needed to estimate mean Doppler frequency. Tissue Doppler imaging has been widely used in cardiac studies (Fig. 12). From the velocity data the local strain may be estimated which is of interest in detection of ischaemic regions where the strain is reduced [132,133].

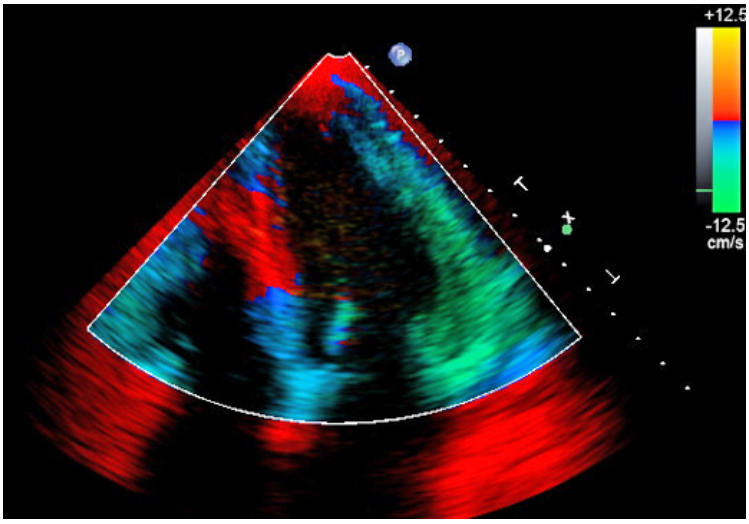

Fig. 12. Doppler tissue image from the heart.

A major limitation of ultrasound imaging, especially colour flow, has been the frame rate which can be achieved. With multiple receive beam-forming [134] frame rates of 200 s^{-1} can be achieved for 2D imaging. This is very high, however achieving high frame rates in 3D imaging and in colour flow is far more challenging using conventional beam-forming techniques. The development of synthetic aperture techniques has led to a vast increase in the amount of data available across all ultrasound imaging modalities. Increase in the information available relies on dispensing with focus-on-transmit. Instead a plane wave or spherical wave is transmitted using all transducer elements, and the image is formed using focus-on-receive. For B-mode imaging frame rates of 10,000 – 20,000 s^{-1} can be generated, with obvious loss of resolution due to the absence of focus-on-transmit. A review of high frame rate techniques for colour flow is provided by Jensen et al. in 2016 [135]. The technique described by Bercoff et al. in 2011 [136] is called 'ultrafast compound Doppler' (UCD) and involves transmission and reception of a series of N plane waves at different angles. Bercoff shows that the frame rate for UCD was 7 times higher than for conventional colour flow, with similar image quality. The availability of such a large amount of data means that choices can be made as to which aspect of image quality to improve. Depending on the application there can be a factor of around 10 increase in frame rate, sensitivity, minimum detectable velocity or minimum detectable vessel diameter. For microvascular imaging, improvements in clutter filtering [137] reduced the minimum detectable velocity from 5 $mm.s^{-1}$ (conventional colour flow) to 0.5 $mm.s^{-1}$, and hence visualisation of vessels down to about 50 micron. New clinical applications have arisen from these developments, in particular 'functional ultrasound' (mirroring functional MRI), concerned with measuring changes in brain activity which are associated with changes in blood flow [138,139].

Colour flow has become an essential feature of the modern cardiovascular ultrasound system. Most clinical practice still relies on spectral Doppler for quantification of blood velocity (see below), with colour flow reserved for qualitative visualisation of the flow-field and of vascularity. While Tissue Doppler Imaging has proven popular in research it has had limited impact on clinical practice. The impact of high frame rate techniques on Doppler ultrasound is still rolling out and it is likely that clinical practice will make more quantitative use of data from colour flow in future.

IX. Measurement of blood velocity and related quantities

A. Estimation of blood velocity

Measurement of blood velocity requires knowledge of the angle between the direction of motion and the Doppler beam. For a single Doppler beam in which the transmit and receive Doppler beams are aligned the velocity may be found by rearranging equation 4:

$$v = \frac{c}{2f}\frac{f_d}{\cos\theta} \quad (10)$$

The ultrasound machine knows the speed of sound and transmit frequency; estimation of velocity therefore requires knowledge of the Doppler frequency shift and the angle θ in the subject.

Early attempts (1970-73) to estimate θ were made using CW and PW Doppler systems. Cumbersome techniques involved finding the orientation of the transducer which is at 90º to the vessel (at which point there is positive and negative symmetry of flow), and then orientating the transducer by a known angle (see page 200 of the text by Evans et al. [49]). The use of 2 or more receive transducers provides an automated method for estimating angle. The velocity component in 2 or more directions is estimated allowing the angle to be calculated [140-143]. These were the forerunner of vector-Doppler techniques described in the next section.

The advent of duplex Doppler provided a clinically-useful means for estimating θ, by enabling the operator to align the angle cursor with the vessel wall. The review on velocity measurement described here relates mostly to clinical ultrasound systems; i.e. those with a single Doppler gate using spectral Doppler.

In clinical use criteria were introduced based on measurement of blood velocity, especially for grading of the degree of stenosis, where the blood velocity increases with degree of stenosis [144]. Typically the maximum Doppler frequency shift has been used (rather than mean frequency), as this is relatively invariant with minor changes in transducer alignment and sample volume position within the vessel.

There was a growing understanding of the Doppler measurement process and the causes of velocity measurement errors following the introduction of the duplex scanner in the 1970s. Implicit within the Doppler equation is the assumption that a single velocity will give rise to a single Doppler frequency shift. In fact a single velocity will give rise to a range of Doppler frequencies; a phenomenon called 'spectral broadening'. Newhouse et al. in 1977 [145] demonstrated that the finite width of the transducer gives rise to broadening as a result of the range of angles which the blood velocity subtends at the transducer. It was shown by Censor in 1988 [146] that the 'geometric spectral broadening' f_{gsb} is given by:

$$f_{gsb} = \left(\frac{2fv}{c}\right)\left(\frac{D\sin\theta}{2L}\right) \quad (11)$$

where D is the width of the Doppler aperture and L is the depth of the Doppler sample volume.

Newhouse et al. in 1976 [147] investigated transit time broadening which is due to the finite time taken for scatterers to cross the beam, then proposed that transit time and geometric spectral broadening were the same phenomenon [148]. This equivalence was accepted for many years before Guidi et al. in 2000 [149] demonstrated that these were different phenomena, however this had actually been proven 14 years previously by Fish in 1986 [150], equation 11.81, p363). The data from Guidi suggest that, around the transducer focus, spectral broadening is dominated by the geometric component (a factor of 6 compared to transit time broadening).

When linear arrays are used to generate the Doppler beam this leads to a large amount of geometric spectral broadening, which in turn leads to overestimation of blood velocity [151,152]. The explanation is that the ultrasound machine angle-corrects to the middle of the array, whereas the highest Doppler frequencies are found at the edge of the array. In fact the equation which is relevant for the highest Doppler frequency shift f_{max} is a combination of equation 4 and 11 [153]:

$$f_{max} = \left(\frac{2fv}{c}\right)\left(\cos\vartheta + \frac{D}{2L}\sin\theta\right) \quad (12)$$

The error in estimated maximum velocity varies with angle, depth and machine. Typical errors are in the range 0-40% for clinical settings and potentially lead to impact on selection of patients for surgery [154]. Despite the known errors there has been no move on the part of manufacturers to provide correct estimation of blood velocity.

B. Estimation of pressure gradient

A method for estimation of pressure gradient across cardiac valves was reported by Holen and then by Hatle [38-41]. The method is based on a consideration of the Bernoulli equation which concerns energy in flow. The pressure drop is:

$$P_1 - P_2 = \frac{1}{2}\rho({v_2}^2 - {v_1}^2) + \rho\int_1^2 \frac{dv}{dt}ds + R(v) \quad (13)$$

where suffix 1 denotes the fluid position element in front of the valve and suffix 2 in the valve jet; P is the pressure, v is the velocity vector of the fluid element, and ds is the path element.

The first term relates to change in kinetic energy, the second term represents acceleration caused by change in velocity with time, and the third term represents viscous loss.

Holen and Hatle argued that the second and third terms were small compared to the first term. Noting also that $v_2 >> v_1$, a simplified equation results:

$$P_1 - P_2 = \frac{1}{2}\rho{v_2}^2 \quad (14)$$

Inserting the value for density, and expressing the pressure difference in mm Hg, the final equation is derived:

$$P_1 - P_2 = 4{v_2}^2 \quad (15)$$

This technique has had widespread clinical adoption, initially using stand-alone CW/PW Doppler systems, then with duplex Doppler.

C. Estimation of volumetric flow

The first attempts to measure volumetric flow were undertaken using multi-gate PW Doppler systems in 1974-75 [69-73]. These systems did not incorporate B-mode imaging so the procedures described in subsection A were used to measure the beam-vessel angle. Assuming that flow was axial (non-rotational) and axially symmetric, the velocity profile could be integrated to produce an estimate of the instantaneous volumetric flow. The mean volumetric flow could then be obtained by integration over the cardiac cycle.

Volumetric flow Q was estimated using a duplex scanner with measurement of diameter d from the B-mode image (from which area is calculated assuming that the vessel is circular) and measurement of the velocity waveform from the Doppler spectral data, noting that this has been angle-corrected by alignment of the angle cursor with the vessel wall:

$$Q = V_{ta}\frac{\pi d^2}{4} \quad (16)$$

where V_{ta} is the time-averaged velocity obtained from the Doppler waveforms.

Early reports of volumetric flow measured using duplex Doppler were published from 1979-1985 [110,155-157]. While the equation used to estimate volumetric flow is straightforward, there are several sources of error which must be considered. The principle problem is the relationship between the Doppler statistic and the mean velocity. Commonly the mean Doppler frequency is used, and it is assumed that the mean frequency when angle corrected is equal to the instantaneous mean velocity. This might be the case were the vessel uniformly insonated. However a typical Doppler beam is thin compared to the vessel diameter so that the mean velocity calculated from mean frequency is usually overestimated; for example for a very thin beam in steady flow the overestimation is 33% [158]. Further complexity arises in pulsatile flow as the velocity profile changes through the cardiac cycle. In addition mean frequency is highly sensitive to small changes in alignment between the vessel and the beam. An extensive discussion of the errors in volumetric flow is provided in by Gill [159], Evans [160] and in chapter 11 of the textbook by Evans et al. [49]. When compared with 'gold standard' measurements it was found the median rms error in flow measurement across several studies was 16% (range 11-34%) [58]. The mean Doppler frequency has remained the statistic of choice in the literature, despite the known problems and errors.

An alternative approach is to use the maximum Doppler frequency, noting that the overestimation of velocity as a result of geometric spectral broadening must be corrected. Maximum frequency has the advantage over mean frequency of not varying for small misalignments of the transducer caused by movement of the operator or patient. The maximum velocity is estimated from the maximum Doppler frequency shift, and there are 2 methods which have been developed which allow estimation of flow from maximum velocity. For estimation of time-averaged flow rate it can be assumed that the average velocity profile is parabolic, provided that flow is fully-developed [161]. In this case the time-average maximum velocity $V_{ta\text{-}max}$ is estimated and flow Q can be calculated:

$$Q = \frac{V_{ta-max}}{2}\frac{\pi d^2}{4} \quad (17)$$

The second method for estimation of volumetric flow from the maximum velocity waveform makes use of the Womersley equations [162]. These equations describe the velocity profiles during fully-developed flow for a Newtonian fluid. The equations are formulated in terms of diameter and flow rate, however these can be modified to allow input of the diameter and the centre-line (maximum velocity) waveform [163] with output of the time-varying velocity profile data. Once the time-varying velocity profile data is available, volumetric flow can be obtained by integration of the profile data, and in addition this technique also gives the time-varying wall shear rate [164] (Fig. 13).

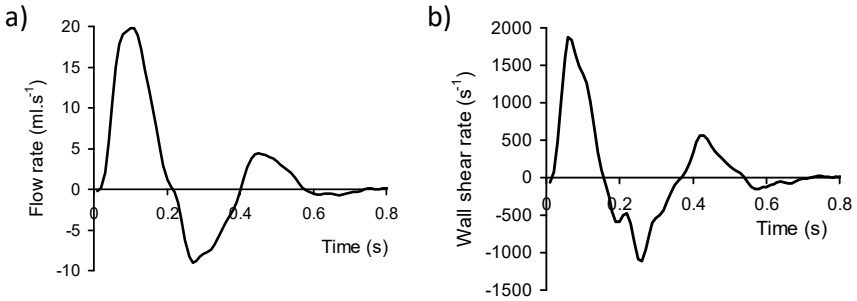

Fig. 13. Estimation of volumetric flow and wall shear stress (Blake et al. 2008 [164]).

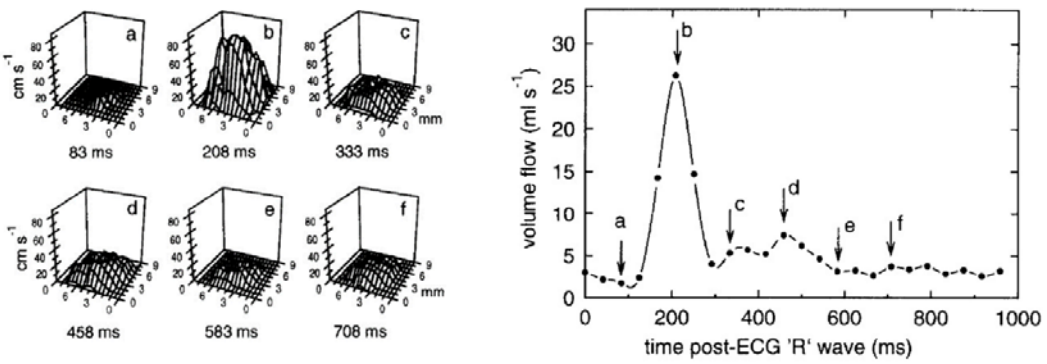

Fig. 14. Estimation of 2D velocity profile and flow. Reproduced from Ultrasound Med Biol, Vol. 21; Picot PA, Fruitman M, Rankin RN, Fenster A; Rapid volume flow rate estimation using transverse colour Doppler Imaging, pp. 1199-1209, copyright Elsevier (1995).

Colour flow provides multi-gate velocity information and has been used to provide estimation of volumetric flow. This has been used to measure the 1D velocity profile, from which volumetric flow has been calculated assuming axially symmetric fully-developed flow [165]. Colour flow has also been used to estimate volumetric flow from the 2D velocity profile using an oblique cross section through the vessel [166], Figure 14.

The discussion of this section has emphasised the difficulties and assumptions in estimation of volumetric flow using the duplex system. Ideally what is required is a method which does not make assumptions related to fully-developed flow or axial symmetry. It is likely that the techniques, discussed in the next section, which are able to measure 3D and 3-component velocity field data are likely to provide accurate and clinically useful information on flow rate and other haemodynamic quantities.

X. 3D AND VECTOR DOPPLER

This section will cover 3D and vector Doppler. These are deliberately combined in that they are attempting to solve the same problem which is a more complete characterisation of the flow field. Early studies, described above, on velocity measurement largely assumed that the blood is flowing parallel to the vessel wall; this assumption is embedded in the phrase 'beam-vessel angle' to describe the angle between the beam and the direction of motion. Understanding of haemodynamics gained momentum in the 1980s and 1990s due to the availability of tools which could measure flow patterns. In the lab optically transparent phantoms were developed where complex flow patterns could be seen in a carotid model [167]. The first numerical simulations were undertaken of blood flow by Perktold et al in 1984 [168] and Friedman and Ehrlich in 1984 [169]. Both ultrasound and MRI were used to demonstrate spiral flow in arteries [170-172]. A key concept is the idea of 'fully developed flow'. This can be understood with reference to flow from a reservoir into a long straight pipe. Near the entrance to the pipe the velocity profile is flat and after a certain distance called the 'inlet length' the velocity profile settles down to a fixed shape (for Newtonian flow this is a parabola). Any change of geometry such as a bend, bifurcation or disease will cause alterations in velocity profile. The underlying assumption of much of the discussion on velocity measurement in section IX is that the flow is fully-developed. In some arteries in health flow may well be fully-developed, for example in the distal regions of arteries in the arm and leg. However many arteries are short with strong curvature and flow will not be fully developed. In addition there are helical components to flow. There is helical flow in the normal aorta [173] and bending of arteries and bifurcations will induce helical flow [174]. Disease such as atherosclerosis and aneurysms will also affect flow profiles and can cause non-axial flow. The common carotid artery has been the subject of considerable interest in ultrasound. Figure 15 is an image of streamlines of flow calculated using computational fluid dynamics showing highly complex flow patterns. A full characterisation of flow requires 3 spatial components, 3 velocity components and time, so 7 components in all. Full time-varying flow field data is sometimes referred to as '7D flow'. It will be seen that the history of Doppler ultrasound is one of progression towards 7D flow.

Early work on 3D Doppler involved mechanically scanning the array [172]. A series of transverse images were acquired while moving the transducer along the length of the artery. There is change in diameter of the artery during the cardiac cycle, so ECG gating was necessary in order to collect data at the same point in the cycle. Colour flow images were acquired which made one of the first observations of helical flow in the carotid artery (Fig. 16). The development of 3D ultrasound is described in the review by Fenster in 2011 [175]. Early commercial 3D systems used a swept linear array. The first 3D ultrasound system based on a 2D array was developed by Volumetrics Medical Imaging (Durham, North Carolina, USA) and available at the end of 1990s (Fig. 17). The Volumetrics system was based on technology developed by the group at Duke University [176]. Later Philips Medical Systems produced a 2D array system with part of the beam-forming within the transducer. In 2020 commercial 3D ultrasound systems are a mix of swept array, 2D array, and (for endoprobe systems) mechanical pullback.

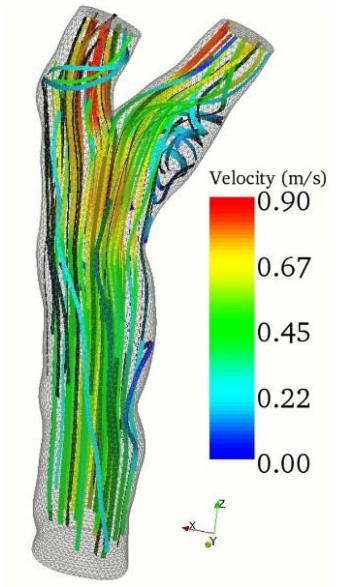

Fig. 15. Streamlines of flow in a diseased carotid artery; showing helical flow and recirculation in the bulb.

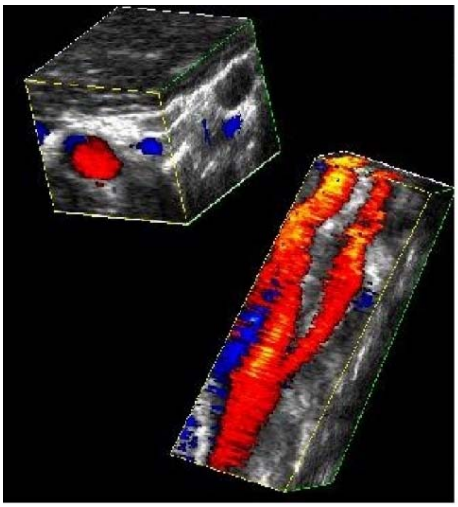

Fig. 16. 3D colour flow images of the carotid artery. From Aaron Fenster, London, Ontario.

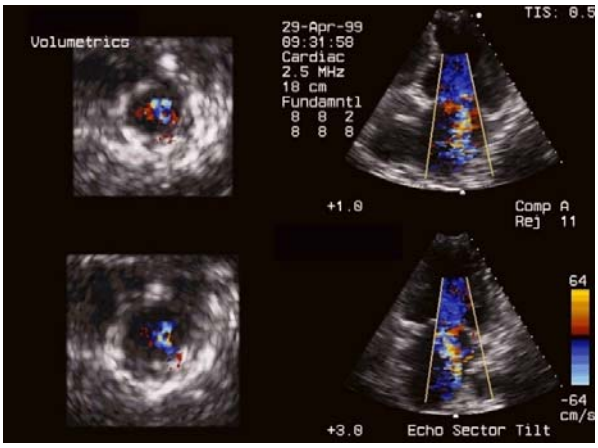

Fig. 17. 3D colour flow imaging using the Volumetrics system from 1999.

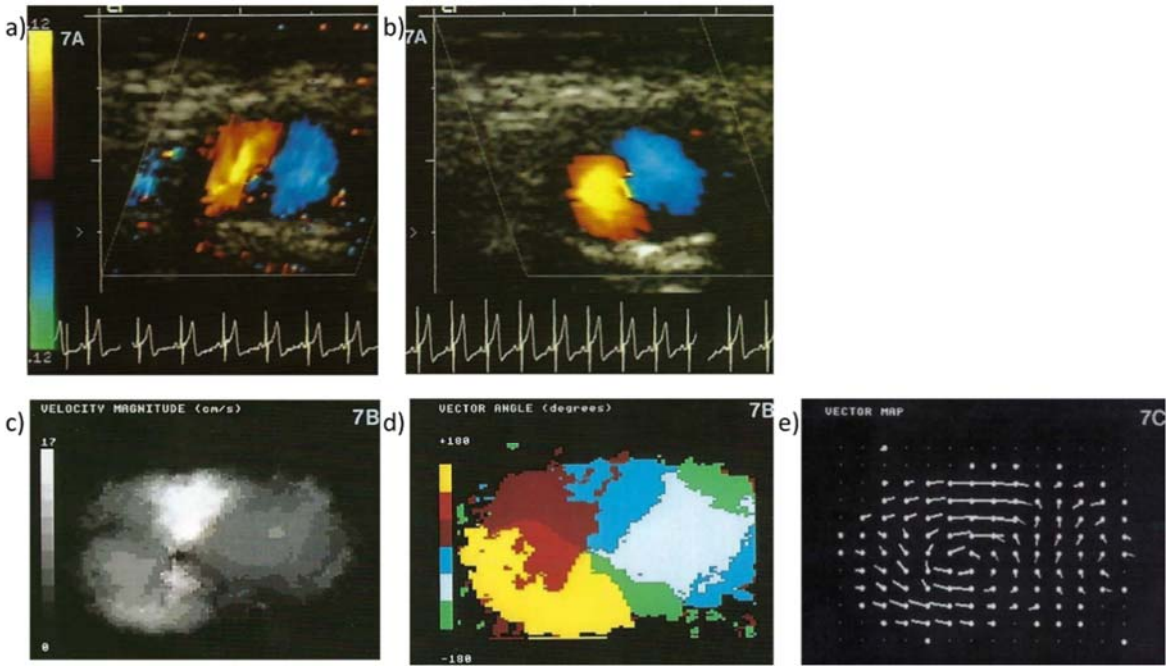

Fig. 18. Vector Doppler images showing spiral flow in the femoral artery. a) and b) colour flow images obtained with the beam pointed to the left then right, c) velocity magnitude, d) vector angle, e) vector display. From Hoskins PR, Fleming A, Stonebridge P, Allan PL, Cameron D. Scan-plane vector maps and secondary flow motions. European Journal of Ultrasound 1994;1:159-169.

Single-beam Doppler systems provide information on 1 velocity component; the component aligned with the ultrasound beam. Vector Doppler involves estimation of velocity components from different directions and compounding these to obtain the velocity magnitude and direction. Vector Doppler systems based on single element transducers were able to make measurements from a single sample volume of 2 velocity components [177-180] or 3 components [178-184]. 2D vector Doppler images may be obtained by using colour flow, with the colour beam steered in 2 different directions [172,185-189]; Fig. 18.

In addition it is possible to use the spectral width to estimate the direction of motion. Combining equations 4 and 11 gives the fractional spectral width:

$$\frac{f_{sw}}{f_{mean}} = \frac{D \tan\theta}{L} \tag{18}$$

where f_{mean} is the mean Doppler frequency, and f_{sw} is the spectral width. Rearranging equation 11 gives:

$$\theta = \tan^{-1}\left(\frac{L}{D}\frac{f_{sw}}{f_{mean}}\right) \tag{19}$$

Equation 12 allows the angle θ to be estimated using a single beam, provided that it is only geometric spectral broadening that is contributing to the spectral broadening. This method has been used to estimate 2 components using a single beam system and 3 components in a 2-beam system [181,182].

A patent was published by Hall et al. in 1995 [190] for an array vector Doppler system in which the array was divided into a central transmit aperture with receive apertures on either side. A prototype array based vector Doppler system was produced by ATL Ultrasound using a similar approach, which proved successful in phantoms and in normal volunteers in acquiring velocity measurements which were angle-independent [191,192]. However despite the ease with which vector Doppler could be adopted on array based systems, there was no commercial adoption of cross-beam vector Doppler and hence no clinical penetration. A review of cross-beam vector Doppler is provided by Dunmire et al. in 2000 [193].

The first commercial colour vector Doppler system was available from BK Medical using the transverse oscillation (TO) method [194,195]. This method creates an oscillation in the transverse direction from which the transverse velocity component can be estimated. Creation of the transverse oscillation is performed in reception; this produces 2 beams from which the transverse velocity component can be estimated. The transmit beam is unchanged in TO from conventional Doppler, so that TO has a higher frame rate and larger field of view than comparable cross-beam techniques.

The TO method has been modified for 3 component velocity estimation [196,197] with 2D 3-component velocity profiles obtained from carotid arteries [198]. The same methodology was also used to acquire 3D 3-component flow in the heart using ECG gating at 50 frames per second [199].

The techniques above, involving cross-beam and transverse oscillation provide real-time 2D imaging with 2 velocity components, so 5 of the 7 dimensions. As noted above recent years have seen the development of high frame rate synthetic aperture techniques involving either plane wave insonation or spherical wave insonation, with focus on receive. Frame rates of over 3000 s^{-1} were achieved for real-time 2D 3-component imaging of flow in the carotid artery [200]. It is using synthetic aperture imaging that real-time 3D 3-component (i.e. 7D) flow imaging is likely to become available in the near future. Further details of recent developments in vector Doppler are available in the reviews by Jensen et al. [135,201].

XI. Chronological summary of major developments by decade

1950s

CW Doppler systems developed by Satomura (Osaka, Japan) for measurement of heart wall motion and blood flow.
Commercial Doppler system; the Ultrasonic Blood Rheograph (Nippon Electric Company).
Invasive Doppler probes developed by Franklin (Seattle, USA).

1960s

Development of CW Doppler technology; flow discrimination methods, real-time single-line display of mean Doppler frequency from a zero-crossing detector, off-line spectral display.
PW Doppler systems.

1970s

Multi-gate Doppler systems, allowing measurement of velocity profile.
Ultrasound angiography.
Duplex Doppler.
Colour flow system using slow-sweep of Doppler beam.
Methods for angle-estimation (based on multiple single elements).
Identification of causes of spectral broadening; geometric and transit time.
Formulation of RI and PI for waveform analysis.
Methods for estimation of pressure gradient across cardiac valves from Doppler measurements of velocity.
Volumetric flow estimation using duplex scanners.

1980s

Real-time colour flow system with applications in the heart, and later in vessels.
Theoretical understanding of the origin of Doppler spectra.
Doppler waveform analysis; quantification of waveform shape and spectral content for use in diagnosis.
Vector Doppler systems (based on multiple single elements).

1990s

Tissue Doppler Imaging (of heart motion, and later vessel wall motion).
Power Doppler.
Colour vector Doppler (off-line then later real-time).
3D colour-flow.
Improved understanding of the estimation of blood velocity and related quantities using Doppler.

2000-2020

High frame rate Doppler and related applications (colour flow, spectral Doppler, vector Doppler, microvascular imaging, functional ultrasound).

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A HISTORY OF HIGH INTENSITY FOCUSED ULTRASOUND (HIFU) THERAPY

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***Abstract*— Therapy ultrasound has, for many decades, been thought of as the poor relation of diagnostic ultrasound imaging. However, its roots predate the scanning applications of this most versatile of energy forms used in medicine. In this review the history of high intensity focused ultrasound (HIFU) is traced from its first mention in 1942 until 1970 at which point there was a lull in interest until its resurgence in the 1980's.**

***Keywords*— therapy ultrasound, minimally invasive, brain, transducer, cancer**

I. Introduction

It could not easily have been predicted that the report by Wood & Loomis in 1927 that ultrasound energy can adversely affect living organisms would trigger areas of research that have led to the development of a number of different therapies [(1)]. The appeal of ultrasound energy for therapeutic purposes is enhanced by its physical characteristics when propagating through water at low megahertz frequencies. These allow tight focusing into volumes and distances from the source that are clinically relevant.

While ultrasound has many potential therapeutic uses, including for physiotherapy treatments of soft tissue injuries [(2, 3)], the acceleration of healing of bone fractures [(4)], and the improvement of drug delivery through the skin (sonophoresis) [(5)], it has been its applications in neurosurgery and cancer therapy that have been the most commonly adopted to date, with the use of high intensity focused ultrasound beams being the most highly favoured, especially in the early days. This technique is referred to interchangeably as HIFU or FUS (focused ultrasound surgery). Historically, it was the applications in the brain that largely drove the development of HIFU devices. In this review, the history of HIFU from its inception until 1970 will be presented.

As can be seen from Figure 1, there was a lull in developments in this area in the 1970's, with a rapid resurgence of interest from the mid 1980's.

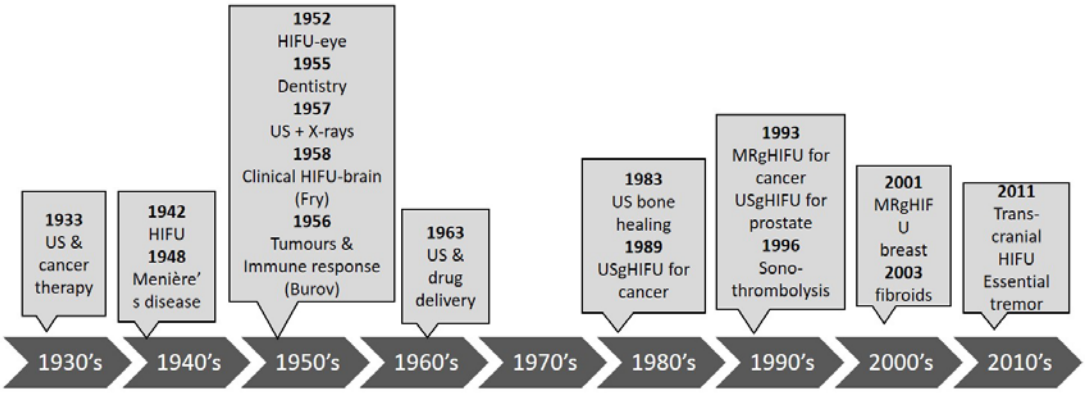

Fig. 1 Timeline, showing introduction of new HIFU applications

From a medical physics perspective HIFU provides many challenges. The different aspects of a HIFU treatment involve initial imaging and treatment planning, treatment delivery and its monitoring, and then follow up. Although early HIFU was always performed under ultrasound guidance (USgHIFU), more recently, magnetic resonance imaging (MRgHIFU) has found favour, largely because of the ability to run thermometry sequences.

II. Principles of HIFU

It has been known since the time of the early Egyptians that heat can play a role in cancer therapy [(6, 7)]. The field of hyperthermia, in which tumours are subjected to either radio- or chemotherapy and elevated temperatures has been extensively explored clinically, and pre-clinically. For this application temperatures of 42-45°C are maintained for times of up to 60 minutes. The heat and adjuvant therapy are applied simultaneously, or, where this is not possible, as close together in time as possible. There are a number of disadvantages to this technique as it requires repeated treatments and, perhaps more difficult technically, it is important that the chosen temperature remains within defined limits for the duration of the heating period. This is difficult to maintain as the body's response is to increase blood flow to provide cooling, adding to the problem of accurate intratumoural thermometry over the entire heated volume.

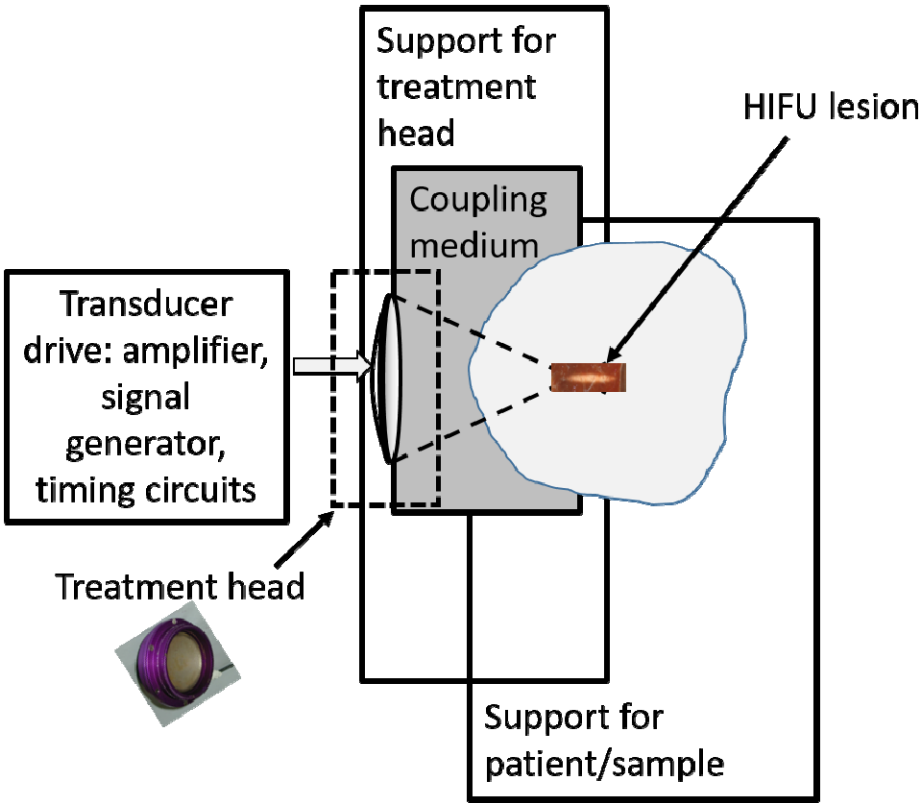

Fig 2. Schematic showing the necessary components of a HIFU system.

HIFU, in contrast, relies on the fact that higher temperatures (in excess of 56°C held for ~ 1 second) can lead to instantaneous cell death (thermal ablation). This can be achieved in a single, short treatment, and as long as thermal necrosis is seen, there is no need for thermometry. Thus, if heating to these temperatures can be achieved selectively in a desired volume (for example, a tumour), with no damage to surrounding tissues, then thermal ablation becomes a viable therapeutic option. While it is often possible to insert a probe capable of generating ablative temperatures into the target (and indeed the Egyptian Edwin Smith papyrus [(8)] suggests the use of a 'fire drill' – a heated stick – inserted into a breast tumour), the non-invasive solution offered by HIFU appears to have significant advantages,

Figure 2 shows a schematic of the principles on which a HIFU treatment works, and the components that are essential to delivering such a treatment. Histologically, the ellipsoidal region of damage created (the 'lesion') has what has been described as an 'island and moat' appearance [(9, 10)], with the boundary between normal and affected tissue being very sharp. Tissue within the 'island' appears almost normal, with cells being heat fixed, and in the 'moat' cells are severely disrupted, with liquefaction, deformed nuclei, and absence of structure.

III. HIFU DELIVERY

While Wood & Loomis' 1927 observations of stimulating and lethal effects in unicellular organisms, small fish and animals, made using a plane transducer [(1)] generated a considerable amount of interest, Lynn et al [(11)] (working in Columbia University, New York) were the first to publish biological effects in focused fields (1942). They claimed that it was the finding of Grűtzmacher that ultrasound waves could be focused with a gain of 150 using a curved quartz surface [(12)] that stimulated their studies. Grűtzmacher had described a constant thickness spherical piezo-electric quartz shell (concave – convex) cut so that the X- crystallographic axis coincided with the beam axis (X-cut), and with electrodes plated on both surfaces.

In their first published paper, Lynn et al describe using 835 kHz ultrasound, generated by putting an oscillating electric potential across the opposite faces of such an X-cut concave quartz crystal. This potential was supplied by "what amounts to a small 0.5 kW radiotransmitter, such as used for code signaling". The power supply design is described as a "full wave mercury rectifier circuit (type 866) with choke input filter". Lynn et al's design of crystal mount [(11)] is one that has been replicated by many. The mounting of a transducer to maximize its acoustic output for therapy has specific requirements. The crystal must be held securely and evenly around its circumference, with as little damping of its vibrations as possible. The

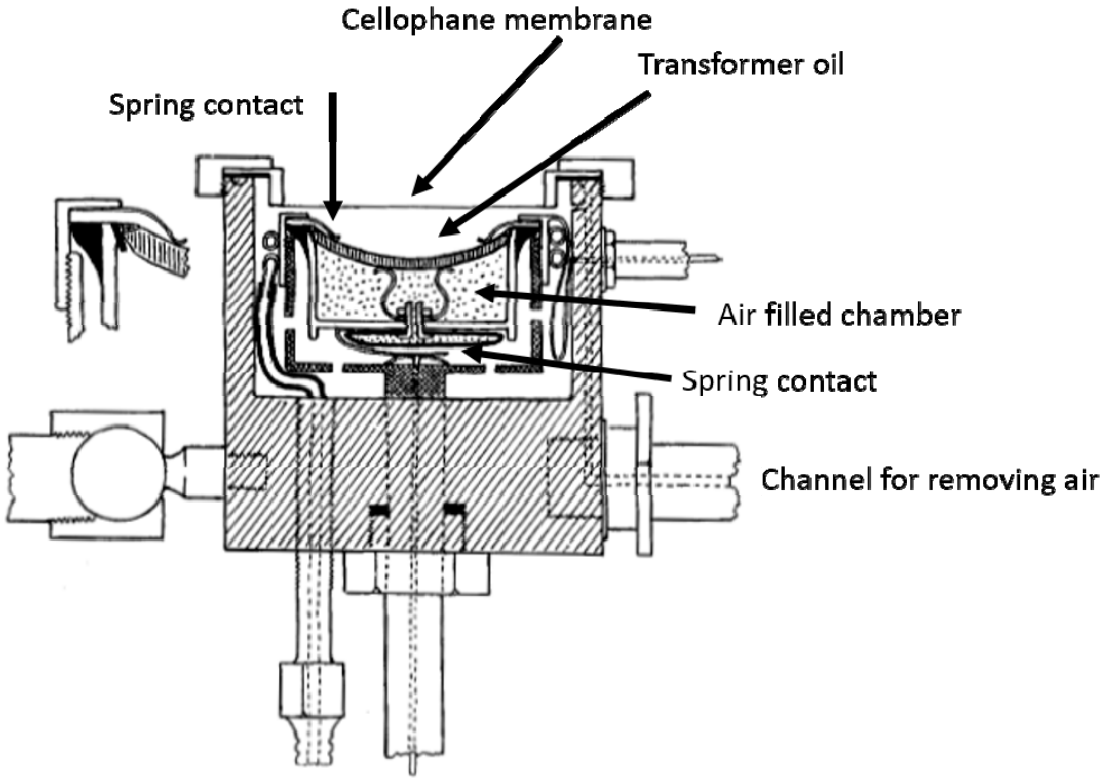

Fig. 3 Transducer mount used by Lynn et al (11).

method of mounting should be such that there is minimum restraint as this can lead to reduced output and mechanical failure. Their solution was to air back the quartz crystal, in a way suggested by earlier authors. Figure 3 shows their transducer design [(11)]. The air backing minimizes damping, but also serves as a total reflector of the sound wave, thus enhancing the forward power from this transducer. This method has been used to effect to the present day.

Changes in focal depth for this system were achieved by mounting the transducer and its immediate casing on a piston structure that enabled it to be moved forwards and backwards relative to the 'thin cellophane diaphragm' that was in contact with the skin. Cooling was provided in part by surrounding the transducer, mounted on a universal joint, with clear transformer oil.

These authors describe the calibration of their system in delightful detail. They placed 'several millimetres' of oil on the top of the front cellophane membrane, and describe the acoustic fountain obtained when the

transducer is activated, pointing vertically upwards : 'a conical column, from the top of which oil droplets were thrown upward and outward so that the entire phenomenon resembled an erupting volcanic cone'. They claimed the height of the cone to be a 'reliable (but crude)' indicator of output, reporting that a plate current of 220 mA, potential 2400V gave a peak output of 900 mA of RF current, and a cone 12.5 cm high with spray extending to twice this height. More details of early calibration methods are discussed in section V below.

Further studies with the Columbia group's device showed that they could produce focal damage in blocks of beef liver only when full power was applied for 10-15 seconds. Follow up experiments in the brains of 3 cats and 2 dogs, resulted in scalp damage in all subjects, and with brain damage only being seen in the 2 animals who were exposed to maximum acoustic power. The necessity of applying sufficient energy very fast when seeking selective focal damage is discussed in this paper. They cite the work of Gohr & Wedekind [(13)] in recognizing that the presence of blood perfusion has an effect on the ability to produce focal lesions.

A. Brain studies

Lynn's group was also the first to publish detailed histology of lesions created in the brain [(14)]. They exposed the spinal columns of 3 dogs, 30 cats and 4 monkeys to 835 kHz ultrasound from a 5.5 cm focal length transducer. They reported 'partial success' with trans-cranial and trans-vertebral focal lesions of the central nervous system. While they were able, at will, to produce irreversible, reversible or partially reversible damage which they refer to as 'disabilities', they inevitably created damage to the overlying skin and soft tissues. The damage included cortical blindness, cerebellar ataxia, monoparesis, and bilateral paresis of the hind extremities. The lesions were conical in shape, and this paper is one of the first to report mechanical tearing and formation of cavities, which, they postulated were due to the 'sudden explosive release of dissolved gases from tissue solution'. They found that ganglion cells were more susceptible to damage than glial elements, and also that the least affected components of the brain were the blood vessels.

A number of other people were also working on this topic at this time (1949 – 1952) but were deterred by the extent of damage to the scalp, and their inability to create lesions deep in the brain when exposing trans-cranially [(15-18)]. Zubiani [(19)] managed to get lesions deep into the brain, but only by using a moving transducer and crossed beams, but he gives little detail about the exposures. He treated 30 patients with brain disorders using low intensities (0.6 – 1.5 W cm^{-2}) at 500 kHz and reported that 'subjective symptoms' disappeared with no concomitant cellular damage. Denier [(20)] also reported treating 3 patients with dementia, Parkinsonism and torticollis to 'some benefit'. Wall et al [(21)] used a focused beam to create lesions in cat and monkey brains through a skull opening.

Petter Lindström (Swedish neurosurgeon, married to the actress Ingrid Bergmann for a while) performed a trephine opening in the skull vaults of dogs and rabbits in order to gain direct access to the brain for 1 MHz ultrasound. This work was published in 1954 [(22, 23)]. They used cone shaped cups of different lengths and angles filled with saline to achieve focusing and to couple sound into the brain. Unsurprisingly, he found that the degree, extent and depth of the tissue damage was a function of intensity, exposure time and size of the sound beam at the entry point into the brain. He discovered that exposing through the dura mater resulted in smaller lesions than when it was turned back. Lindström reported that he could make graded, controllable lesions extending deep into the white matter with comparatively little damage to the cortex. He therefore concluded that the risks and complications arising from the use of ultrasound to produce functional changes to replace lobotomy clinically would be minor compared to operative lobotomy. The team went on to try this on 20 patients between May & October 1953, 17 of whom had intractable pain from inoperable cancers, exposing through a 1.5 inch (3.8 cm) 'button' opening in the calvaria. 1MHz ultrasound was used, with the output being described as 7 W cm^{-2} 'close to the crystal', but being focused using the cone shaped cup described above [(23)]. It is reported that there were no post-operative complications, and 10/17 had practically complete relief from pain over the 2 week – 11 month observation period, appeared to be 'more relaxed' and had 'better appetites'. Four more patients showed improvement with pain decreasing after treatment. The remaining three patients gained no relief from the treatment, although one was offered two more, after which pain relief resulted. 15 patients died from their cancers following (but not connected with) treatment, providing access to post mortem treated brain tissue. Apart from the expected minimal damage under the dural flaps that were lifted in the first patients treated, most patients appeared normal, with minimal damage, where seen, being in the white matter.

Lindström recognized the fact that it would be important to keep the head stationary while treating, and worked with his Swedish colleague Lars Leksell to produce a stereotactic device that would enable ultrasound treatment of psychiatric disorders. There is little published record of this, but in a 1951 paper, Leksell [(24)] shows his frustration with the need for a craniotomy, and moved on, developing stereotactic frame based radiosurgery, and later developing the gamma knife for which he is most famous (1967). Figure 4 shows the transducer and frame developed by Lindstrom and Leksell in 1949-1950. They successfully produced a system for creating periventricular lesions.

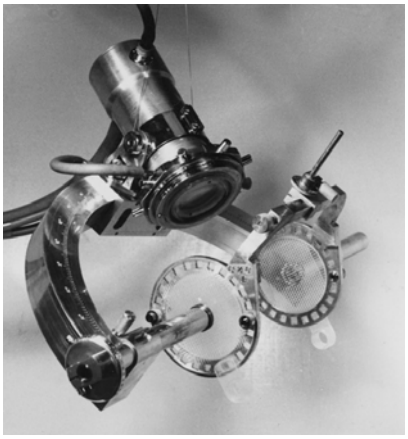

Fig. 4 Transducer head and stereotactic frame developed by Lindstrom & Leksell. (photo courtesy of Dan Leksell)

Lindström concluded that the methods he used needed further improvement. Although the required skin incision was not troublesome, the trephining was unpleasant, with patients complaining of the 'noise' and the 'shaking'. In a footnote, we are told that higher intensities (12 W cm^{-2}) had been trialed. In Lindström's opinion, a multiple beam system as described by Dussik [(25)] would only have restricted use in clinical neurosurgery since it would require several holes, or a large bone flap to be created. However, the same conclusion was not reached by the Fry brothers (William & Frank) working in the Bioacoustics Laboratory in Urbana, Illinois. After 5 years working with simple systems, they developed a treatment head that used four focused transducers mounted so that the beams overlapped at a common point, thus creating a high intensity region [(26)]. Quartz crystals, resonant at 1 MHz, fronted by polystyrene lenses were used. Coupling was achieved using physiological saline degassed by boiling. In their preclinical work, an acoustic window was created by removing a portion of bone overlying the target region. The animal's head was held in a standard stereotactic apparatus, and the depth of the focal region altered by vertical movement of the transducer mount. The system was tested on 31 cats. It was found that nerve fibres were damaged most readily, with nerve cells and blood vessels remaining intact, even in the focal zone. The paper describes the methods used to localize the focal region in the brain and to calibrate the acoustic output. This group has

Fig. 5 4-transducer head developed by the Frys in Illinois for use producing HIFU lesions in the brain

described an absorbing probe that can be used for mapping acoustic fields (see Section V below) [(27, 28)].

The Illinois team published prolifically on their research into the effects of ultrasound in the brain in the 1950's and early 60's [(10, 21, 26, 29-47)]. A publication in Science describes the lower threshold required to produce lesions in white matter (given as 51 atmospheres of acoustic pressure, acoustic particle velocity $4.8x10^3$ cm s^{-1}, 1 second exposure) than in grey (53 atmospheres of acoustic pressure, acoustic particle velocity $4.9x10^3$ cm s^{-1}, 2.5 second exposure) [(42)]. This was followed up a decade later by more very detailed studies in feline white matter which showed a linear relationship (with log(acoustic intensity) decreasing with increasing log(pulse duration)) in the threshold conditions for producing a lesion in the range $10^2 – 2.10^4$ W cm^{-2}, 7 -2.10^{-4} seconds at 1,3 and 4 MHz [(48)]. Here they took a closer look at possible mechanisms involved, and hypothesized that at the low intensity/long pulse end of the graph the mechanism was thermal, and at the high intensity, short pulse end it was predominantly cavitational.

Frank Fry outlined the devices used by the Illinois group in a detailed, and informative paper [(31)]. He lists the required components for a HIFU system as a means for focusing the sound waves, coupling medium, rigid mechanical supports and moving devices for the transducer, supporting structure for the specimens, electronic drives and precision timing devices. They were refreshingly rigorous in also requiring a calibrating and field plotting system. The first system built by the Fry brothers was said to be designed as a prototype for human neurosurgery, and was far from portable. They used a double layer room structure. The upper room contained the positioning and support system for the transducer and the power amplifier driving it. The animal support structure, and the electronic controls were all in the lower room. The co-ordinate system and the connection between the two rooms is described as being accurate and rigid, allowing placement accuracy of 'a few thousandths of an inch' in the brain.

A smaller system, which also included the 4-transducer head configuration, was built in Urbana and installed in the University of Iowa hospital. This too was designed to be very rigid, with the patient's head being held in position by 4 steel pins located in 'depressions' drilled into the skull. X-ray imaging was incorporated into the system in order to establish the co-ordinates of relevant structures within the brain. The 'replacement accuracy' is quoted as being a few thousandths of an inch, allowing the patient to return for treatment after a planning session. Coupling was achieved using an air inflated 'rubber gasket' between the bottom of the open pan in which the 4-transducer head is immersed and the patient's head [(49)]. This system was used to expose the ansa lenticularis or substantia nigra of Parkinson's patients to treat their tremor [(50)]. Its use to expose human pituitary glands for patients with breast cancer met with limited success [(46)]. A second generation, simpler, clinical device was set up in the University of Indiana University Medical School in 1970. This was fully computer controlled and had 5-degrees of freedoms for spatial coordinate control. It had the added novelty that it provided B-mode ultrasound images that were synchronized with the HIFU transducer for targeting and monitoring treatments. It was used for treatment of brain cancer, and was called the "Candy machine" after a young lady who had brain cancer and was treated by Dr. Heimburger [(51)].

At the same time as the Illinois group were working in this area, there were teams in Massachusetts following similar lines. Padmaker Lele (a neurophysiologist with a laboratory at MIT) described a modified system that used a single transducer for producing lesions in the cat brain. He discussed ceramic piezo-electric transducers, but dismissed them on the grounds of their temperature dependent acoustic power output, and consequent accuracy and stability of calibration. He recommended the use of plane X-cut quartz crystal plano-concave plastic lens combinations, the lens being attached to the silvered front face of the transducer by a uniform thin film of degassed castor oil and held in place by a retainer ring. Conical applicators of different lengths were attached to the transducer/lens combination to give different available depths of the focus in tissue. The applicator's open end was sealed with a rubber condom. Coupling was achieved through distilled water or normal saline that was boiled for more than an hour, and stored under a layer of degassed mineral oil until siphoned off for use [(52, 53)]. Lele used this system to perform extensive studies of the reproducibility of lesioning in plastics and in tissue.

At Massachusetts General Hospital (MGH) a team (comprising Ballantine, Ball (a neurosurgeon), Hueter, Bell, Nauta (a neuroanatomist), and Cosman) were also creating a HIFU delivery device [(54-58)]. They describe single transducer heads operating at 1 or 2.5 MHz. These frequencies were chosen to provide appropriately sized foci for the small animal studies they were conducting. The choice made in Massachusetts from the outset was to use a single transducer head comprising a quartz crystal fronted by a tuned steel plate, a plano-concave polystyrene lens and a conical water filled applicator. They used both frequencies to study paraplegia following the exposure of mouse spinal cords. A study was undertaken to investigate the ultrasound dose (here defined by the intensity and exposure time) required to produce damage, the correlation between dose and extent of damage, interdependence of intensity and time as measured by the paraplegia endpoint, and, importantly, the mechanism of action. It was found that damage was much more extensive at 1 MHz, which was attributed to the larger focal region at this lower frequency. The mechanistic studies indicated that when the

intensity exceeded 150 W cm^{-2} the damage was predominantly thermal in nature, whereas at lower intensities mechanical stresses appeared to be more important.

The studies with this system in mice led to work on larger mammals in 1960, with human exposures being the end goal [(56)]. A stereotactic device was developed. This combined a support for the subject (here, a cat), a counter-balanced column of a mobile X-ray set, a calibrated cross feed to allow vertical and lateral movement of the transducer and a commercial Horsley-Clarke machine (a stereotactic frame designed for making electrolytic lesions in the brains of large animals [(59)]). The radiofrequency generators and transducer head were developed by Bernard Cosman, who also worked on RF lesioning of the brain, and went on to start his own company, Cosman & Co., which became Radionics in the 1950's [(60)]. The frontal plane was targeted by moving the animal's support. It was still necessary to create an acoustic window by removing a portion of the overlying skull. This system was used to conduct a dosimetric study in cat brain, the exposure conditions for lesion production being found to be comparable to those used in the earlier mouse experiments. Presciently, one of the conclusions to their 1956 paper is that use of 'ultrasound lesions may offer a useful method for investigation of the nature of the blood brain barrier'[(57, 61)].

In the UK, a group working in the anatomy department of Guy's Hospital Medical School, also developed a system for studying HIFU lesions in the brains of large animals [(62-66)]. This was a collaboration between Roger Warwick (Professor of anatomy and one of the authors of the renowned Gray's Anatomy), researcher John Pond (physicist and inventor) and later Ken Taylor, research student. This team was the first to use a spherical segment (focused bowl) piezo-electric transducer to provide the acoustic field. In the reverse of the usual stereotactic procedure, the animal was placed on a milling machine that allowed its positioning with a quoted accuracy of a thousandth of an inch in 3 perpendicular planes relative to the transducer. Two different transducers were available, with focal lengths of 7.8 and 9.0 cm, and fundamental frequencies of 1 and 2 MHz respectively. Most of the work reported in their 1968 paper was conducted at 3 MHz (the third harmonic of the 1 MHz transducer, used because a 1 MHz transducer is more robust than its thinner 3 MHz counterpart). The focal position was determined using a thermocouple, its position then being indicated by a pointer fixed in relation to the transducer. In common with the other systems described above, the transducer head had a truncated Perspex cone attached to allow beam coupling to the brain. Both the Guy's and the MGH groups used goldbeater's skin to close these coupling cones. This is made from the gut of oxen, soaked in a dilute solution of potassium hydroxide, washed, stretched and beaten flat. This versatile membrane has been used to laminate gold in order that several sheets can be beaten at a time in producing gold leaf, and also, amongst other things is used for the repair of vellum, construction of early airships and the sealing oboe reeds [(67)]. In the ultrasound context, it has the attraction of being strong, elastic, hydrophilic and very thin (~25 μm), and so is better for transmitting high acoustic powers than, for example, latex.

Warwick & Pond describe in detail the microscopic appearance of the lesions obtained in the brains of cats and monkeys. They were able to confirm the 'island' and 'moat' appearance first described by Barnard et al [(10)]. While agreeing in the main with previously published histology (68) they found that whereas, for example, Bakay et al [(69)] had found that although there may be some changes in vessel walls, blood flow was not impaired, in these UK studies apparently intact blood vessels could be blocked.

The HIFU systems described above were typical of those used until the advent of phased array technology for high power applications [(70-79)]. This allows electronic movement of the focal volume without the need for mechanical translation of the transducer head. Similarly, the introduction of time reversal and other phase correction (adaptive focusing) techniques obviated the need for skull bone removal to provide an acoustic window, as judicious choice of signal phase and amplitude allows focusing behind a highly scattering medium [(80-86)].

B. Ophthalmology, cancer and other applications

Probably the first indication that high intensities of ultrasound can destroy specific ocular structures came in 1952 from Lavine et al [(87)] working at The Catholic University of America, Washington, DC, who demonstrated that cataracts were formed when the lens was targeted. Similar studies were carried out in Western Reserve University School of Medicine, Cleveland, Ohio by Purnell and colleagues [(88)]. In contrast to laser treatments in the eye, ultrasound effects are not reliant on absorption by pigmented structures, and can be focused at any point in the eye independently of the optical properties. It can be focused through the cornea and lens, through the conjunctiva and sclera, through a combination of these routes or onto the back of the eye. In the 1960's, the Ohio team used 3.5 and 7 MHz ultrasound to create circumscribed chorioretinal lesions and localized destruction of the ciliary body and suggested possible uses of this type of exposure in repair of retinal tear detachment, cyclodiathermy and destruction of intraocular tissue [(88-92)]. They used plane quartz transducers, 55 mm in diameter, with polystyrene lenses. The 3.5 MHz treatment head had an estimated maximum acoustic power at the focus of 900 W cm^{-2}, was in the form of a 'round bottomed cylinder' sealed by a polyethylene membrane, with the focal point capable of being positioned between 1 and 28 mm from the front.

For the 7 MHz head, the reach of the focal point is said to be 1-17 mm from the front of the truncated cone sealed with a vinylidene chloride membrane, and the maximum focal power is quoted as being 60 W cm^{-2}. 240 rabbit eyes were used in a study that demonstrated that selective chorioretinal lesions could be created [(90)]. Despite encouraging results, the team suggested that the danger of inadvertent cataract production and the prolonged exposure times made treatment of retinal detachment hazardous at that time.

The team best known for its work on therapy ultrasound and ophthalmology was a collaboration between the physicist Fred Lizzi at Riverside Research Institute, New York, and Jackson Coleman, an ophthalmic surgeon at Cornell University [(93-98)]. They worked on a number of conditions of the eye including the treatment of glaucoma, retinal detachment and vitreous haemorrhage, and founded the company Sonocare. The timing was perhaps unfortunate as laser techniques were being developed at the same time, with these being perceived as being simpler to use.

For many years, patients with Ménière's disease had limited options for treatment. Krejci applied ultrasound in an attempt to provide relief from the vertiginous attacks that are a symptom of this affliction [(99, 100)]. He used a narrow ultrasound beam to irradiate the vestibular portion of the inner ear, after doing an operation to expose the bony labyrinth. He eliminated the vestibular function, but preserved cochlear function. Michele Arslan, working in Padua took his technique further, using a narrow beamed 0.8 – 1.0 MHz ultrasound transducer to expose the bony wall of the lateral semicircular canal in order to destroy vestibular function [(101, 102)]. He took care to prevent lateral transmission of sound to the facial nerve by shielding, and heating was prevented by cooling. The technique appeared to resolve the problem of vertigo in 95% of patients, and so was taken up by a number of other centres [(103-114)]. In some cases, it was not possible to get sufficient intensity through the lateral semi-circular canal to get the require effect. A Bristol team under Angell James therefore developed a method of reducing the thickness of this structure to aid transmission.

Although there is now considerable interest in using HIFU (sometimes in conjunction with radiotherapy or chemotherapy) for the treatment of cancer, in the time frame under discussion here, there was little activity in this area. Probably the first mention of the use of ultrasound for the treatment of cancer was in 1933. Szent Györgyi from Budapest wrote a letter to Nature about an ultrasound exposure of 723 kHz; 'The effect of this radiation on Ehrlich's carcinoma has been studied by B Gözzi and found to have no specific effect on this tumour' [(115)]. The final sentence of this letter is dispiriting : 'For lack of funds our investigation has been broken off'. This discouraging view of ultrasound in cancer therapy (albeit, not strictly for HIFU) continued for more than 2 decades, despite some encouraging pre-clinical results from Carl Dittmar in Frankfurt [(116)]. The Congress on Ultrasound in Medicine in Erlangen, Germany in 1949 published a resolution (unsurprisingly, the 'Erlangen Resolution') which stated that 'Ultrasound is not suitable for cancer therapy and its clinical use should be discontinued'. Pohlman [(117)] performed an analysis of 133 clinical cases, finding improvement in only 17% of patients. Stuhlfauth [(118)] added to this stance, pointing out that, apart from Woeber's work combining X-rays with ultrasound therapy [(119)], '..the fact that …. tumours have never been successfully treated in human beings is due to the too low ultrasound intensities used which may also lead to an opposite effect'. He therefore called for people to 'refrain from sound treatment of tumors, since rapid growth of metastases was observed after sound treatment'. It is clear that there was little outside knowledge of research being carried out in the USSR in the 1950's, most probably because the publications were only in the Russian literature. An extraordinary man, Andrey Konstantinovich Burov, was head of the Laboratory of Anisotropic Structures, Academy of Sciences, and corresponding member of the Academy of Construction and Architecture

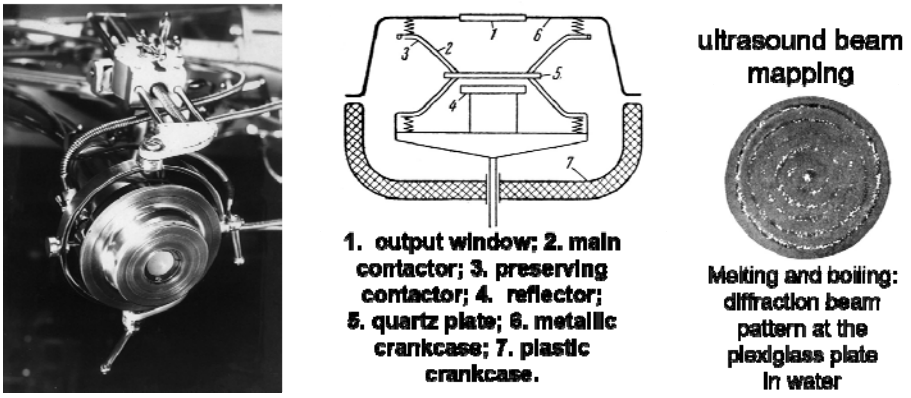

Fig. 5. System used by AK Burov & co-workers to treat Brown Pearce tumours in rabbits. (Photo courtesy of Burov archives)

of the USSR. At the end of a stellar career as an architect, where he worked with many of the greats of that era, including Le Corbusier, and he designed film sets for Eisenstein, he turned his attention to treating tumours with ultrasound. They used unfocused 1.5 MHz beams produced by quartz transducers, operating at 200 W cm^{-2} (continuous wave) (~ 500 W cm^{-2} in a pulsed regime) [(120, 121) (see Figure 5)]. They used the pulsed regime to treat Brown-Pearce tumours in rabbit testicles, finding that in 40-80% of cases, the tumour was resolved completely several months later, and a clear immune response was observed with metastases in the rabbit eyes disappearing [(122, 123)]. This was said to be a non-thermally induced response to the short (1-2 sec) exposure. The team also conducted clinical trials of melanoma treatments at the N.N.Blokhin Institute of Experimental Pathology and Therapy of Cancer. They treated 10 patients mainly in terminal stages of melanoma. Complete resorption of the tumour was seen in some, but not all, patients [(120)].

IV. HIFU TREATMENT PLANNING & MONITORING

In the 1950's and 60's pre-clinical HIFU was planned and monitored, if at all, using ultrasound imaging. The quality of the imaging was limited by the technology available in those years. Ultrasound lesions appear as hyperechogenic regions on a B-mode image. MRI was not proposed for guiding these treatments until 1992, but soon took off, especially for treatments in the brain [(83, 124-130)]. A major advantage of MR guidance of ultrasound therapies is the ability to overlay temperature rise maps on top of anatomical images. Ultrasound guidance of HIFU in patients is first mentioned in a paper by Heimburger in 1974 [(131, 132)]. Further descriptions of ultrasound guidance wait until the late 1990's with the advent of the ultrasound guided clinical devices being developed by groups in China, the UK, France and the USA for use in cancer, uterine fibroids and the prostate [(133-155)].

V. HIFU MODELLING & CALIBRATION

Gerald W. Willard and H.T. O'Neil, working at Bell Telephone Labs in New Jersey studied focused sources in some detail. In a 1949 seminal paper, O'Neil developed the theory for describing the field from a concave spherical transducer [(156)]. Despite only dealing with linear conditions, this is still used as the basis for field modelling by many people today. He was foresighted in suggesting that the results might be useful for design purposes and for correcting measured pressures and intensities to take into account the finite size of the measuring device. The solutions presented are for the distributions of pressure, particle velocity and intensity along the beam axis, but, perhaps more interestingly, also in the focal plane.

The source modelled is a concave spherical radiator with a circular boundary of diameter large relative to a wavelength (λ) and to the depth of the concave surface. Constant amplitude and phase are assumed, with the amplitude being sufficiently small that cavitation and non-linearity can safely be ignored. The pressure amplitude P_A at the centre of curvature is given by $P_A = \rho c u_0 k h$ where ρ is the mean density, c is the speed of sound in water, u_0 is the normal velocity of the surface, h is the depth of the concave surface, and k is given by $2\pi/\lambda$. This is close to the highest pressure in the field, except when kh is small. The relative pressure in the focal plane at an angle θ from the axis is the same as that at a large distance from a flat circular piston ($2J_1(ka \sin\theta)/ ka \sin\theta$) where J_1 is a first order Bessel function and a is the transducer radius. O'Neil notes that his theory matches Willard's experimental data [(157, 158)].

It is imperative that, when these high intensity focused fields are used for therapeutic benefit, they can be accurately characterised. Willard and Virginia Griffing & Francis Fox (from The Catholic University of America, Washington, DC) [(159)] were amongst the first to take this seriously, with both teams describing the use of the fountain obtained when the beam is fired through water on to an angled reflector, and thence to an air interface, to assess transducer output, as first described by Wood & Loomis [(1)].

Willard's 1949 paper [(157)] shows some of the first Schlieren images of focused fields, although probably the first reference to this method is in Richardson in 1940 (160) who in turn cites its description by Hiedemann & Osterhammel in 1937 [(161)]. While focused fields can be readily visualized using Schlieren techniques, this has only recently become quantitative [(162, 163)]. He also describes the use of highly attenuating, non-reflecting materials placed at the beam focus for investigating the beam. He refers to rubber, phenol fibre and methacrylate plastics (eg. Lucite) as being suitable. He described the appearance of 'little protuberances of melted material, and an 'odour of overheating' in rubber and phenol fibre. He was able to produce localized internal heating using plastics that were less absorbing. This builds on the work of the Columbia group who used paraffin blocks to demonstrate melting [(11)].

Willard also mentions, as an aside, that 'when a person's finger is placed at the focus, an input of >100 volts (1W) produces a sensation of burning, though none of the normal burn characteristics (redness or blistering….). He does not recommend this as a method of determining the focal position, providing the caveat that there may be considerable danger of causing 'serious internal injury' if the feeling of discomfort is ignored.

This paper also describes a simple radiation force balance [(157)]. This allowed what Willard termed 'fairly quantitative' estimation of the acoustic intensity. An absorbing target is mounted on a vertical arm mounted on to a horizontal bar resting on two supports. Two horizontal arms are attached to the axle supporting the vertical bar. These have moveable weights mounted on them. On one side is a counterweight whose position can be adjusted to give a zero (horizontal) setting in the absence of a sound beam. The other arm mounts a balance weight. The radiation force from a sound beam incident on the absorber upsets the equilibrium until the position of the balance weight is adjusted to bring the system back to the zero position. Hill [(164)] also describes this type of system. The insertion of a diaphragm with a hole between the target and the source allows assessment of, for example, the intensity at the focus. Figure 6. shows the radiation force balance used to perform acoustic power measurements at the Institute of Cancer Research. The reflecting stainless steel coated target (A), horizontal rod (B), masses (D) and counter-balance masses (C) are shown. The length of the vertical rod (l), mass displacement (x) and direction of the acoustic field are also seen. Measurements were made after first balancing the horizontal rod following submersion of the target in degassed water. On the horizontal bar, accurately known masses (m) were placed at measured distances from the pivot (l_1) in order to counter-balance the radiation force. The radiation force (F_{rad}) could then be calculated by equating moments around the pivot, $F_{rad}\ l_0 = m\ g\ l_1$ where the target is a distance l_o from the pivot. The conversion of force to acoustic power for an absorbing target is 69 mg/Watt [(54)]. The development of radiation force methods for the measurement of acoustic power and intensity is described in more detail elsewhere in this history: www.mpijournal.org/pdf/2021-SI-05/MPI-2021-SI-05.pdf pp519-536.

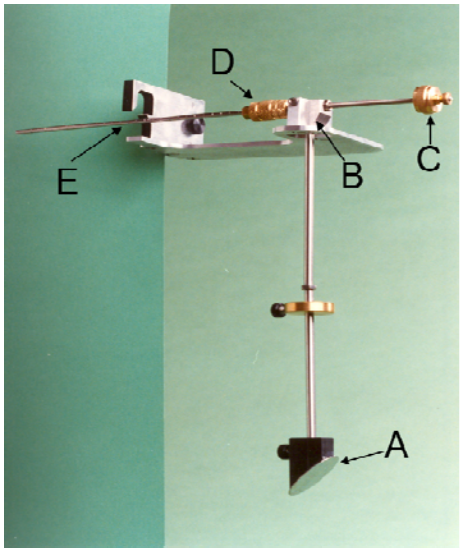

Fig. 6 Equilibrium balance for measuring radiation force

Safe and effective delivery of an ultrasound treatment also necessitates detailed knowledge of the pressure distribution. Fry & Fry described the use of thermocouples for this purpose [(27, 28)]. The probe consisted of a 12 μm wire thermocouple whose junction was imbedded in sound absorbing medium (a thin disc of castor oil) that matched the impedance of water. The oil is encased in a thin polyethylene membrane. The thermal emf is recorded on a galvanometer. They describe an initial steep temperature rise when the probe is in an ultrasound field, followed by an 'almost linear' portion of the curve. The steep rise is attributed to viscous heating due to the relative motion of the wire and the oil. These probes have been described in more detail in www.mpijournal.org/pdf/2021-SI-05/MPI-2021-SI-05.pdf p540. More recently thermistor probes coated in absorbent rubber have also been explored [(165)]. Modern beam plotting is more likely to use miniaturized ceramic piezoelectric probes or PVDF membrane hydrophones [(166)].

VI. Conclusions

Therapy ultrasound in general, and HIFU in particular has generated considerable interest with consequent potential applications in the 21st century. While it is generally recognized that this field has its roots in the 1940's and 50's, even before imaging ultrasound techniques, the intervening history is often overlooked. In many ways, current pre-clinical knowledge of HIFU and the way in which it interacts with tissue has not moved on far, but what has changed is technology. We now have the ability to image targets at depth within the body, to reconstruct a focus after transit through a strong scatterer such as the skull, and create ablated volumes using phased arrays. Used judiciously, for appropriate applications, there is no doubt that HIFU will continue to play a role in a number of medical specialties.

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Gail ter Haar is Professor of Therapy Ultrasound at The Institute of Cancer Research, London. She studied for her PhD in Physics at Guy's Hospital, in London, holds a D.Sc. in clinical medicine from Oxford. Gail's research interests have always lain primarily in understanding the way in which medical ultrasound interacts with tissue, especially the physical mechanisms involved in producing bio-effects (primarily heating & acoustic cavitation) with a view to understanding its safety when used in diagnosis, and to harnessing these effects for therapeutic benefit. Most recently her research has concentrated on developing devices and protocols for ultrasound based treatments of cancer.

Gail is founding President of the International Society for Therapy Ultrasound (ISTU). She is an honorary member of BMUS, honorary fellow of the American Institute for Ultrasound in Medicine, and fellow of the Acoustical Society of America and IPEM. She is Deputy Editor of "Ultrasound in Medicine and Biology", associate editor of "Ultrasonics" and on the editorial boards of International Journal of Hyperthermia and Medical Physics. Gail has written 5 books and 32 book chapters, and over 250 peer reviewed research papers.

THE DIASONOGRAPH STORY.

Tony Whittingham

APPENDIX

Sections *A* - *K* of this appendix contains copies of Diasonograph and related sales brochures and leaflets

Section *L* contains a list of the locations of archived Diasonographs in the UK

A. Nuclear Enterprises Ltd. Bulletin No. 327- Diasonograph NE4101. February 1970.

B. Nuclear Enterprises Ltd. Bulletin No. 64 - New Diasonograph NE4102 Diagnostic Ultrasonic Scanner. September 1972.

C. Nuclear Enterprises Ltd. Bulletin No. 88 - New Diasonograph NE4200 with Greyscale Storage Display. July 1976.

D. Nuclear Enterprises Ltd. Bulletin No. 434 - Ultrasonic Greyscale Facilities. Edinburgh. April 1975.

E. Nuclear Enterprises Ltd. Bulletin No. 90 - NE4104G Greyscale Storage Display Accessory for NE4102 Diasonograph Systems, July 1976.

F. EMI Medical Ltd. Bulletin No. 112 - EMISONIC 4200. October 1977.

G. EMI Medical Ltd. Bulletin No. 116 - EMISONIC 4201. January 1978.

H. Fischer Ultrasound Ltd. 4200S Console. High Resolution Imaging Combined with Maximum Operating Convenience.

I. Fischer Ultrasound Ltd. Articulated Scan Arm.

J. Fischer Ultrasound Ltd. MARTI.

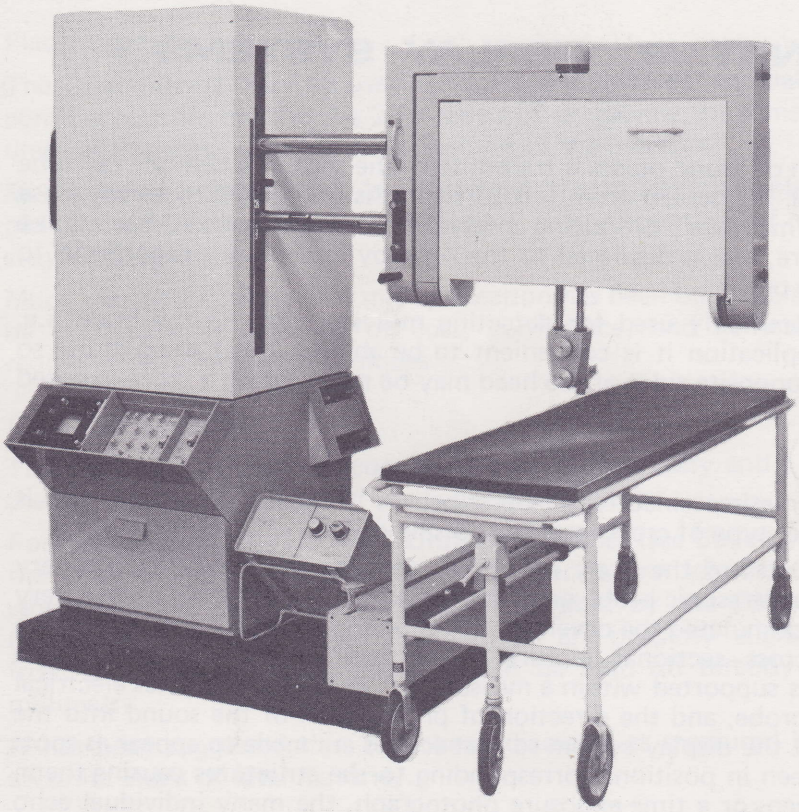
K. Fischer Ultrasound Ltd. LINUS.

L. Locations of archived Diasonographs in the UK.

A

Nuclear Enterprises Ltd. Bulletin No. 327- Diasonograph NE4101.
February 1970.

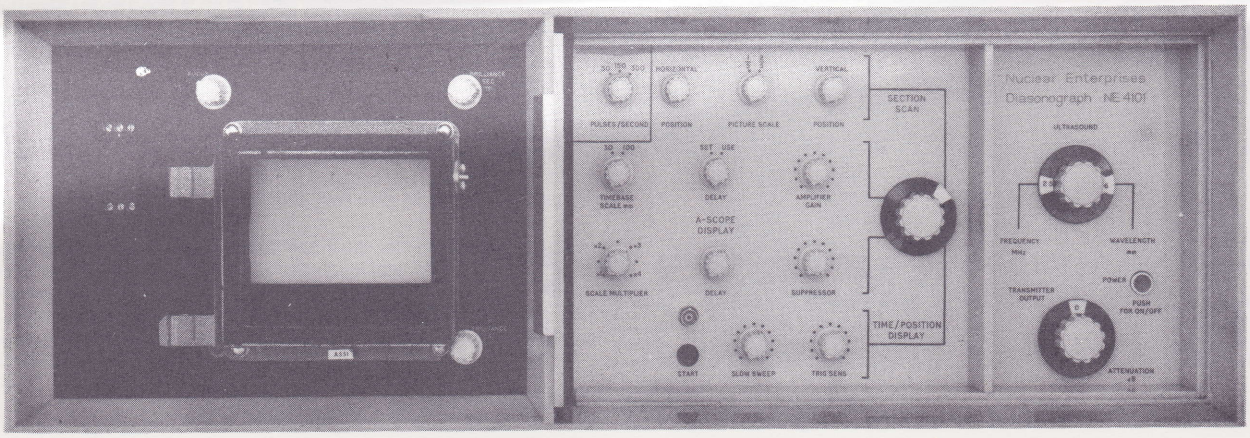
DIASONOGRAPH
NE 4101



for
ROUTINE
CLINICAL
DIAGNOSIS
in
Obstetrics
and
Gynaecology
and
for
RESEARCH
in
General
Medicine

The Diasonograph is a multi-purpose system for the investigation of soft-tissue structures in the body by pulsed ultrasonic waves. The operating principle is basically the same as that of underwater echo sounding or 'sonar'. However, the sound waves used have a much higher frequency, corresponding to a wavelength in tissues of 1 mm or less.

The ultrasound is emitted in short pulses, and echoes are detected from discontinuities in the tissues through which the sound pulses pass. Since the velocity of sound in soft tissue is known to be approximately the same from one tissue type to another, the elapsed time between transmission of a sound pulse and reception of the echo gives a measure of the distance of the tissue structure from the transmitting and receiving probe.



The Control Panel of the Diasonograph

THREE ALTERNATIVE DISPLAY SYSTEMS

A–Scope Display

Just before each of a rapid succession of sound pulses is transmitted, the spot of a cathode ray tube is made to start moving rapidly over the screen from left to right. As echoes are received, these cause vertical deflections of this 'time base', producing a series of vertical 'spikes'. Each spike represents a separate echoing structure, and is displaced to the right by a distance proportional to the distance of the structure from the probe.

This is the type of display most commonly used for detecting mid-line shifts in the brain, i.e. "echoencephalography". For this application it is convenient to be able to invert the picture, so that comparative echo patterns from opposite sides of the head may be recorded on a twice-exposed photograph of the screen.

Cross–sectional Display

(Note: Cross–sectional display is sometimes incorrectly described as 'B–scope' display, which is more properly a certain, rather limited, type of cross–sectional display).

Echo patterns received from most sites of the body are rather complex, and may change very rapidly with small movements of the ultrasonic probe or of the patient. Interpretation is often very difficult, if not impossible. This has stimulated the development of pictorial display systems, which present the echo information as a cross–sectional 'picture' of the tissues being examined. In the Diasonograph, the ultrasonic probe is supported within a measuring frame. This generates electrical signals defining the position of the probe, and the direction of propagation of the sound into the body. These electrical signals control the display system so that echoes are made to appear as spots of light on the cathode ray tube screen in positions corresponding to the structures causing them. By the use of a long-persistence screen, or a time-exposure photograph, the many individual echo patterns produced as the probe is moved round the body surface, can be summated to yield a cross–sectional picture of the soft tissue structures being investigated.

By this means it is possible, for example, to visualise the gestation sac only a few weeks after conception, to locate the placenta and define its precise margins with a great deal of confidence, and to carry out many other diagnostic investigations on soft tissues which have no counterpart in conventional radiological, or isotopic techniques.

Time/Position Display

Many soft tissue structures of the body move, for example, with the heartbeat, or with respiration. In some instances the form of this movement has diagnostic significance.

The best-known examples are possibly the movement pattern of the mitral valve as an index of the severity of rheumatic heart disease, and the separation and synchronous movement of the pericardium and myocardium which reveals the presence and degree of pericardial effusion.

The Time/Position display system utilises the horizontal timebase of the A–scope system and presents the echo signals as bright spots on this line, rather than as deflection 'spikes'. At the touch of a button, the whole trace moves, relatively slowly, from the bottom to the top of the screen. Echoes from stationary structures thus trace out straight vertical lines, while structures which are moving trace out the pattern of the movement in great detail.

The speed of the slow vertical sweep is adjustable over wide limits, enabling the operator to choose a speed best suited to the time scale of the phenomenon he is investigating.

APPLICATIONS

A Summary

Without doubt the principal routine clinical applications of the Diasonograph lie in the obstetric field, though the practical significance of the instrument in gynaecology, and the investigation of the liver, spleen, and renal disorders, should not be overlooked. The following is only a brief summary of the main applications which have already been extensively investigated.

OBSTETRIC APPLICATIONS

Early Pregnancy

Positive confirmation of pregnancy from six weeks amenorrhoea onwards, diagnosis of extra-uterine pregnancy, multiple pregnancy, blighted ovum, level of nidation, and incomplete abortion.

Positive diagnosis of hydatidiform mole, and, more significantly, exclusion of this possibility in cases of threatened abortion.

Placentography

The Diasonograph may be used to locate the placenta, and define its margins with a high degree of confidence from around the 28th week of pregnancy. In some cases it may be possible to observe unusual thickness, or other abnormalities of the placenta.

Much earlier in pregnancy, 14–16 weeks, the approximate placental site may be identified, but a positive location and the firm exclusion, or confirmation, of placenta praevia should await a later stage of pregnancy.

Much clinical experience of this application has been built up, and the ultrasonic method appears to be preferable on grounds of cost, confidence, and general convenience to either soft-tissue radiography or isotope techniques.

Fetal Cephalometry

The fetal head is readily recognised by ultrasonography and a refinement of the technique permits the accurate measurement of the bi-parietal diameter.

For this purpose, the NE 4131 Ultrasonic Caliper (see Bulletin No. 284) is used in conjunction with the Diasonograph. The bi-parietal diameter is first identified by cross–sectional scanning, after which the two bright marker spots generated by the Caliper are superimposed on the display of the echo pattern. When the markers are properly aligned with the entry and exit echoes from the fetal head, the diameter can be read off directly to an accuracy of better than one millimetre.

It has been suggested that fetal head diameter as measured in this manner, is a very reliable and accurate index of fetal maturity.

General Obstetric Applications

Determination of fetal presentation in the obese patient or otherwise difficult situation. Diagnosis of hydramnios, anencephaly and hydrocephaly. Investigation of multiple pregnancy and the determination of the relative sizes of the fetal heads. Avoidance of the placenta in amniocentesis. Investigation of all cases of antepartum haemorrhage. Determining co-existence of pregnancy and pelvic tumour.

GYNAECOLOGICAL APPLICATIONS

Diagnosis of ovarian cyst, and determination of whether unilocular or multilocular. Differential diagnosis of ovarian cyst, fibroid, and other pelvic tumours. Differentiation of different types of ascites. Study and management of retention of urine.

GENERAL MEDICAL APPLICATIONS

Investigation of hepatic tumours, and hepatomegaly. Investigation of splenomegaly. Demonstration of hydronephrosis, and diagnosis of polycystic disease. Abdominal abcesses and hematomata. Demonstration of abdominal aortic aneurysm by cross–sectional scanning and time/position display. Demonstration of Pericardial effusion and mitral stenosis by time/position display. A–scope "echoencephalography".

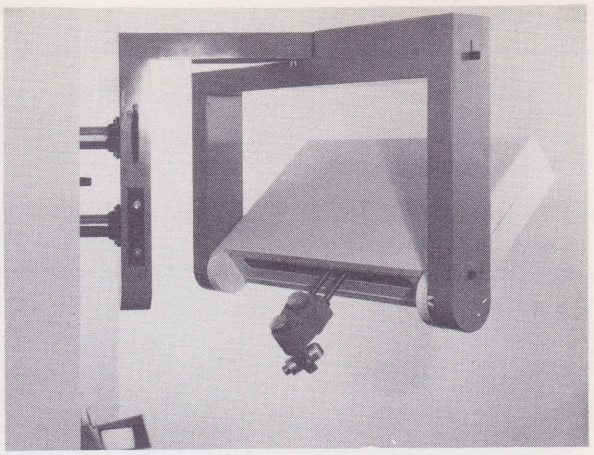
MECHANICAL FACILITIES

In many applications the diagnostic information can only be obtained if the plane of the cross-sectional scan passes through a particular small region. Also there are instances when the inclination of this plane relative to the tissue structures is critical.

A unique and very important feature of the Diasonograph is the extreme degree of freedom of adjustment of the measuring frame — and hence the cross-section examined — relative to the patient. This is achieved by the combination of three linear, and two angular movements. The whole measuring frame can be rotated about a vertical pivot from the transverse to the longitudal direction, and may be locked in position at 5° steps. Further the plane of the scan may be tilted out of the vertical, and also locked at 5° steps (see illustration).

Calibrated scales are provided for each of these five degrees of freedom of adjustment, so that the position of any particular scan may be recorded and repeated.

It is this convenience and stability of this freedom of adjustment of the plane of the scan which accounts for much of the versatility and clinical utility of the Diasonograph.



SPECIFICATION

Dimensions

Overall Height	: 202cm (79.5in)
Maximum width without couch	: 108cm (42.5in)
Overall Length of measuring frame assembly and supporting guide tubes	: 233cm (91.7in)
Height of couch top	: 76cm (30.0in)
Width of couch top	: 59cm (22.0in)
Overall Length of couch	: 198cm (78.0in)
Gross weight, inclusive of packing for transit by air	: Approx 800kg (1760lb)

Power Requirements

Suitable for use on ac mains of 50 or 60Hz. (Frequency and operating voltage should be specified at time of ordering). Power consumption approx 600 watts.

Photographic Recording

An NE 4135 camera is supplied as part of the Diasonograph system. This has an object/image ratio of 1:0.7. The camera uses Polaroid type 107 film packs which give 8 exposures 83 x 108mm (3.25in x 4.25in).

Ultrasonic Probes

3 special focussed probes are provided as part of the Diasonograph system. These operate at 1.5MHz (NE 4161), 2.5MHz (NE 4162) and 5MHz (NE 4163). In addition, hand-held, non-focussed "low-noise" single transducer probes are available at extra cost.

Recommended Operating Frequencies

For most abdominal work, 1.5MHz and 2.5MHz will be found to give best results. 5MHz may be used for experimental investigations but its use is generally confined to the examination of tissue structures close to the skin surface.

Transmitting System

The electrical input to the transducer does not exceed 0.001 joules per pulse.

A stepped attenuator on the output of the transmitting system provides 0-45dB attenuation in nominal 5dB steps.

CROSS-SECTIONAL DISPLAY ("Section Scan")

The picture scale, as measured on a finished print from the NE 4135 camera may be selected as $\frac{1}{5}$ or $\frac{2}{5}$ of full size.

The measuring frame allows the transducer freedom of measured movement within a rectangle 50 x 25cm (20 x 10in approx). The transducer may be rotated through $\pm 100^{\circ}$ from the medial position.

The display system will follow any scanning pattern chosen by the operator, though the most useful type of scan is usually the "compound sector scan" described in the literature.

A long-persistance cathode ray tube is provided for visual monitoring. This reproduces exactly the information on the photographed display, but the 'afterglow' effect enables the operator to carry out a systematic visual search for particular sections which contain data of diagnostic significance.

Special circuits are incorporated in the Diasonograph which enable echo signals of widely-differing amplitudes to be recorded on the same display. The signal processing circuits also decrease the apparent pulse length, thus improving the resolving power of the equipment, and preventing small signals following closely on larger ones from being lost.

A-SCOPE DISPLAY

Time base ranges are 30mm and 100mm nominal, in soft tissue for full-scale range, with a continuously adjustable range multiplier from X1 to X5. The start of the timebase may be delayed relative to the instant of transmission of the sound pulse enabling the operator to study in detail the echo pattern obtained from a small region some distance from the body surface. A delay "Set/Use" switch enable the operator to pre-select the region to be examined.

The A-scope display is normally arranged to occupy half the available screen height, and may be inverted by the function selector switch.

TIME/POSITION DISPLAY

The slow-speed vertical timebase is initiated by a push button, or an external signal, and causes the horizontal, intensity-modulated, trace to move at an adjustable speed from the bottom to the top of the screen.

SWEPT GAIN SYSTEM

A swept gain system is incorporated which compensates both for the sensitivity-versus-range characteristics of the probes in use, and also for the progressive weakening of echo signals from tissue structures remote from the probe.

Not all applications require the same swept gain conditions and for the convenience of the user, four separate sets of pre-set swept gain controls are provided. These are selectable by a four-position switch. As supplied from the factory, the four positions give progressively increasing degrees of swept gain compensation. However, a user may modify these settings as he sees fit for his particular clinical or research applications.

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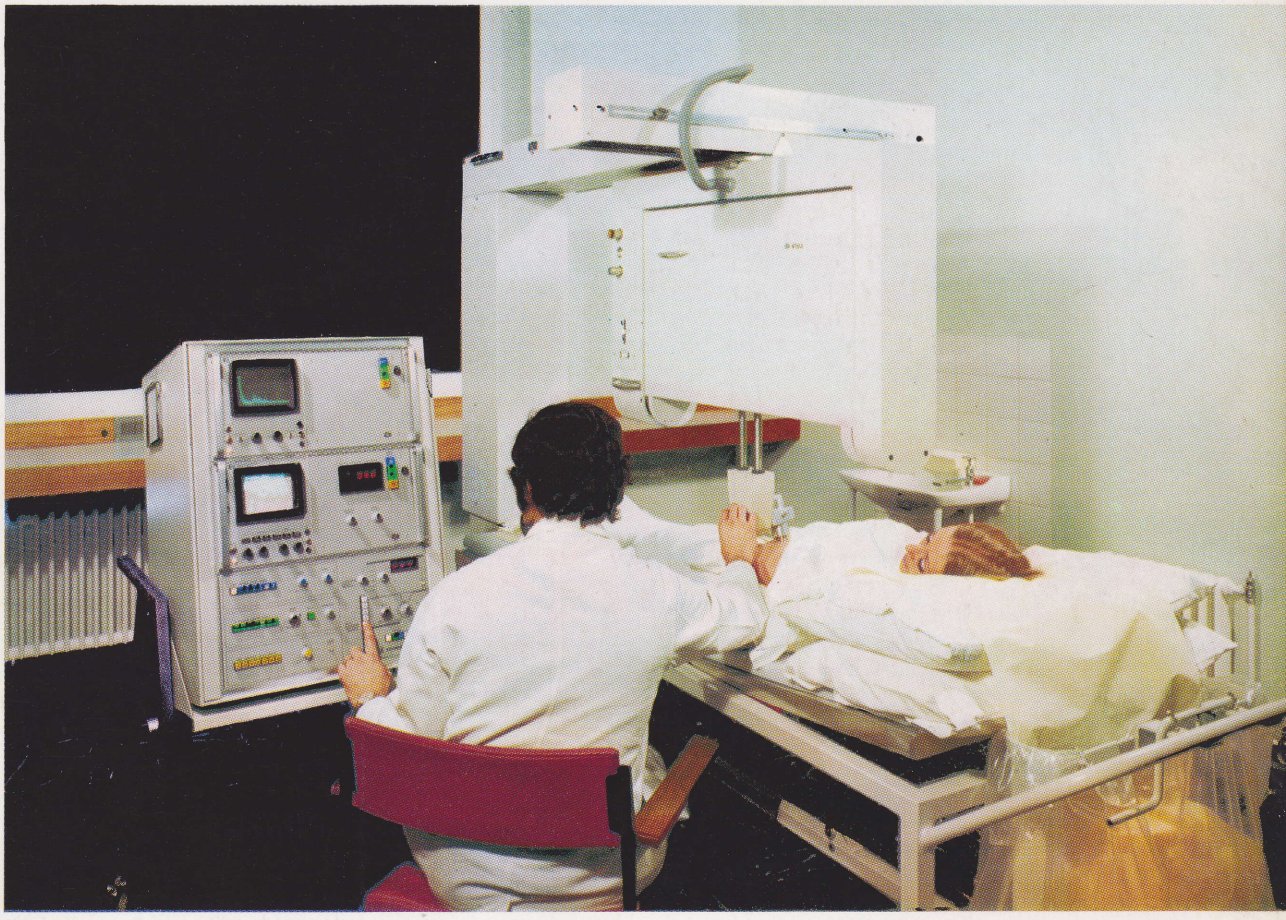
B

Nuclear Enterprises Ltd. Bulletin No. 64 - New Diasonograph NE4102 Diagnostic Ultrasonic Scanner. September 1972.

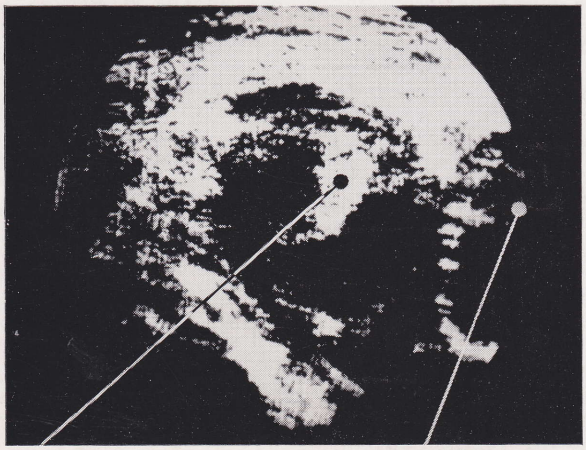
NEW

DIASONOGRAPH

NE 4102 Diagnostic Ultrasonic Scanner



Diasonograph NE 4102 in operation at Queen Mother's Hospital, Glasgow



FOETUS BLADDER

Longitudinal cross-sectional picture of nine week old foetus (Courtesy Queen Mother's Hospital, Glasgow)

- ★ Very high resolution combined with fine control, for detailed examination of small structures
- ★ Twin cathode-ray tube displays for simultaneous search and identification of structures
- ★ Wide choice of scan planes, simply selected and extremely stable
- ★ A-scan, Cross-sectional and Time/Position Displays
- ★ Integrated circuitry, for high electrical stability, accuracy, and extreme reliability
- ★ Caliper System with large-scale direct digital readout
- ★ Compact and flexible, permitting considerable freedom in patient examination
- ★ Designed for a single operator
- ★ Wide range of Obstetric, Gynaecological and General Medical applications
- ★ Lowest possible ultrasonic output

DIAGNOSIS BY ULTRASOUND

The new Diasonograph NE 4102 has been developed and evaluated by groups of workers whose involvement in diagnostic ultrasonic scanning equipments extends over the past twenty years. Typical of the thought and expertise designed into the NE 4102 are the methods used to ensure that the ultrasonic power output is reduced to the lowest possible level. Firstly, for fixed pulse rate applications the number of pulses transmitted per second is held at 600. Some equipments use pulse repetition rates up to three or four times higher than this. Secondly, during searches, a unique velocity controlled pulse repetition system ensures that transmission only takes place whilst scanning is being carried out and at a rate controlled by the speed of scanning. Thirdly, the sensitivity of the equipment is controlled by reducing the transmitted power rather than using high power and reducing the gain of the system.

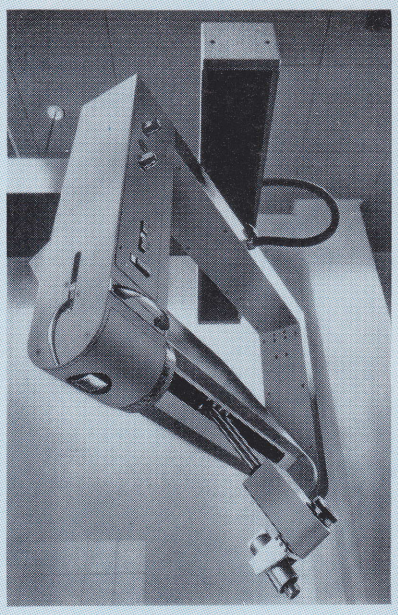
Some features can be considered traditional. The NE 4102 measuring system has long been shown to have exceptional freedom of positioning of the scanning plane, whilst maintaining extremely tight control of the chosen scanning plane. The one-to-one measuring system directly coupled to the scanning probe has unequalled measuring accuracy which contributes to the outstanding ability of the NE 4102 to find and resolve structures smaller than hitherto possible.

Very comprehensive display facilities and a wide choice of operating modes enable the new Diasonograph to cover a wide field of applications. The controls are simple. Colour coding is used to indicate function, and illumination to indicate choice (see specifications p. 6).

Most investigations carried out by ultrasound come within three main groups:

1. Accurate measurement of structural dimensions in vivo, e.g. Foetal Cephalometry—*A-scan*
2. Visualisation of a structure, e.g. Placental Localisation—*Section Scan*
3. Visualisation of moving structure, e.g. Mitral Valve Investigation—*Time/Position Scan*

It is often necessary to observe in two modes simultaneously, as in Foetal Cephalometry. With the new Diasonograph, accurate measurements can be made on the A-scan Display of echoes identified on the Section Scan Display.



Positioning of the scanning plane is achieved with push-button controlled magnetic locks (Courtesy Design Magazine and Philip Sayer)

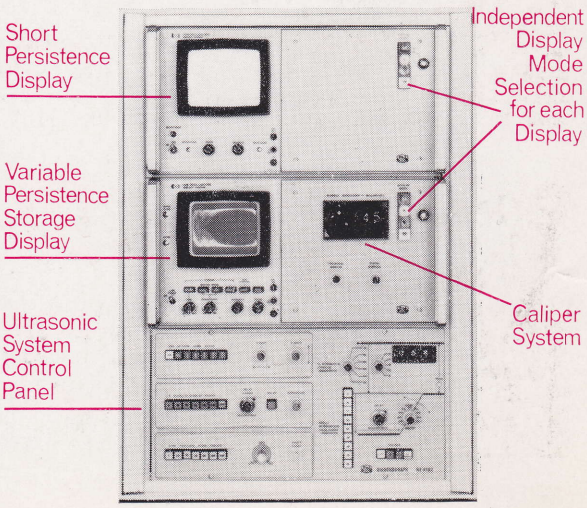
PRINCIPLE OF OPERATION

The technique for investigations of soft tissue structures in the body by pulsed ultrasonic waves is basically the same as that employed in underwater echo-sounding or sonar. However, the sound waves used have a much higher frequency, corresponding to a wavelength in tissue of 1mm or less. The ultrasound is emitted in short pulses, and echoes are detected from discontinuities in the tissues through which the sound pulses pass. Since the velocity of sound in soft tissues is known to be approximately the same from one tissue type to another, the elapsed time between transmission of a sound pulse and reception of the echo gives a measure of the distance of the tissue structure from the transmitting and receiving probe.

Display Systems

The NE 4102 has two display oscilloscopes which are interchangeable. Display of section scan, A-scan or time/position scan may be presented independently on either of the two units (see illustration right).

The upper display unit consists of a short persistence oscilloscope. The lower incorporates a variable persistence oscilloscope and Caliper System, which uses a large scale illuminated digital readout. This layout permits unobscured viewing of both displays and digital readout while the caliper settings are being adjusted. This is the standard arrangement but it can be modified to suit individual requirements. For specification details see page 6.



Electronics console of UE 4102

THREE ALTERNATIV

A-Scan Display

A-scan display is most commonly used for detecting mid-line shifts in the brain (echo-encephalography). For this application it is convenient to be able to invert the picture so that comparative echo patterns from opposite sides of the head may be recorded on a twice-exposed photograph of the short persistence screen, or on the variable persistence oscilloscope.
When the echo comparison is not required, the full screen is used to make accurate location of the caliper markers easier.

Just before each of a rapid succession of sound pulses is transmitted, the spot of a cathode-ray tube is made to start moving rapidly across the screen from left to right. As echoes are received, the time base is deflected producing a series of vertical 'spikes'. Each spike represents a separate echoing structure.

Cross-sectional Display

With the new Diasonograph's cross-sectional display facility, it is now possible to visualise the gestation sac only a few weeks after conception. The placenta can be located and its precise margins defined with a great deal of confidence. Many other diagnostic investigations on soft tissues, which have no counterpart in conventional radiological or radioisotope techniques, can be carried out.
Formerly, because of the complex nature of the echo patterns received from most sites on the body, and lack of stability of probe positioning on the patient, interpretation of results was very difficult, if not impossible. Now, cross-sectional pictorial display is combined with the very high system resolution of the new Diasonograph and an extremely accurate and stable measuring frame. Consequently, the pictures of small structures obtained with the NE 4102 are exceptionally detailed.

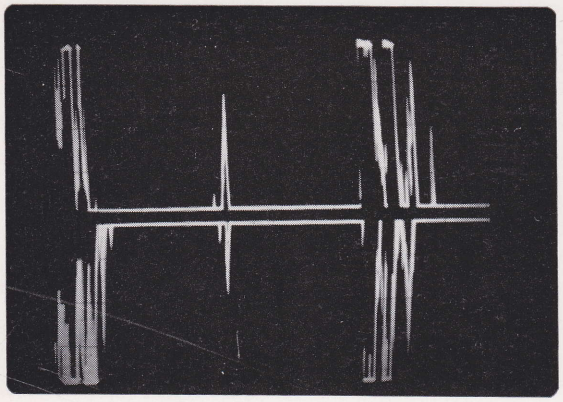
In the new Diasonograph, the ultrasonic probe, supported within a measuring frame, generates electrical signals defining the position of the probe and the direction of propagation of the sound into the body. These electrical signals control the display system so that echoes are made to appear as spots of light on the cathode-ray tube screen in positions corresponding to the structures causing them. By the use of a long-persistence screen, or a time-exposure photograph, the many individual echo patterns produced as the probe is moved round the body surface, can be integrated to yield a cross-sectional picture of the soft tissue structures under investigation.

Time/Position Display

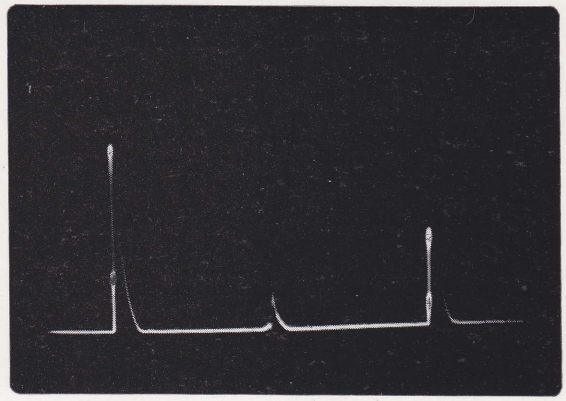
Many soft tissue structures of the body move, for example with the heartbeat, or with respiration. The form of this movement can have diagnostic significance. For example, the movement pattern of the mitral valve gives an indication of the severity of rheumatic heart disease, and the separation and synchronous movement of the pericardium and myocardium can reveal the presence and degree of pericardial effusion.
In the Time/Position display mode, the pattern of movement of the structure under examination is shown. The vertical sweep velocity is adjustable over wide limits permitting the operator to select conditions best suited to the particular investigation.

The Time/Position display system uses the horizontal timebase of the A-scan system and presents the echo signals as bright spots on this line, rather than deflection 'spikes'. The whole trace can be made to move relatively slowly from the bottom to the top of the screen. Echoes from stationary structures trace out straight vertical lines, while structures which are moving trace out the pattern of their movement.

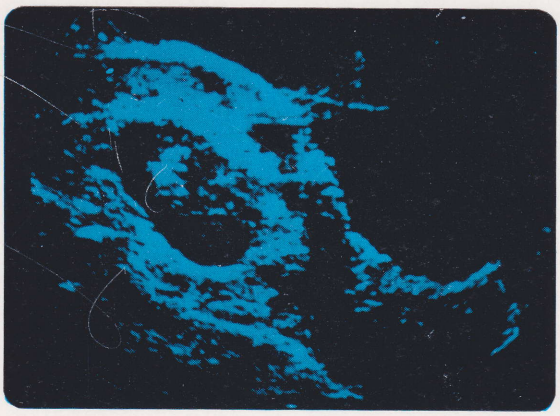
'E DISPLAY MODES



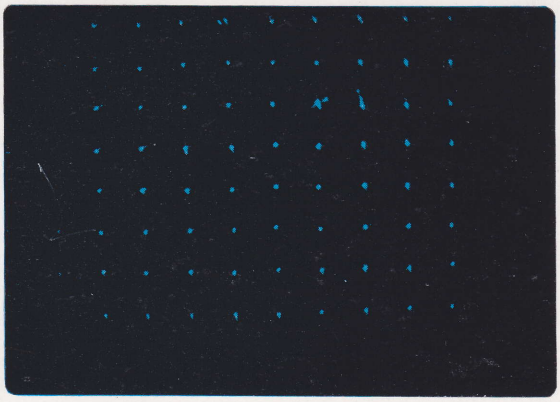
Echoencephalography Trace



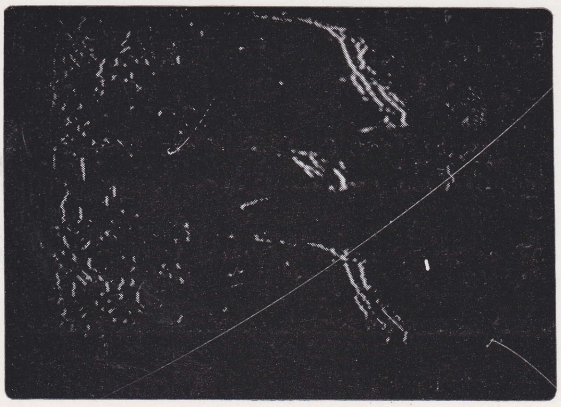
Foetal Cephalometry Trace



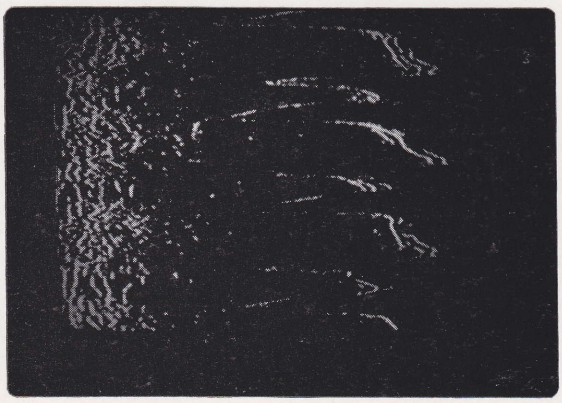
Longitudinal cross-section of an early gestation sac (Courtesy Queen Mother's Hospital, Glasgow)



Cross-sectional picture of a matrix of wires spaced at 0.5in (12.7mm) intervals in a water tank



Time Position Trace showing the movement of a mitral valve



Time Position Trace of identical mitral valve showing larger number of heartbeats

WIDE RANGE OF APPLICATIONS

Obstetric Applications

Positive confirmation of pregnancy from six weeks amenorrhoea onwards, diagnosis of extra-uterine pregnancy, multiple pregnancy, blighted ovum, level of nidation and incomplete abortion.

Positive diagnosis of hydatidiform mole, and, perhaps more important, exclusion of this possibility in cases of threatened abortion.

Placentography

The Diasonograph may be used to locate the placenta and to define its margins with a high degree of confidence from around the 28th week of pregnancy. In some cases it may be possible to observe unusual thickness, or other abnormalities of the placenta.
Much earlier in pregnancy, 14-16 weeks, the approximate placental site may be identified, but the positive location and the exclusion or confirmation of placenta praevia should await a later stage of pregnancy.
Much clinical experience of this application has been built up, and the ultrasonic method is often preferred on grounds of cost, confidence, safety, and convenience to either soft-tissue radiography or radio-isotope techniques.

Foetal Cephalometry

The foetal head is readily recognised by ultrasonography and a refinement of the technique permits the accurate measurement of the bi-parietal diameter. The bi-parietal diameter is first identified by cross-sectional scanning, after which the two bright marker spots generated by the Caliper are superimposed on the display of the echo pattern. When the markers are properly aligned with the entry and exit echoes from the foetal head, the diameter can be read off directly to an accuracy of better than one millimetre. It has been suggested that the measurement of foetal head diameter obtained in this manner, is a very reliable and accurate index of foetal maturity.

General Obstetric Applications

Foetal presentation may be determined in the obese patient and in otherwise difficult situations. Diagnosis of hydramnios, anencephaly and hydrocephaly may also be made. Other applications include the investigation of multiple pregnancy and the determination of the relative sizes of the foetal heads, the avoidance of the placenta in amniocentesis, investigation of all cases of antepartum haemorrhage, and determining the co-existence of pregnancy and pelvic tumour.

Gynaecological Applications

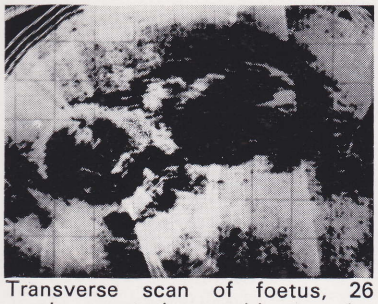
Diagnosis of ovarian cyst, and determination of whether unilocular or multilocular. Differential diagnosis of ovarian cyst, fibroid, and other pelvic tumours. Differentiation of different types of ascites. Study and management of retention of urine.

General Medical Applications

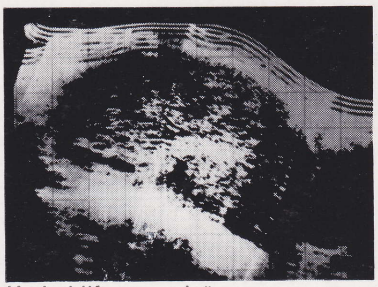
Investigation of hepatic tumours, and hepatomegaly. Investigation of splenomegaly. Demonstration of hydronephrosis, and diagnosis of polycystic disease, abdominal abcesses and hematomata. Demonstration of abdominal aortic aneurysm by cross-sectional scanning and time/position display. Demonstration of pericardial effusion and mitral stenosis by time/position display. A-scan 'echoencephalography'.

TYPICAL RESULTS

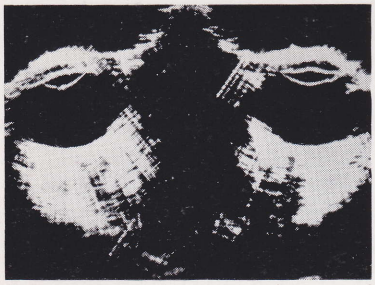
Transverse scan of foetal head at 36 weeks gestation*



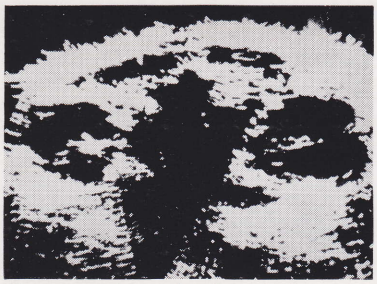
Transverse scan of foetus, 26 weeks gestation, with anterior placenta*



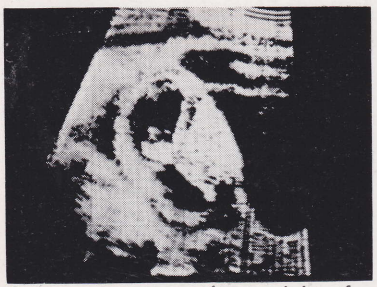
Hydatidiform mole*



Cross-sectional scan of normal eyes†



Cross-section of infant kidneys†



Early pregnancy (see right for enlarged picture of gestation sac)†

Life-size picture of embryo†

References

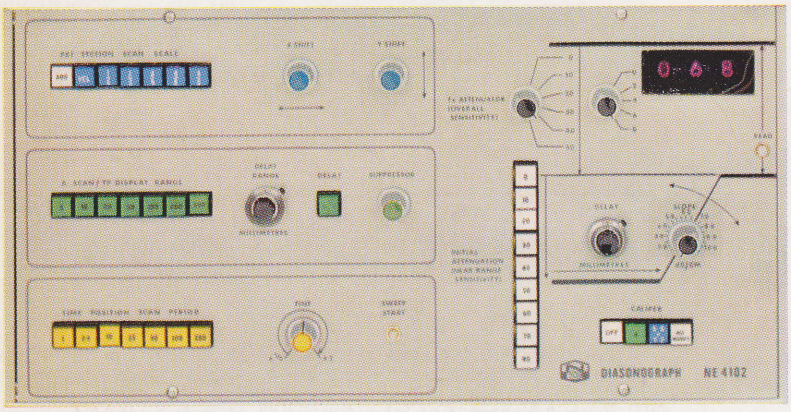
S. Campbell, J. Obstetrics and Gynaecology of Brit. Com., **75**, *p. 568, 1968.*
I. Donald and U. Abdulla, J. Obstetrics and Gynaecology of Brit. Com., **75**, *p. 993, 1968.*
P. Morley and E. Barnett, Brit. J. Radiology, **43**, *p. 602, 1970.*
E. Barnett and P. Morley, Brit. J. Radiology, **44**, *p. 733, 1971.*
E. A. Lyons et al, Brit. Med. J., p. 2689, 1972.
R. A. Mountford et al, Brit. J. Radiology, **44**, *p. 860, 1971.*
R. A. Underhill et al, Brit. Med. J., p. 3736, 1971.
S. Campbell and C. J. Dewhurst, Lancet, p. 1002, 1971.
I. Donald et al, J. Obstetrics and Gynaecology of Brit. Com., **79**, *p. 304, 1972.*
Investigations into Possible Chromosome Damaging Effects of Ultrasound, Brit. J. Radiology, **45**, *p. 319, 1972.*

* Courtesy St Bartholomew's Hospital, London
† Courtesy Queen Mother's Hospital, Glasgow

NE 4102 SPECIFICATION

Two cathode-ray tube display units are provided, one with a short persistence cathode-ray tube, and the other with a variable persistence storage type tube. The short persistence tube is most useful for the photographic recording of section scans, where the optimum system resolution is required, or for foetal cephalometry with the Caliper. The variable persistence tube is most valuable when 'searching' during a section scan, since it is possible to vary the fade on the tube, and having located the required place of scan, to store the required picture. Photographs of the stored picture may be taken if required. A-scan, cross-section or time/position display may be set up independently on either display unit.

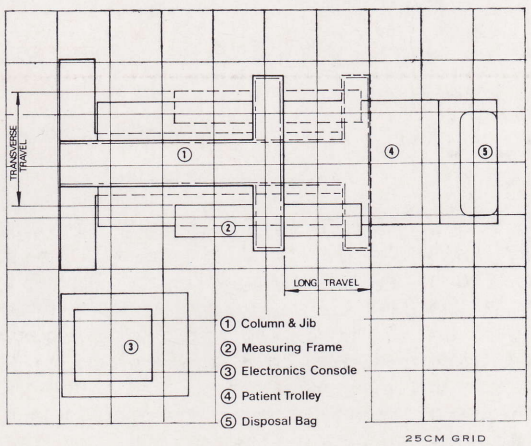
Operating controls are distributed in ergonomic groupings to assist operators without specialised knowledge of physics or electronics. Operating controls are calibrated in cm (or mm) in tissue where appropriate (see Control Panel below).

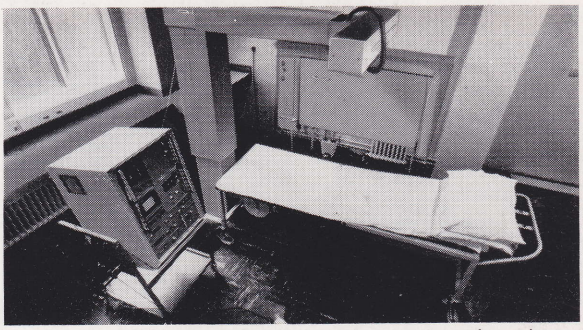


Operating Frequencies	0.5, 1.5, 2.5, 5 and 10 MHz
Energy	Maximum available ultrasonic intensity is a function of the probe in use. Using the standard range of probes the following figures are typical.
Normal Mean Acoustic Power	1.5MHz – 0.70 microwatt 2.5MHz – 6.92 microwatt 5.0MHz – 84.0 microwatt
Maximum Available Mean Acoustic Power	1.5MHz – 7.0 milliwatt 2.5MHz – 6.92 milliwatt 5.0MHz – 2.7 milliwatt
	Note Maximum operating sensitivity of the equipment is controlled by adjustment of the output power.

Transmitter Pulse Repetition Rate	(a) Fixed PRF 600 pulses/second (b) Velocity Controlled PRF (Section Scan only) 60 to 1000 pulses/second
Sensitivity Control by Transmitter Attenuator	0 to 50db in 10db steps + 0 to 8db in 2db steps
Swept Gain	Initial reduction: 0 to 80db in 10db steps Initial delay: 0 to 25cm continuously variable Slope: 1.5 to 7.5db/cm continuously variable
Picture Scales (Cross-Section only)	1/5, 2/5, 3/5, 4/5, or 5/5 of full scale
Scanned Area (Cross-Section only) measuring frame in vertical position	Horizontal: 20in (500mm) nominal Vertical: 10in (250mm) nominal Probe rotation: ±100° from vertical in the plane of scan
Cross-Section Display	Intensity modulation of compound scan pattern on a 7.5 x 9.9cm display area.
A-Scan Display	A-scan display occupying half or whole available screen height. (Selected by preset switch.) Inversion by panel switch to enable accurate photographic comparison to be made between two successive echo patterns.
A-Scan Range	5, 10, 20, 50, 100, 200 and 500mm switched steps.
A-Scan Delay	0 to 500mm continuously variable A delay set/use switch allows the operator to preselect the region to be examined.
Time/Position Display	Intensity modulation of horizontal sweep.
Vertical Sweep (Time/position Display only)	Single vertical sweeps triggered by push button. Sweep speed continuously variable between x ½ and x 2 of the following switched ranges: 1, 2.5, 10, 25, 50, 100, 250mm/second
Horizontal Range and Delay (Time/position Display)	A-scan controls operative.

Plan of Layout





In the new Diasonograph, maximum use has been made of floor space without any sacrifice in the facility of selecting the more difficult scanning planes. The construction of the scanning system and the control cabinet affords considerable **freedom of choice in the layout** of the examination room. Total system weight is 650kg. (Photo courtesy Design Magazine and Philip Sayer)

NUCLEAR ENTERPRISES LIMITED

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Bulletin No. 64
September 1972

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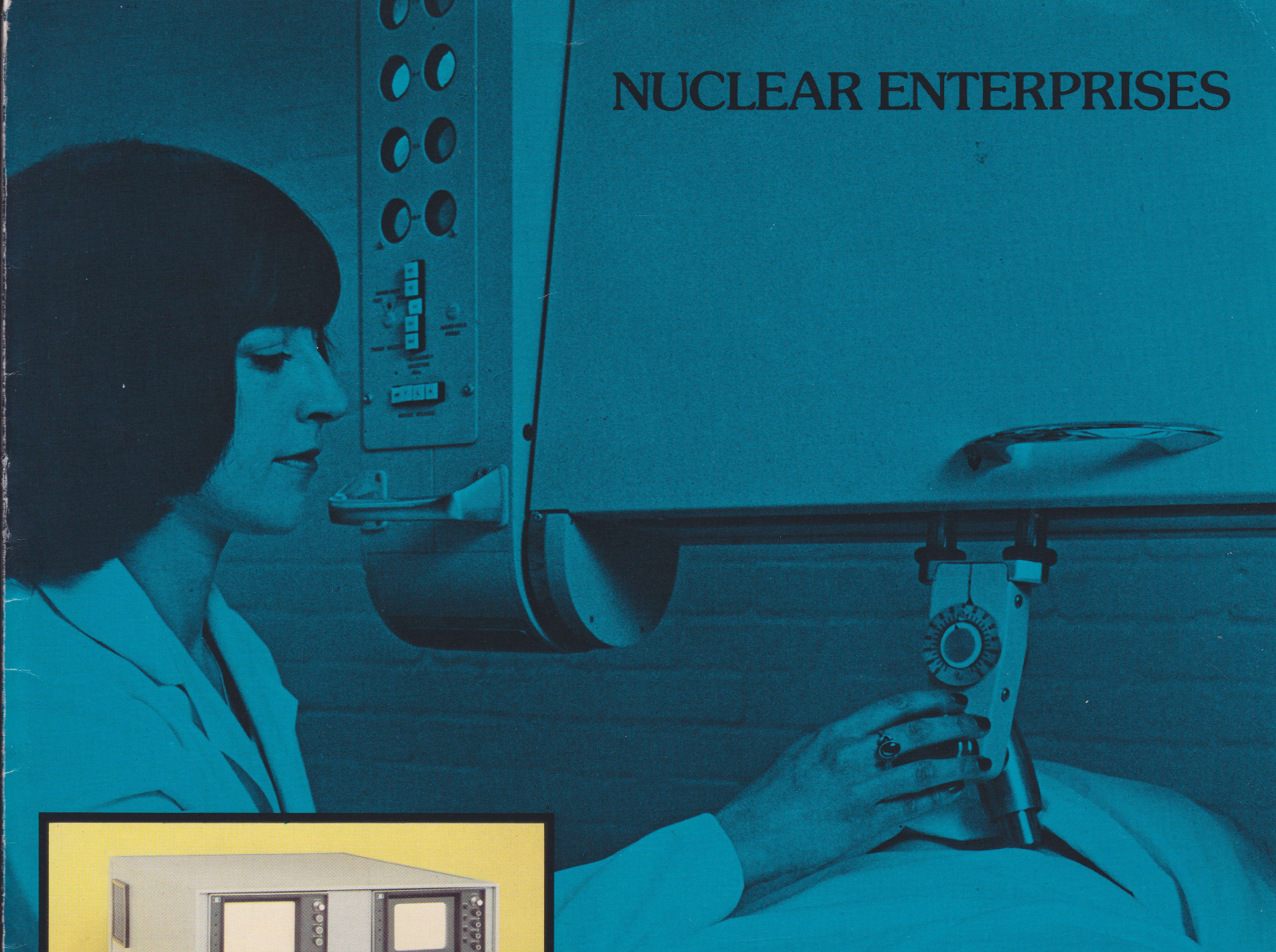
Nuclear Enterprises Ltd. Bulletin No. 88 - New Diasonograph NE4200 with Greyscale Storage Display. July 1976.

NUCLEAR ENTERPRISES



New NE4200 Diasonograph

with Greyscale Storage Display

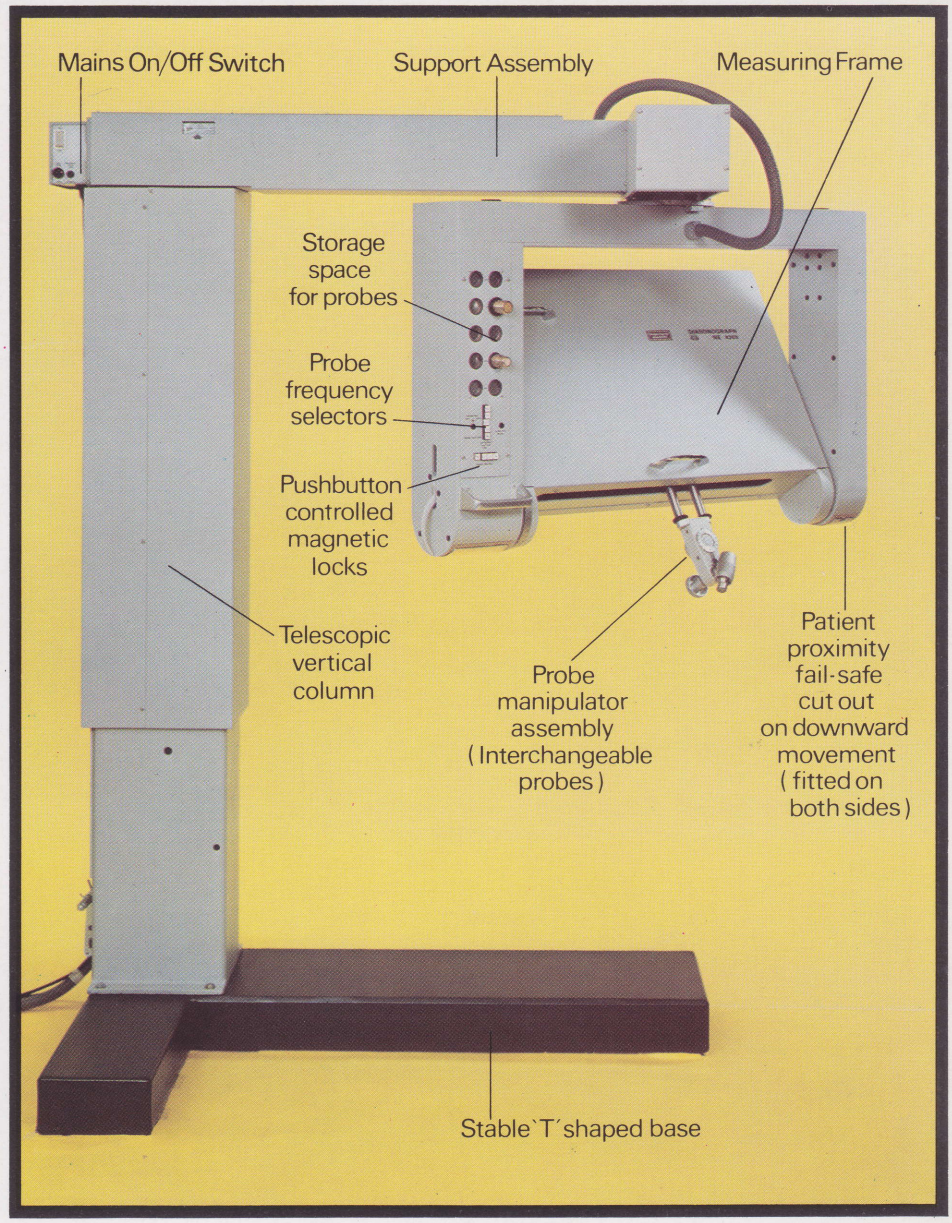
New NE4200 Diasonograph with Greyscale Storage Display

The NE 4200 Diasonograph is the latest addition to the world-leading range of diagnostic ultrasonic scanners developed by Nuclear Enterprises over the past ten years. It has full greyscale capability with colour option for expansion of greyscale, and maintains the high standard of resolution and the same outstanding operational capability as the Award-winning NE 4102 Diasonograph now in use in hospitals in over forty countries. Improved control console facilities are a major feature of the new scanner which operates with the exceptionally stable scanning system and a wide choice of accessories.

The basic Control Console (opposite) now includes improved display oscilloscopes and the following facilities are provided as standard: display of swept gain waveform; calibrated intensity controls for oscilloscopes; and amplifier processing mode selection. Careful ergonomic grouping and colour coding of pushbutton controls ensure simplicity of operation. The recording facilities offered include: photographs (Polaroid, 70mm film), hard copy, strip chart and video.

The Nuclear Enterprises policy of seeking equipment evaluation from groups of workers with widespread experience in the field of medical ultrasound continues and the new NE 4200 is designed to meet individual requirements as fully and as simply as possible. The system can be supplied complete with the NE 4204G Greyscale Storage Display and NE 4103C Cardiac Module (as illustrated on the front cover); with either facility or in its basic form. Nuclear Enterprises Ultrasonic Division will gladly advise on selection.

Scanning Assembly with exceptional Measuring Accuracy

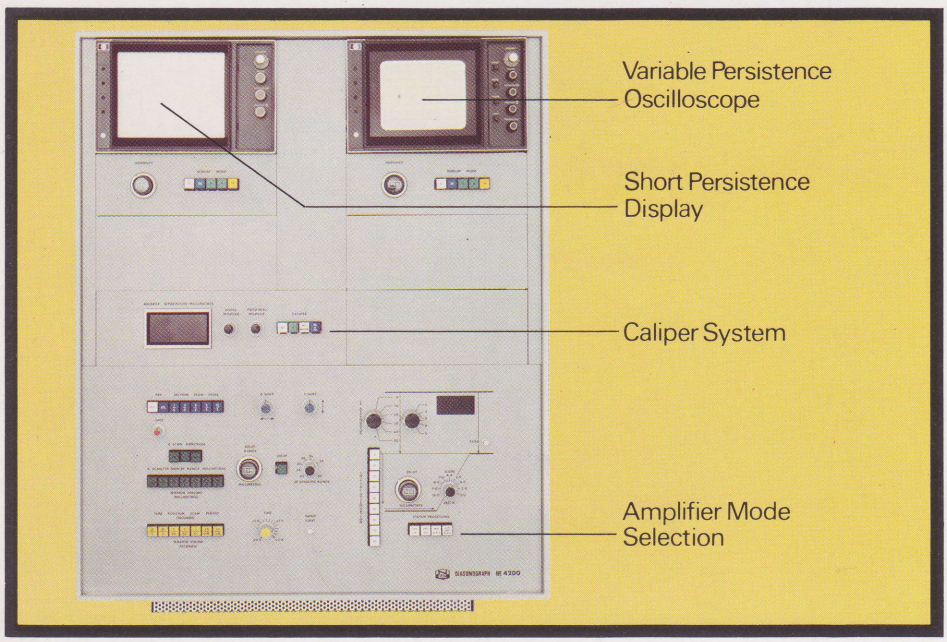


Measuring System

The NE 4200 measuring system affords an operator exceptional freedom in positioning the scanning frame and allows the chosen scanning plane to be easily maintained and reproduced. The one-to-one measuring system coupled to the scanning probe has unequalled measuring accuracy, and all movements are calibrated. Rapid changing from longitudinal to transverse scanning is another important facility offered by the 4200 measuring system.

The NE 4162 2.5MHz focused probe is supplied as standard with the system but a wide choice of interchangeable bayonet fitting probes for different applications is available. These probes may be exchanged without switching off the equipment, and changed from one side of the assembly to the other to suit the examination. The probe may be locked at a specific angle within its travel both for calibration and operational requirements as when it is used with an NE 4167 Biopsy Probe. Optional facilities for locking the X and Y movements are also available.

New Improved Control Console



Basic Control Console

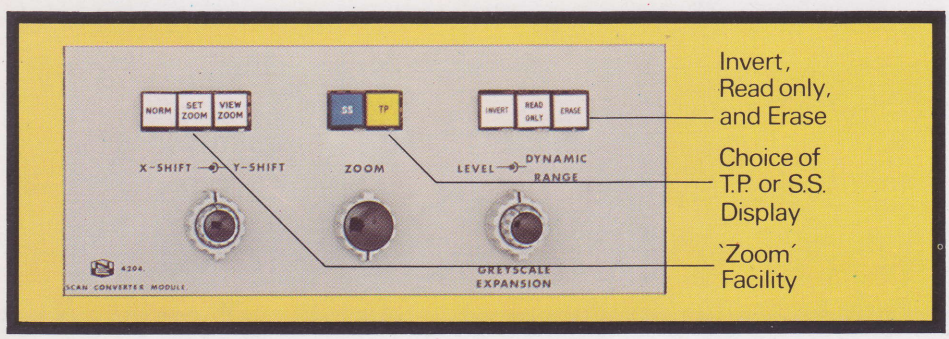
Two recently introduced high performance cathode ray display units are now included in the trolley-mounted electronics console. One is a short persistence HP 1332 and the other is a variable persistence/storage HP 1335, with foot-operated erase switch. A caliper device, a calibrated swept gain display, calibrated oscilloscope intensity controls and amplifier processing mode selection are now standard facilities in the electronic system. The controls on the console are simple. Colour coding is used to indicate function and illumination to indicate choice.

The very comprehensive display facilities with the wide choice of operating modes enables the Diasonograph to cover a wide field of applications. It is possible to display section-scans (B mode), A scan, inverted A scan, time-position scan (M mode) and "open shutter" greyscale scans, including those from the storage greyscale unit, independently on each cathode ray tube display.

Accurate measurements of structural dimensions *in vivo* are possible with the caliper system which has a large scale illuminated digital readout. The caliper pips can be displayed simultaneously on all modes.

The basic NE 4200 Diasonograph has a maximum dynamic range of 40dB, which may be reduced to 22dB by the panel control. The operator may vary the appearance of the scan he wishes to observe by using the combination of the dynamic range control and the amplifier mode selection. "Non-Greyscale" gives a non-greyscale picture ideal for producing outline-type scans of important structures. "Greyscale" operates with two types of signal processing ,"Diff In" and "Diff Out" which provide a scan with a range of grey tones. The incorporation of the NE 4204G Greyscale Storage Display described below allows the complete range of grey tones to be observed.

Integral NE4204G Greyscale Storage Display



Greyscale Storage Display

The NE 4204G Greyscale Storage Display offers the clinician the benefits of conventional storage tubes and, at the same time, a picture on the video monitor with the full range of grey tones. This can enormously simplify the obtaining and interpreting of pictures of more complex structures.

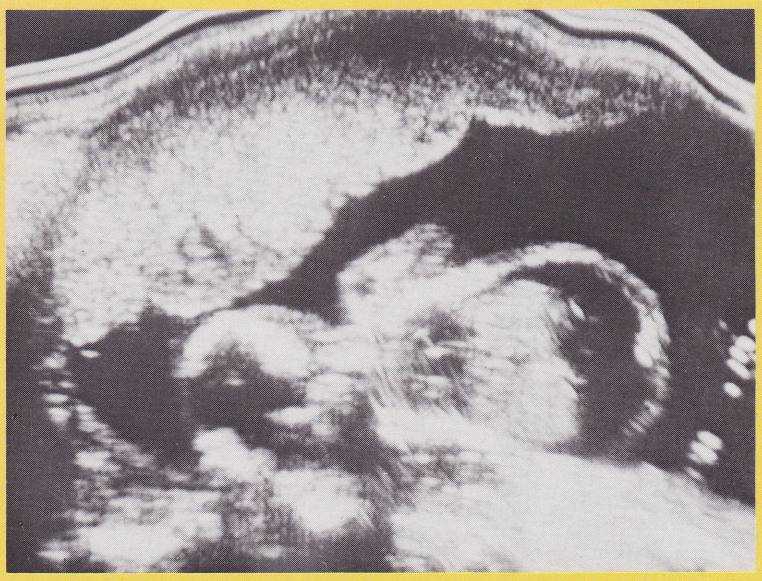
With the NE 4204G it is possible to achieve almost complete freedom from the overwriting effects which can seriously degrade a compound scan result. The user may, without writing out the picture, "compound" a scan in order to collect as much clinically useful information as possible. This significantly improves quality and simplifies diagnosis in some types of examination.

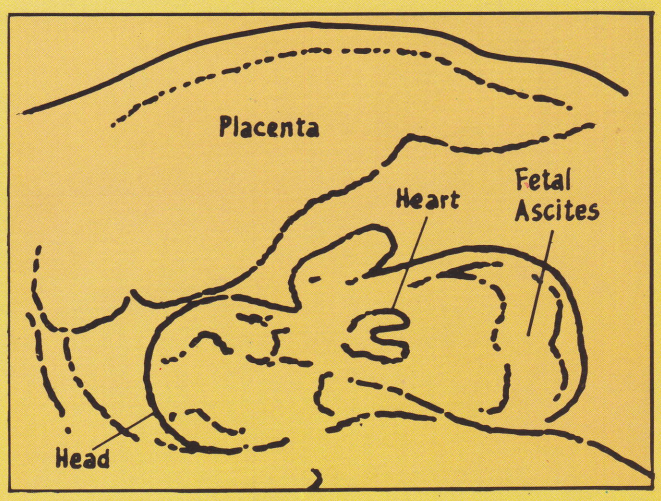
A ten step wedge greyscale may be displayed on the screen for standardisation of the greyscale displays and thereby optimum discrimination of the tissue under examination is obtained.

The NE 4204G time share, read/write facility allows the operator to see the build-up of a picture during the scanning process. Any area of interest may be preselected and with the use of the zoom control switched to cover the total screen width. A magnification 4 × life size is possible without significant loss of resolution. Time-position and Section Scans may be displayed. Permanent records may be obtained by conventional Polaroid photography or by the addition of a hard copy unit. Alternatively, 70mm film recordings can be made.

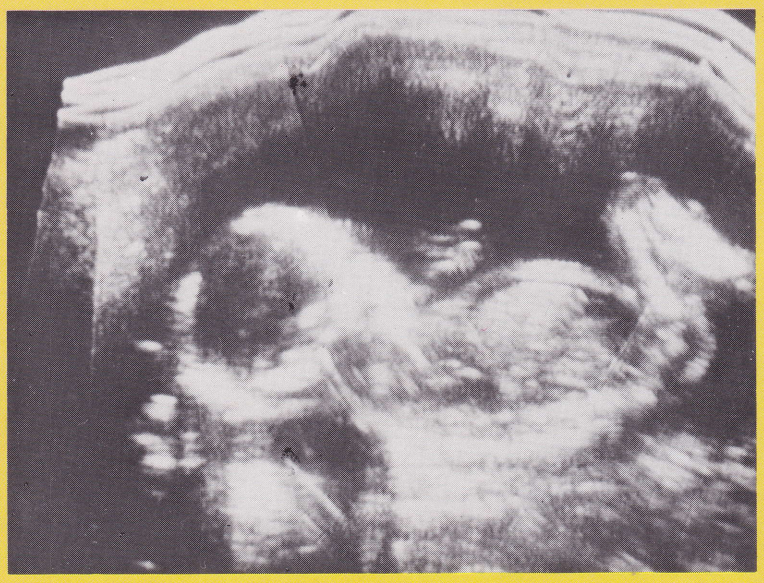
As electronic signals are now available in standard TV form, slave or remote TV monitors may be used for display and video tape recorders employed to store and replay information. Thus monitors may be sited in lecture theatres or in consulting rooms.

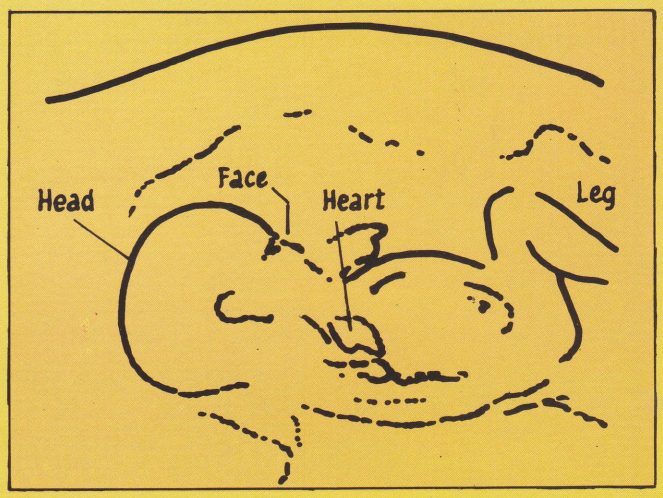
Typical Greyscale Results



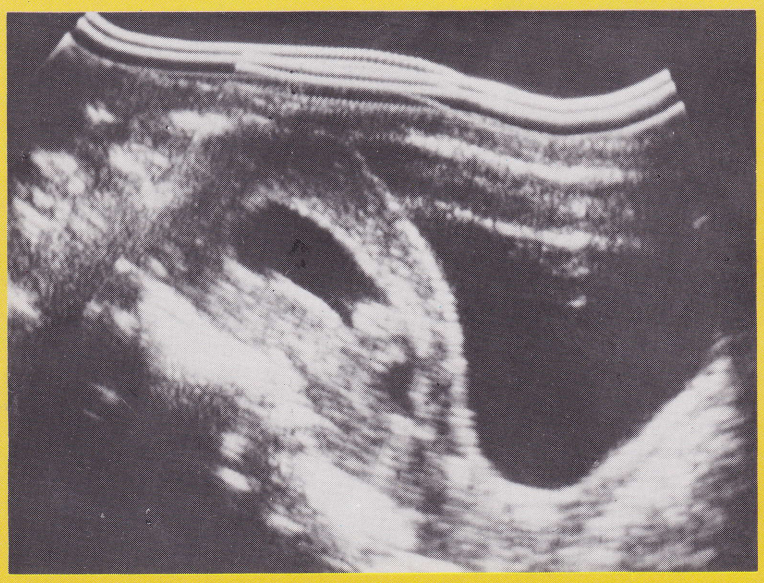


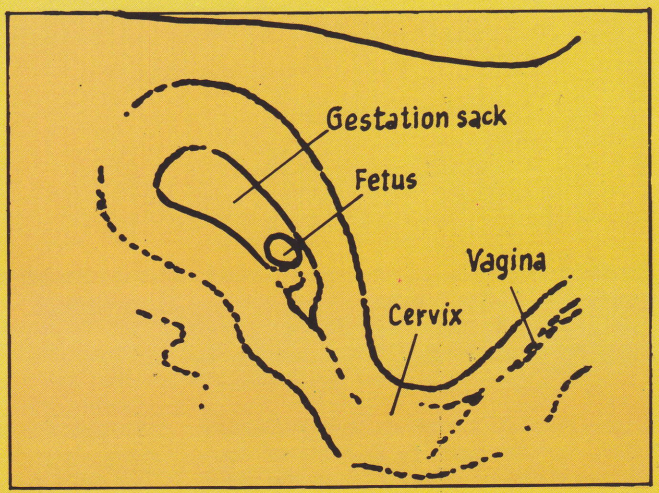
Fetus at 22 weeks with marked ascites due to severe Rhesus iso-immunisation.





Fetus at 24 weeks.



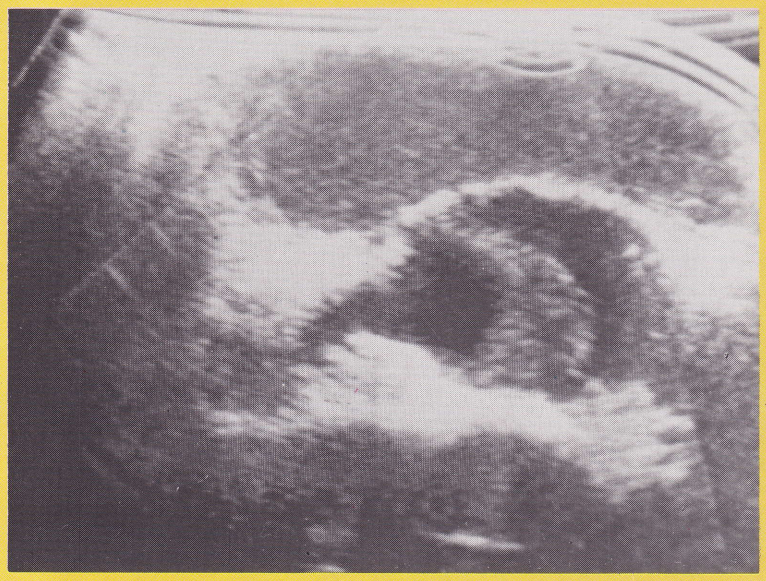


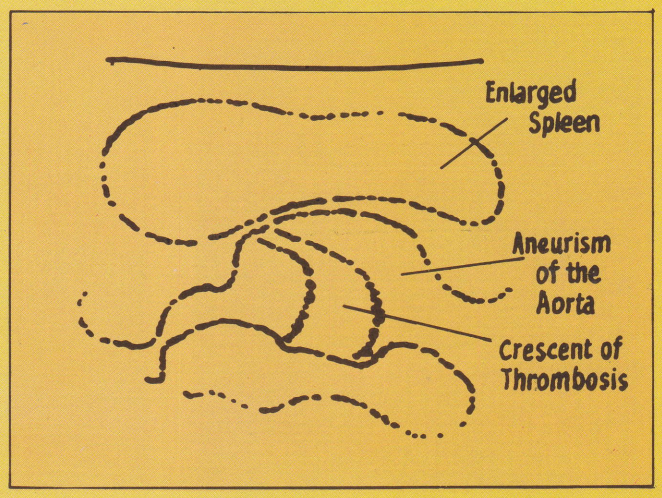
Longitudinal scan of 9 week pregnancy.

Illustrations on this page courtesy Queen Mother's Hospital, Glasgow

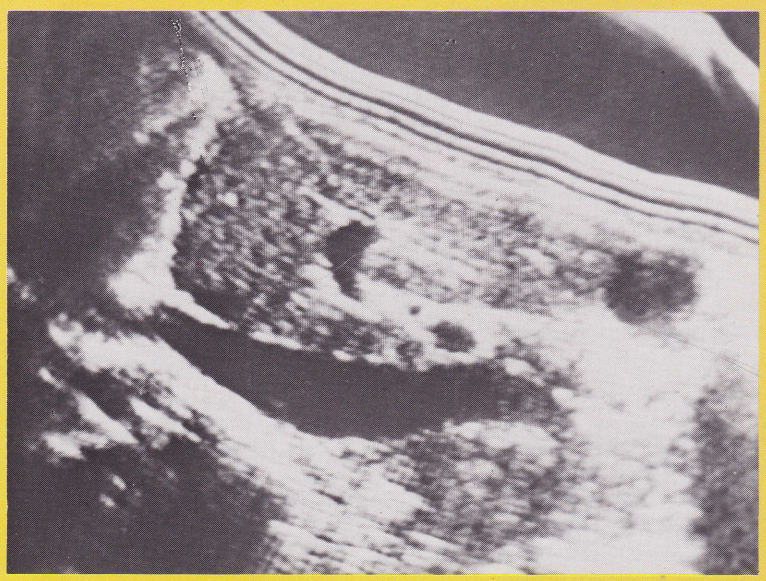
Applications in :

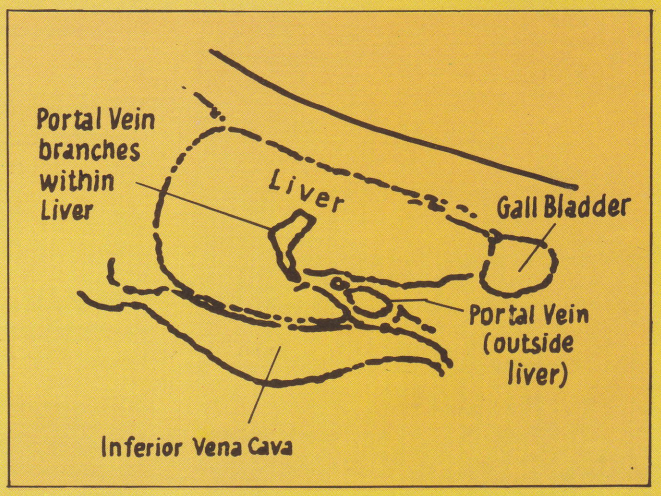
* **Obstetrics**
* **Gynaecology**
* **Abdominal and Pelvic Pathology**
* **Cardiology**
* **Ophthalmology**
* **Oncology**



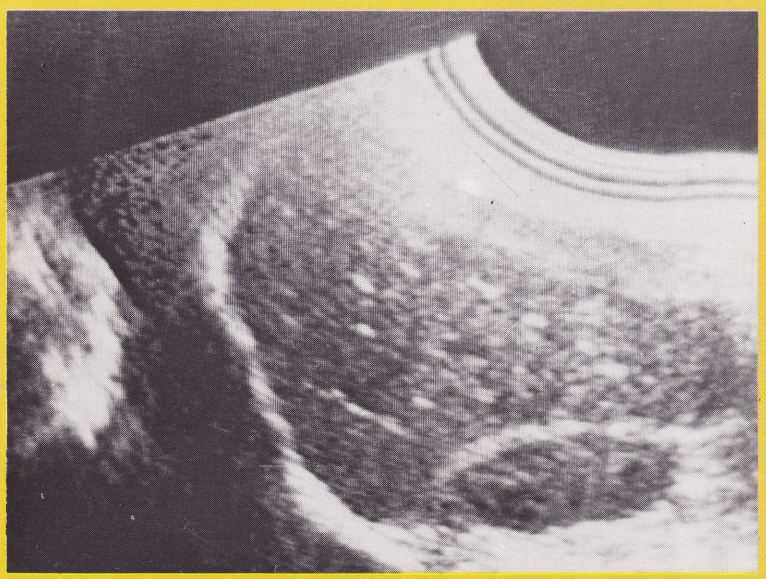


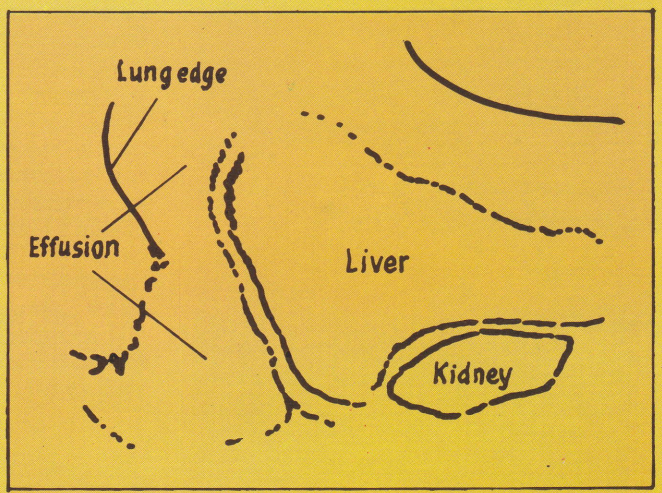
Longitudinal scan—spleen and aorta. Clinically unsuspected aneurism of the aorta behind the enlarged spleen. A crescent of thrombosis present is in the aneurism.





Longitudinal scan—liver. Shows inferior vena cava and portal vein.



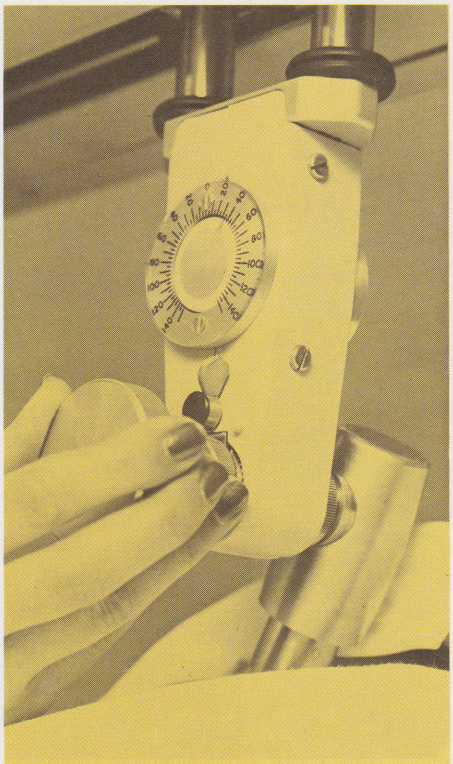


Longitudinal scan—liver and right kidney. A large pleural effusion is present above the diaphragm.

Illustrations on this page courtesy Northwick Park Hospital, Harrow

For Application Notes, (available on request) see p. 8.

Accessories



The ultrasonic power output in all Nuclear Enterprises scanners is reduced to the lowest possible level. The sensitivity of the equipment is controlled by reducing the transmitted power rather than using high power and reducing the gain of the system. For fixed pulse rate applications the pulse repetition frequency is held at 600. During searches a unique velocity controlled pulse repetition system ensures that transmission only takes place while scanning is being carried out at a rate controlled by the speed of scanning.

1
NE 4211 Patient Trolley with paper sheet dispenser and facility for raising or lowering the patient's head to allow adjustment of posture for comfort and/or ease of investigation.

2
NE 4161 1.5MHz focused bayonet fitting probe.

3
NE 4166 5MHz non-focused bayonet fitting probe.

4
NE 4155 2.5MHz hand held probe (cardiology).

5
NE 4167 Biopsy probe—especially useful for amniocentesis 2.5MHz.

6
NE 4141 Water Bath for immersion scanning techniques.

7
NE 4110 Echo Generator for system calibration checks.

8
Shackman Super Seven Camera with Polaroid or 70mm film backs.

9
NE 4210 Remote Photographic facility for use with the NE 4204G Greyscale Storage Display comprises a 6in (152 mm) TV monitor with hinged adaptor to accept a Polaroid or 70mm camera. This TV monitor can be used as a remote viewing monitor and may be placed away from the main ultrasonic scanning area.

10
NE 4204C Colour Conversion Unit is all that is required to upgrade a greyscale storage display to full colour capability. It comprises a special 20in (508mm) colour TV display monitor and additional circuitry. The addition of a colour "window" to the greyscale picture allows much simpler differentiation between adjacent grey shades and this makes examination of fairly homogeneous structures such as livers more easily visualised.

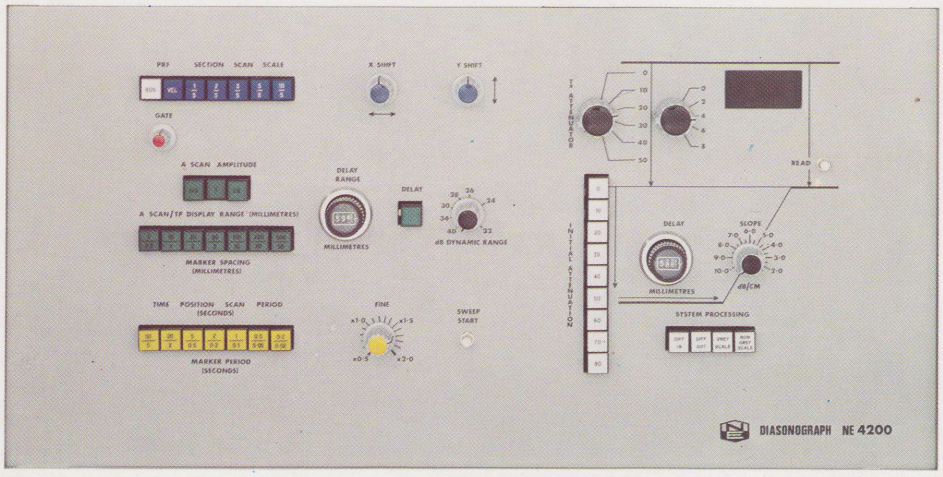
11
NE 4108 Video Cartridge Recorder is basically a PAL Colour $\frac{1}{2}$ in (12.7mm) magnetic tape recorder. It is specially modified to suit the particular requirements of recording scans from the greyscale/colour storage display. Operation is extremely simple, all major operations being pushbutton. The standard $\frac{1}{2}$ in (12.7mm) cartridge tapes supplied have a playing time of 36 minutes and it is possible to produce a simultaneous audio recording using the dubbing facility.

12
NE 4106 Hard Copy Unit produces high quality paper copies of the greyscale information stored on the NE 4204G Greyscale Storage Display video monitors. The copies are large ($8\frac{1}{2}$ in $\times$ 11 in or 216mm $\times$ 279mm), and are produced quickly (approximately 12 seconds).

13
NE 4103C Cardiac Module extends the use of Diasonograph systems in the field of cardiology. This module allows simultaneous presentation of Time-Position Scan, ECG and PCG traces. In addition, the display of ECG triggered cross-section scans of moving structures is possible. The pulse repetition rate is raised to 1,800 pulses per second and synchronisation of the Time-Position Scan from the ECG waveform can be achieved.

14
NE 4105 Fibre Optic Recording Oscilloscope provides a flexible, non-integrating display-record facility. In addition to continuous chart recording (m-mode) conventionally used in cardiological investigation, the NE 4105 can provide permanent hard copy records of section scans, time-position scans and A scans. All NE 4200 series Diasonographs are manufactured with a fibre optic recorder interface socket.

Specifications



Control Panel

NE4200 Specification

Operating controls are distributed in ergonomic groupings to assist operators without specialised knowledge of physics or electronics. Controls are calibrated in cm (or mm) in tissue where appropriate (see Control Panel above).

Sensitivity Control by Transmitter Attenuator:
0 to 50dB in 10dB steps
+0 to 8dB in 2dB steps

Swept Gain (Depth Compensation): Initial reduction: 0 to 80dB in 10dB steps
Initial Delay: 0 to 25cm continuously variable.
Slope: 1.5 to 10.5dB/cm continuously variable.

Picture Scales: (short persistence, variable persistence and scan converter displays)—Section Scan only: 1/5, 2/5, 3/5, 4/5, 5/5 of full-scale.
Scanned Area: Horizontal: 500mm nominal
Vertical: 250mm nominal
Probe rotation: ±135° from vertical in the plane of scan.

A-Scan Display: Occupies half or whole of screen (preset switch selection). Inversion of ½ scale by panel switch to enable accurate photographic comparison to be made between two successive echo patterns. Swept-gain characteristic can be superimposed on the A-scan display (selected by pushbutton control).

A-Scan Range: 5, 10, 20, 50, 100, 200, 500mm switched steps.

A-Scan Delay: 0 to 500mm continuously variable. A delay set/use switch allows the operator to preselect the region to be examined.

Time-Position Display: Intensity modulation of horizontal sweep.

Vertical Sweep (Time-Position only): Single vertical sweeps triggered by push button. Scan period continuously variable between ×0.5 and ×2 of the following ranges:—
50, 20, 5, 2, 1, 0.5, 0.2 in seconds.

Horizontal Range (Time-Position Display): 5, 10, 20, 50, 100, 200, 500mm switched steps (same control as A-Scan range).

Operating Frequencies:
0.5, 1.5, 2.5, 5, 10MHz.

Energy: Maximum available ultrasonic intensity is a function of the probe in use. Using the standard range of probes, the following figures are typical.
Normal mean
Acoustic Power: 1.5MHz 0.70 microwatt
2.5MHz 6.92 microwatt
5MHz 84.0 microwatt
Maximum Available Mean Acoustic Power: 1.5MHz 7.0 milliwatt
2.5MHz 6.92 milliwatt
5MHz 2.7 milliwatt
NOTE: Maximum operating sensitivity of the equipment is controlled by adjustment of the output power.

Transmitter Pulse Repetition Rate:
(a) Fixed 600 pulses/second.
(b) Velocity controlled (section scan only) 0 to 1800 pulses/second.
(c) With cardiac facility (time position scan) 1800 pulses/second.
(d) Triggered section scan with cardiac facility 10 to 100 pulses/cardiac cycle.

Horizontal Delay (Time-Position Display):
0 to 500mm continuously variable (same control as A-Scan range).

Dynamic Range of Echoes: 22 to 40dB continuously variable.

Ultrasonic caliper: Marker separation 0 to 199.9mm in 0.1mm steps over entire A-Scan range.

Power requirements: 100 to 120/200 to 240V ac, 50/60Hz 800VA.

NE4204G Greyscale Storage Display Specification

Description: High resolution unit for storage and display of video and graphic information.
Storage Medium: Princeton Electronic Products PEP 400R Scan Conversion and Image Storage Unit.

Resolution: 1350* to 2100**
*TV lines per diameter at 50% depth of modulation.
**TV lines per diameter limiting resolution.

Output: Television 1V video signal with composite synch.
625 lines, 50 fields or 525 lines, 60 fields.

Display Unit: Electrohome 28cm monochrome monitor.

Front Panel Controls:
(Continuously variable)
"Zoom"—magnifies the image being viewed.
"X"—controls horizontal position of the area being magnified by zoom.
"Y"—controls vertical position of the area being magnified by zoom.

Pushbutton switch selection:
"Norm"—normal full-size viewing of the image.
"Set Zoom"—normal full-size image with "region of interest" superimposed—this is a rectangular box defining the area to be covered by the zoom. Its position and size are controlled by the X, Y and "Zoom" controls.
"View Zoom"—enlarges area within "region of interest" to fill the screen.
"Invert"—changes the displayed picture from positive to negative. Used for making positive pictures on negative film.
"Read only"—stops the writing of images on the storage unit when the probe is moved.
"Erase"—clears screen for the next scan.
"SS"—selects section scan picture storage.
"TP"—selects time-position scan picture storage.

Preset Controls on the Storage Unit:
"Read"—Preset control for optimisation of image read from the silicon storage target.
"Focus"—Focus of image read from the storage target.

Rear Panel Outputs/Inputs:
"Video Out"—1V composite synch video signal for TV monitor or video tape recorder. BNC 750hm socket.
"Video In"—1V composite synch video input for colour processing.
"Colour TV"—Multiway connector for colour TV monitor for NE 4204C.

Dynamic Range of system with NE 4200: 36dB.

Greyscales Displayed: 10.

Power Requirements:
100 to 120/220 to 240V ac, 50/60Hz, 200VA.

The following Application Notes are available on request:

1. Ophthalmic Ultrasonography—G. R. Sutherland FRCP(E), FFR. Southern General Hospital, Glasgow.
2 & 3. Obstetric Ultrasonography—Hugh P. Robinson MRCOG. Queen Mother's Hospital, Glasgow.
4. Echocardiography—D. A. R. Robertson FRCR. Southern General Hospital, Glasgow.
5. Ultrasonic Scanning of the Abdomen—Patricia Morley MB, DMRD, and Ellis Barnett FRCP, FRCR. Western Infirmary, Glasgow.
6 & 7. Short Case Notes on Abdominal Scanning—Hylton B. Meire MB, BS, DObst RCOG, FRCR. Northwick Park Hospital, Harrow.

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Bulletin No 88 July 1976

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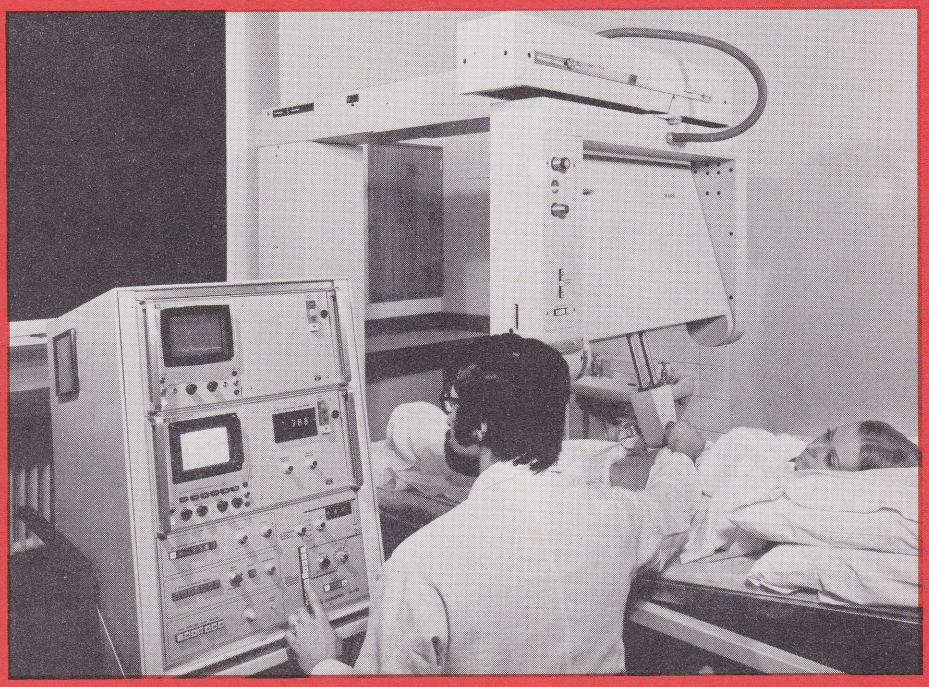
Nuclear Enterprises Ltd. Bulletin No. 434 - Ultrasonic Greyscale Facilities. Edinburgh. April 1975.

DIAGNOSIS BY ULTRASOUND

GREYSCALE FACILITIES



with
DIASONOGRAPHS
NE 4102 & 4102A

Nuclear Enterprises Limited

ULTRASONIC

Interest is becoming very widespread in the application of greyscale techniques. A summary of equipment available for Storage Display and details of the capability of Nuclear Enterprises standard scanners in this field are given below. For full information please contact the Ultrasonic Sales Department, Tel: 031-443 4060.

NEW GREYSCALE STORAGE DISPLAY

The new NE 4104 Greyscale Storage Display offers the clinician the benefits of conventional storage tubes and, at the same time, a picture on the screen with the full range of grey shading. This can enormously simplify the obtaining and the interpretating of pictures of more complex structures. (The four pictures on this page are printed by kind permission of the Queen Mother's Hospital, Glasgow).

Easier Operation and Improved Picture Quality

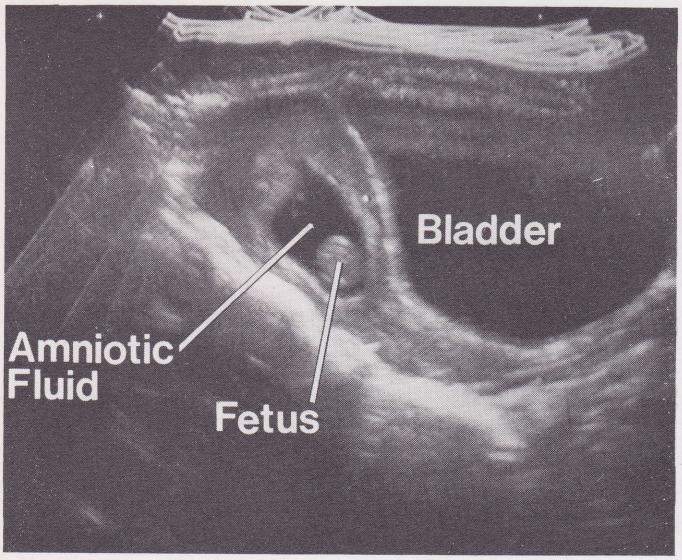
With the NE 4104 it is possible to achieve almost complete freedom from the overwriting effects which ruin many potentially good pictures obtained with conventional (non-storage) greyscale techniques. The user is encouraged to 'compound' every scan in order to collect as much clinically useful information as possible. This automatically improves quality and simplifies diagnosis.

Ten discernible shades of grey may be displayed on the screen for optimum discrimination of tissues under examination. In addition, special dynamic range compression and contrast enhancement circuitry allow the echo emplitudes of interest to be spread across the grey range of display.

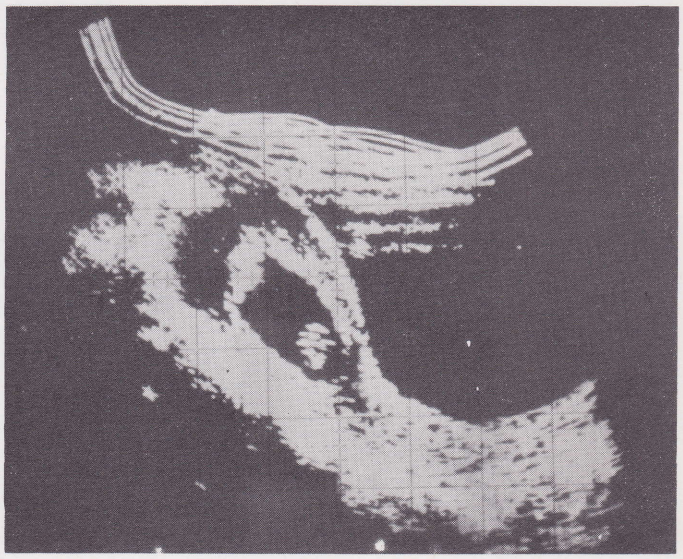
Display and Record Facilities Increased

The NE 4104 time share, store/view facility allows the operator to see the build-up of a picture during the scanning process. Any area of interest can be selected and magnified to cover the screen and a maximum overall magnification 4x life size is possible without significant loss of resolution. TP and Section-scans may be displayed, and, with Diasonograph NE 4102B only, up to 4 section scans may be stored and viewed. Permanent records may be obtained by conventional Polaroid photography or by the addition of a hard copy unit.

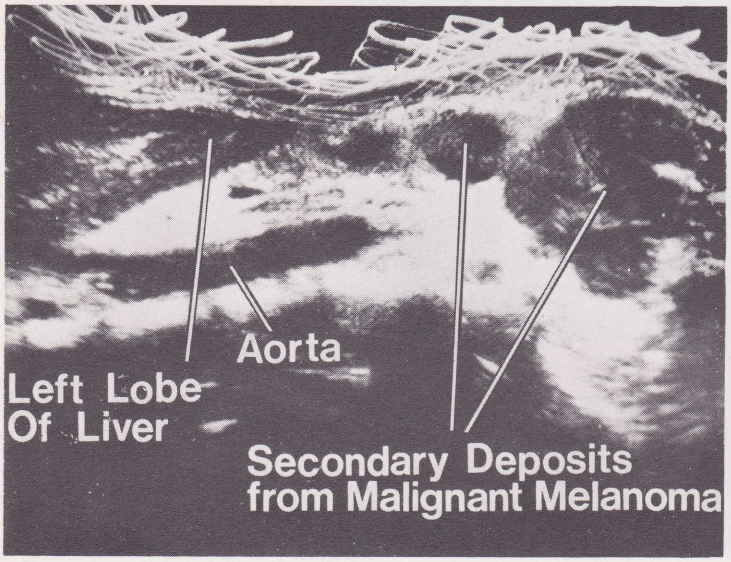
As electronic signals are now available in standard TV form, slave or remote TV monitors may be used for display and video tape recorders employed to store and replay information. Thus monitors may be sited in lecture theatres or in consulting rooms.



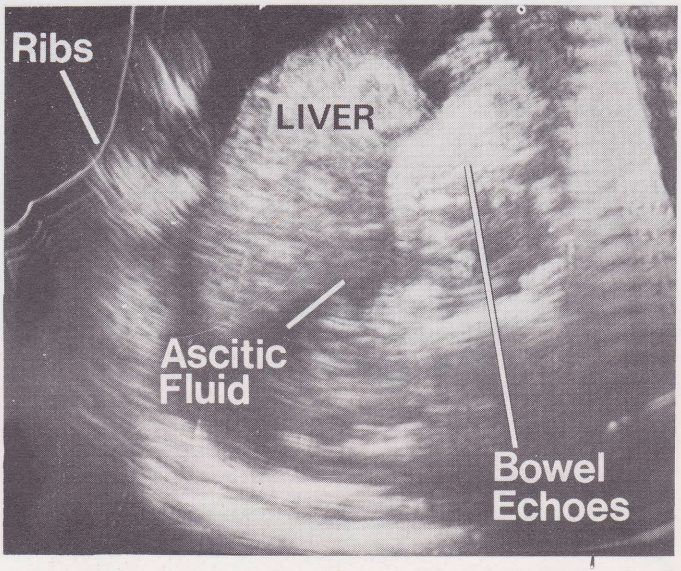
LONGITUDINAL SECTION OF EARLY PREGNANCY



LONGITUDINAL SECTION OF EARLY PREGNANCY (NO GREYSCALING)



LONGITUDINAL SCAN OF ABDOMEN



HIGH TRANSVERSE SCAN

CS - GREYSCALE FACILITIES

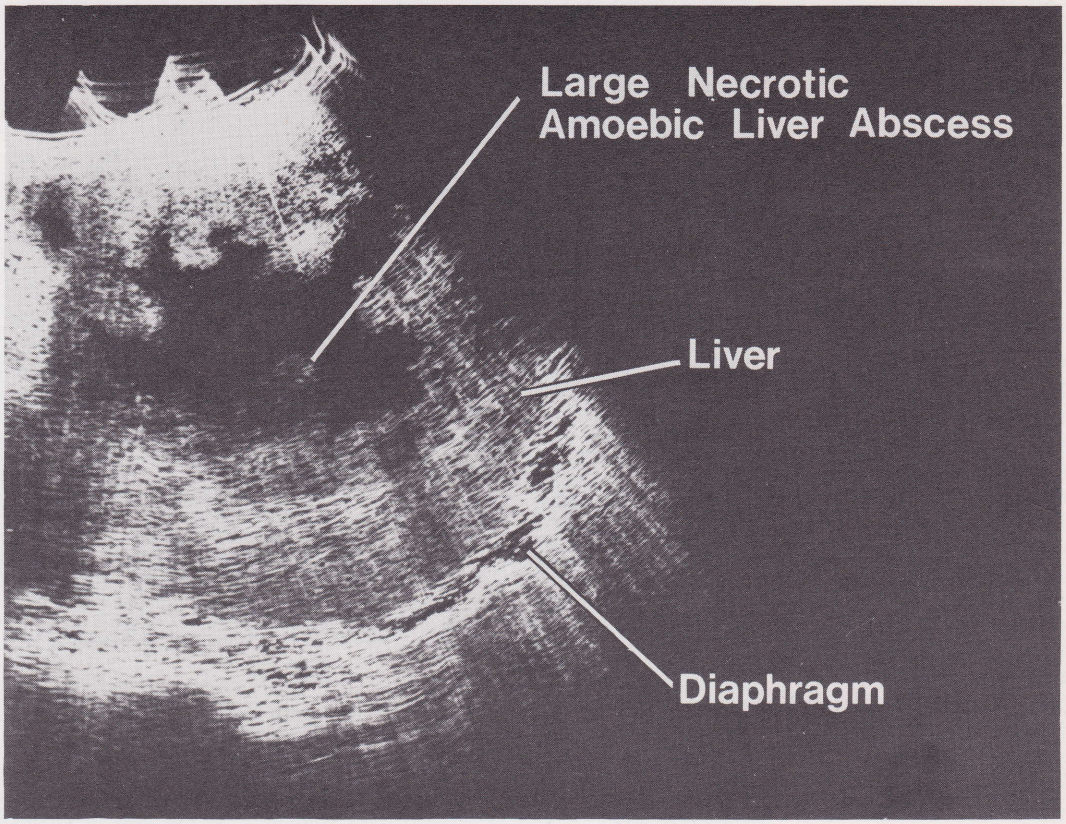
GREYSCALE CAPABILITY OF STANDARD EQUIPMENTS

NE 4102 and NE 4102A Diasonographs

All Nuclear Enterprises scanners can be used to produce greyscale pictures by time exposure photography of the non-storage display. The technique can also be used on the variable persistence display when this is switched to the non-store (conventional) mode. However, signal processing on the NE 4102 and NE 4102A includes differentiation for resolution enhancement which reduces greyscale contrast. Nuclear Enterprises can now supply a Modification Kit which permits non differentiated signals to be used and ensures maximum greyscale contrast for applications where greyscale is the most important picture characteristic.

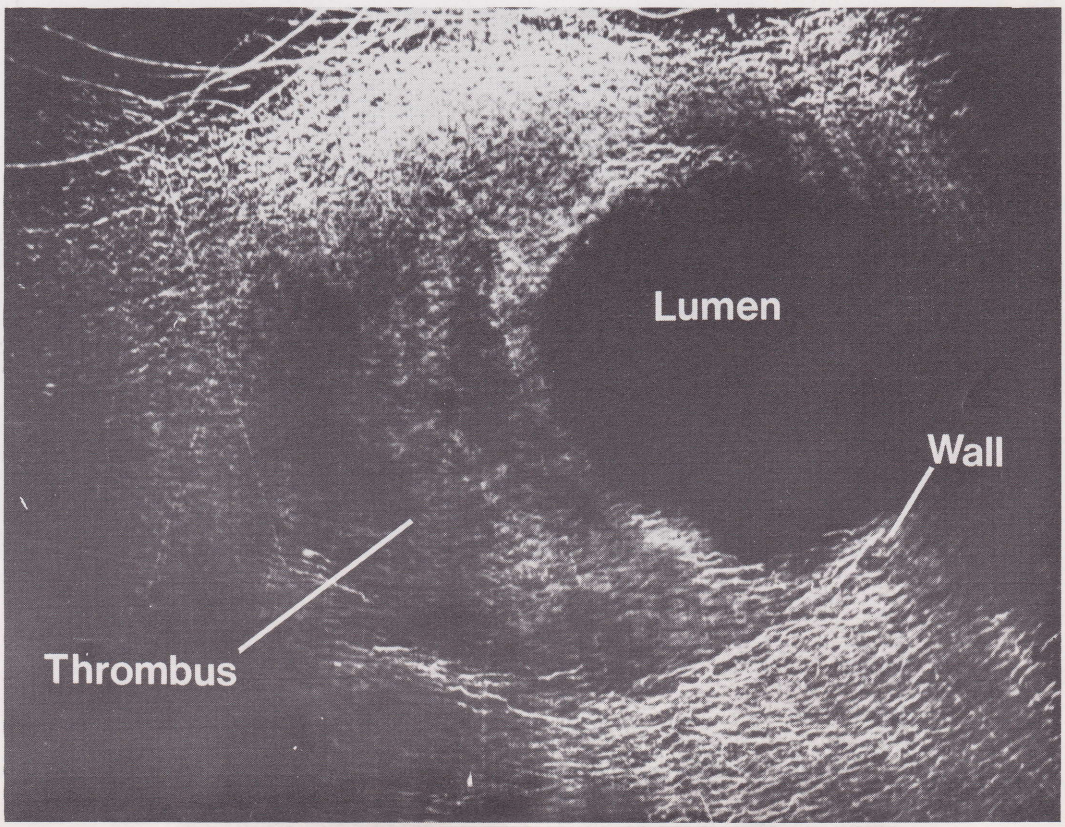
The Kit is easily incorporated in all equipments and, when the standard NE 4102 has been modified, two signal processing modes are available. The first mode "Differentiation," is suitable for high resolution greyscale visualisation. The second Mode, "Non - Differentiation", emphasises greyscale, but compromises boundary resolution or "sharpness" of the displayed result.

N.B. All the greyscale conditions specified above for the modified standard NE 4102 (or NE 4102A) are automatically included in the NE 4102B, so no modification is required for NE 4102B. The four pictures on pages 3 and 4 were taken with the



LARGE NECROTIC AMOEBIC LIVER ABSCESS

TRANSVERSE SCAN. LARGE ABDOMINAL AORTIC ANEURISM



NE 4102B and are printed by kind permission of Dr. D. A. R. Robertson, Dept. of Radiology, Southern General Hospital, Glasgow and Dr. R. Railton, Dept. of Clinical Physics and Bio-Engineering, Glasgow.

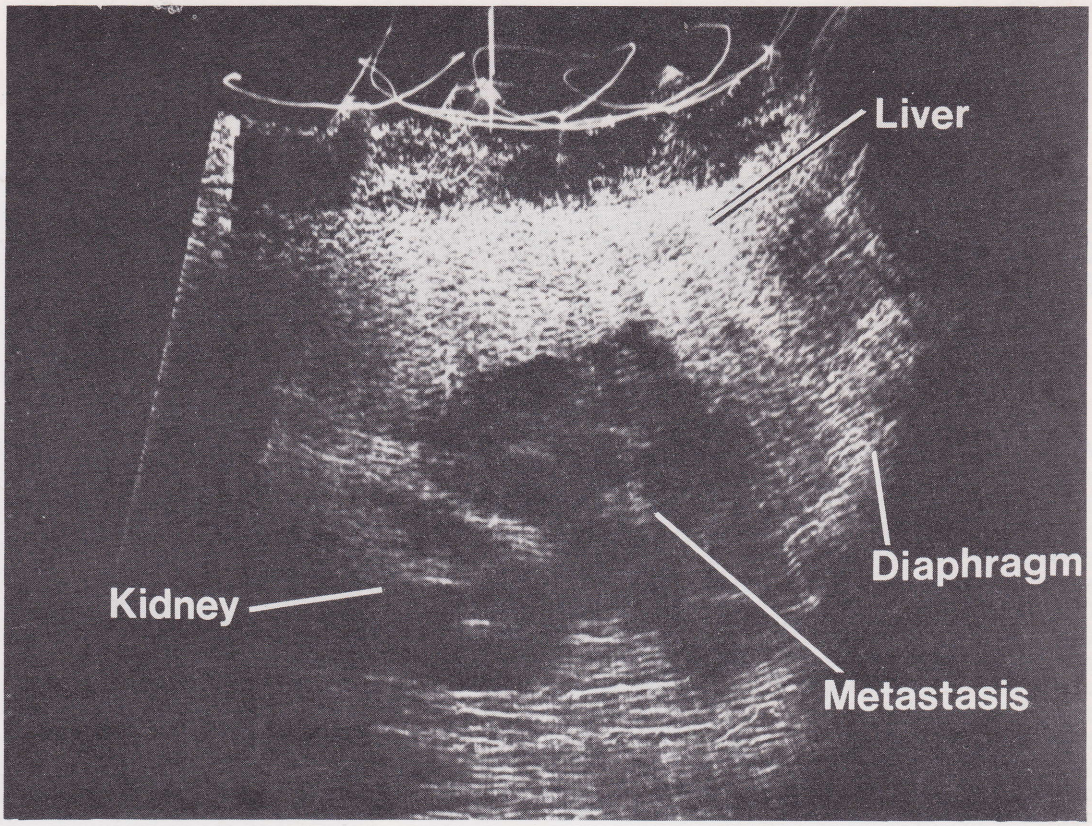
NE 4105 Fibre Optic Recording Oscilloscope

The NE 4105 can be fitted to the NE 4121 Diasonoscope and to all the NE 4102 Series Diasonographs to provide permanent records at comparatively low cost. The differentiated and non-differentiated signal processing option is a standard facility in this unit. The NE 4105 reproduces up to five grey tones unequivocally and, in the section scan mode, considerable over scanning without deterioration of picture quality is possible.

Records of single frame time position scans, A-scans, and continuous time position scans for long term observation of moving structures, can also be obtained with the NE 4105.

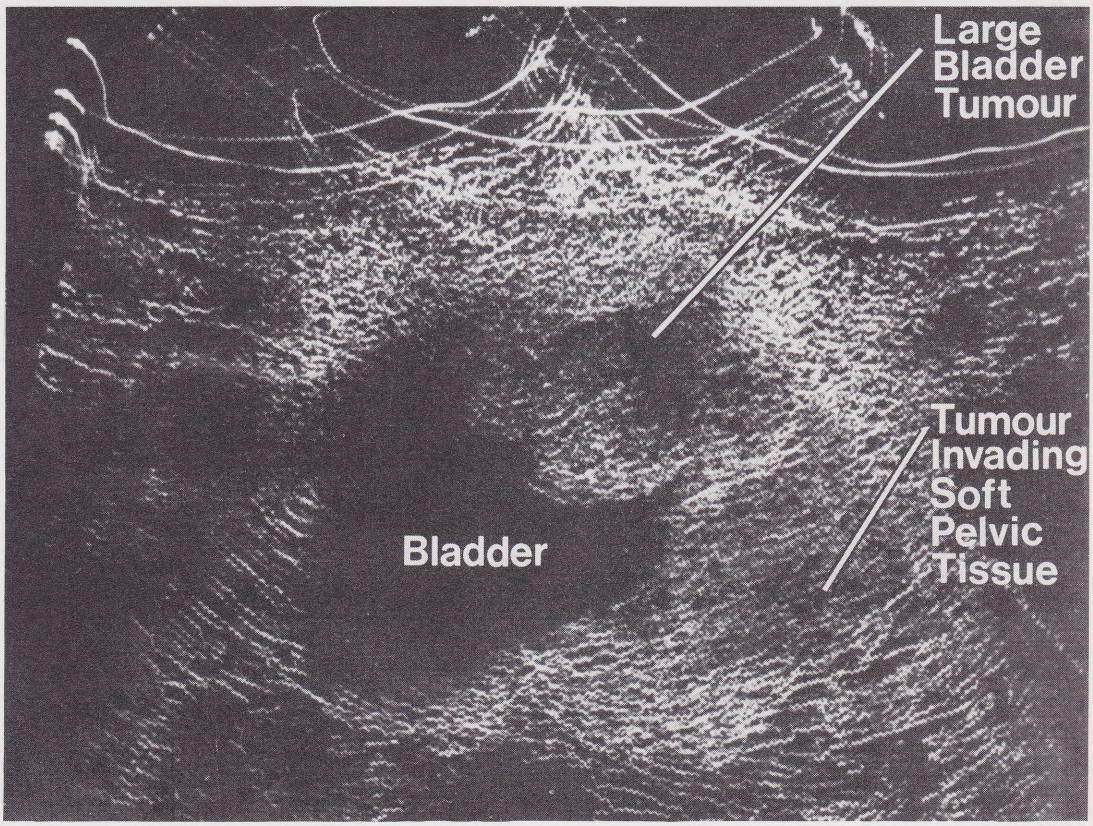
NE 4107 Greyscale Calibration Unit

The NE 4107 provides a calibration signal into the input of the receiver amplifier, and a drive waveform to the swept gain unit. This enables a thorough check to be made of the performance of all critical elements affecting the greyscale performance — not merely as individual elements but as a complete system.



LONG SCAN. LARGE METASTASIS IN RIGHT LOBE LIVER

TRANSVERSE SCAN, BLADDER. TUMOUR: INTRA-LUMINAL PLUS EXTRA-VESCICAL SPREAD



Nuclear Enterprises Limited

Bulletin No. 434
April 1975

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* Registered Office Registration No. 31256 Scotland

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E

Nuclear Enterprises Ltd. Bulletin No. 90 - NE4104G Greyscale Storage Display Accessory for NE4102 Diasonograph Systems, July 1976.

DIAGNOSIS BY ULTRASOUND

NE4104G GREYSCALE STORAGE DISPLAY ACCESSORY FOR NE4102 DIASONOGRAPH SYSTEMS

HIGH DEFINITION LARGE SCREEN NON-INTEGRATING GREYSCALE STORAGE DISPLAY CAPABILITY

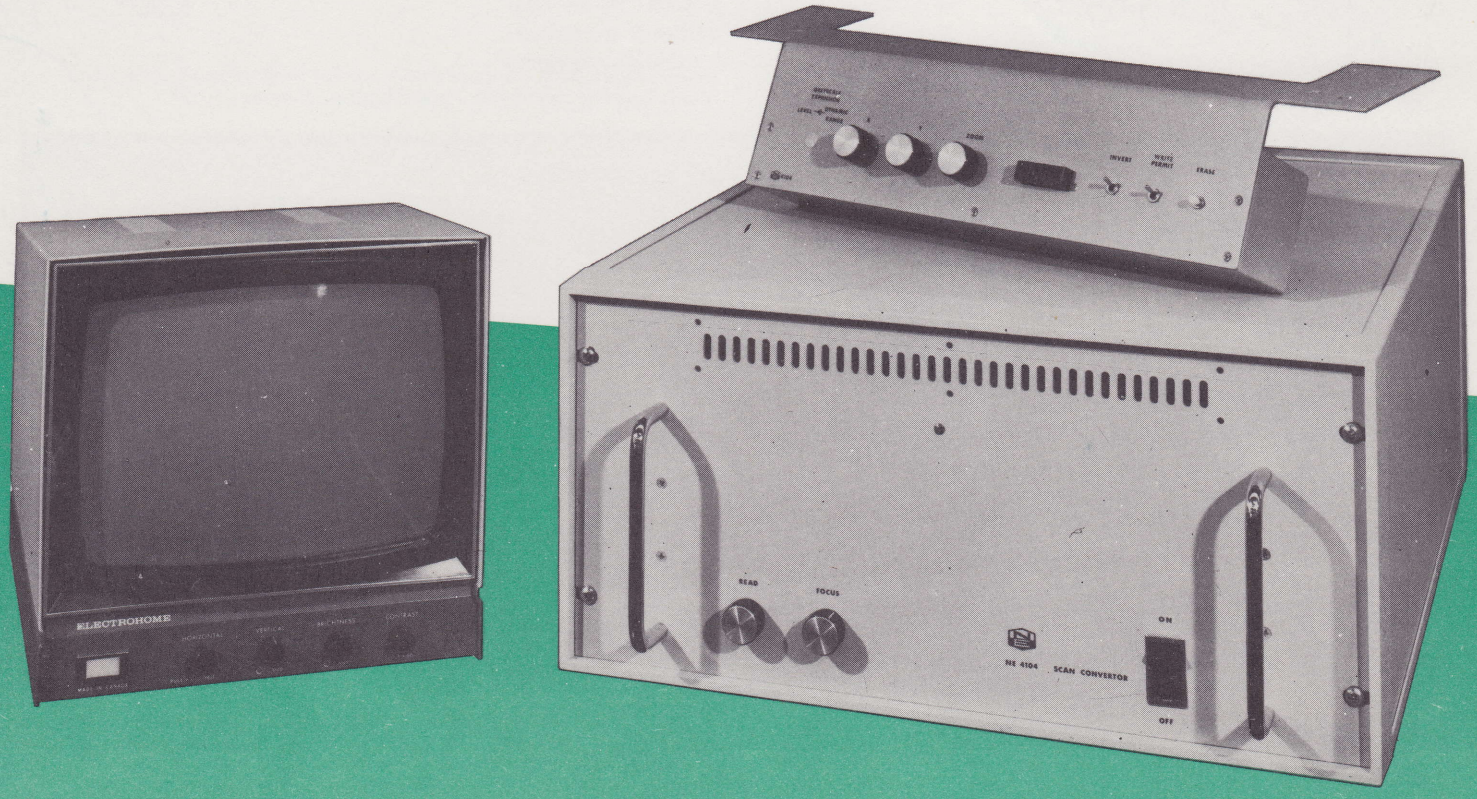
The NE 4104G has been designed for use with the NE 4102, NE 4102A and NE 4102B Diasonographs. The unit is supplied in its own cabinet which fits on the trolley below the main electronics console together with the necessary interfacing circuitry and display unit.

The NE 4104G Greyscale Storage Display offers Diasonograph users the benefits of conventional storage tubes and, at the same time, a picture on the TV monitor screen with the full range of grey shading. This can enormously simplify the obtaining and the interpreting of pictures of more complex structures.

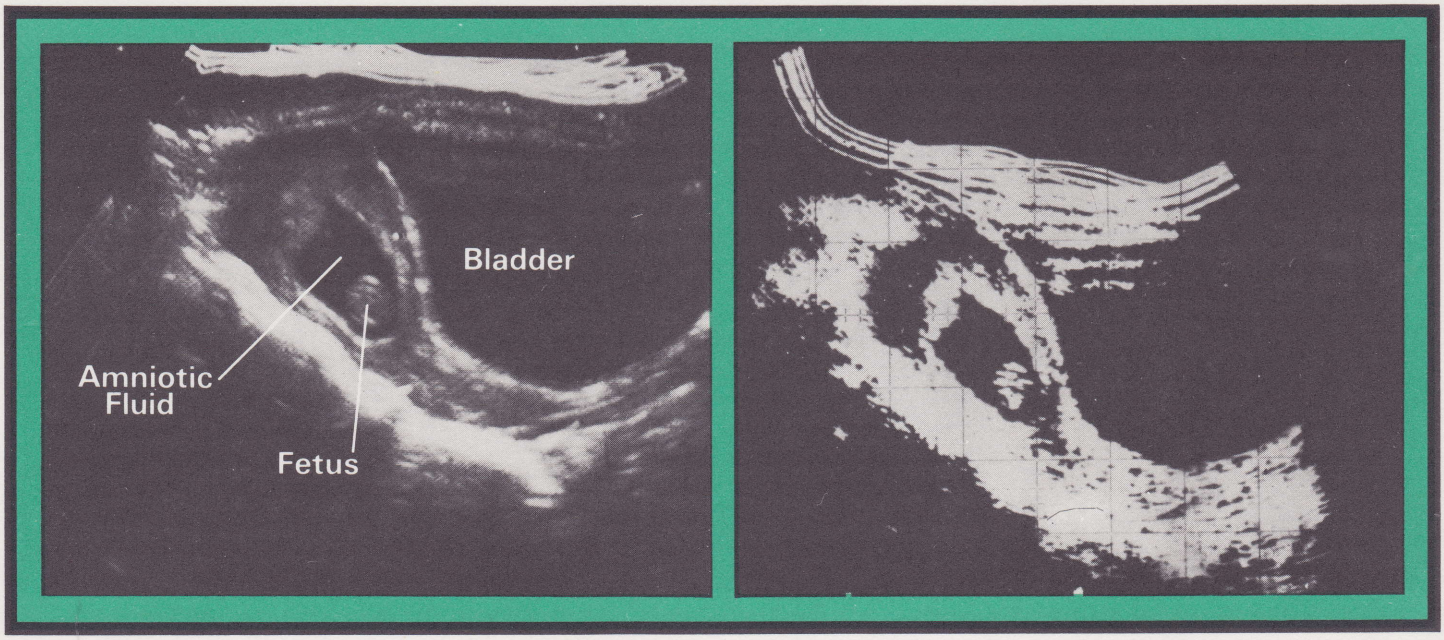
Easier Operation and Improved Picture Quality

With the NE 4104G it is possible to achieve almost complete freedom from the overwriting effects which spoil many potentially good pictures obtained with conventional (non-storage) greyscale techniques. The user may, without writing out the picture, 'compound' a scan in order to collect as much clinically useful information as possible. This significantly improves quality and simplifies diagnosis in some types of examination.

Ten discernible shades of grey may be displayed on the screen for optimum discrimination of tissues under examination. In addition, special dynamic range compression and contrast enhancement circuitry allow the echo amplitudes of interest to be spread across the grey range of display.



Typical Results



Top left: Longitudinal section of early pregnancy

Top right: Longitudinal section of early pregnancy — no greyscaling

Courtesy Queen Mother's Hospital, Glasgow

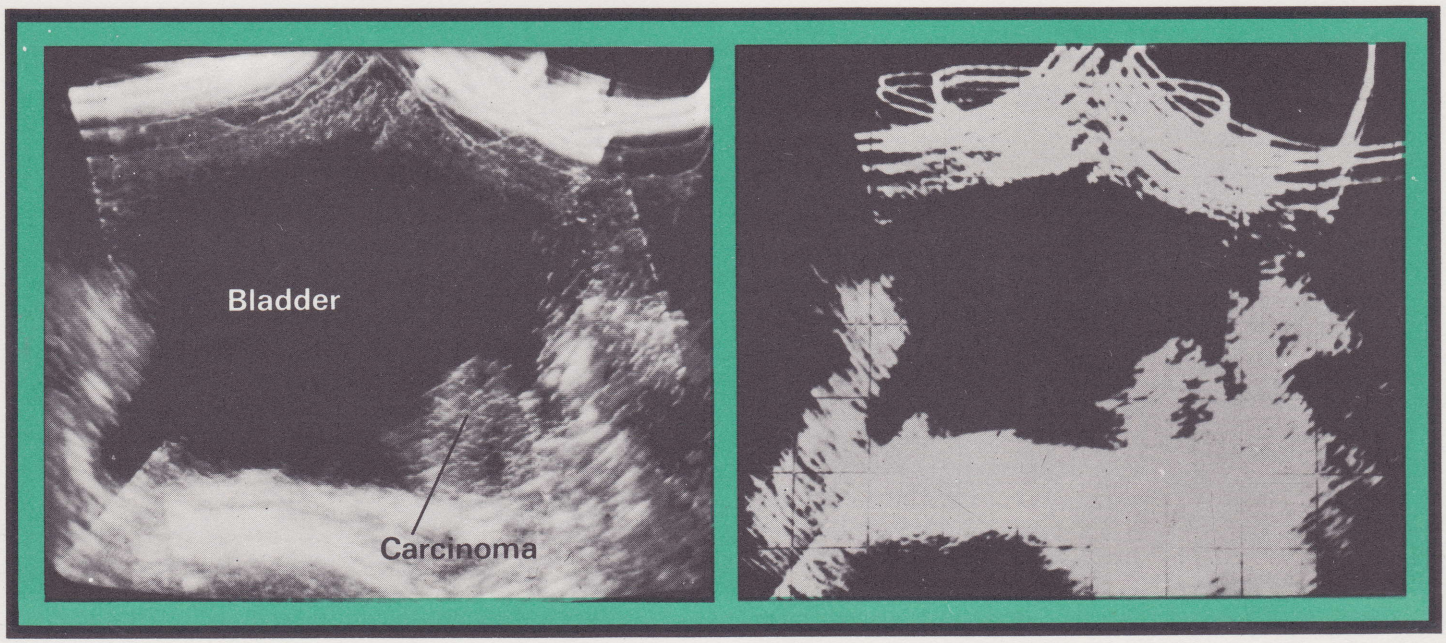
Display and Record Facilities Increased

The NE 4104G time share, read/write facility allows the operator to see the build-up of a picture during the scanning process. Any area of interest can be selected and magnified to cover the total screen width. A maximum overall magnification 4× life size is possible without significant loss of resolution. TP and section scans may be displayed and, with Diasonograph NE 4102B only, up to 4 section scans may be stored and viewed. Permanent records may be obtained by conventional Polaroid photography or by the addition of a hard copy unit.

As electronic signals are now available in standard TV form, slave or remote TV monitors may be used for display, and video tape recorders employed to store and relay information. Thus monitors may be sited in lecture theatres or in consulting rooms.

Bottom left: Transverse section of a female bladder with a left posterior carcinoma

Bottom right: Same study as shown left but without greyscaling. Both results courtesy Northwick Park Hospital, Harrow



Technical Description

The NE 4104G Greyscale Storage Display has been designed to give greyscale pictures from Diasonographs. The heart of the system is the scan converter tube which stores the image. The stored picture cannot, however, be viewed directly but is electronically processed so that it is in the format of a standard television video waveform and is viewed on standard TV monitors.

The image is stored on a silicon chip or target within the scan converter. The principle of operation is for the target to be charged to a high voltage. The signals, which drive the display tubes on a Diasonograph, are used to discharge the voltage on this target, and thus the pattern of charge remaining on the target is the same as the pattern on a storage oscilloscope tube. The target is scanned by a beam of electrons, the scanning being in the same pattern as on a TV screen. The number of electrons in the scanning beam is determined by the charge on the target where the beam is aimed, and since the movement of electrons is an electric current then these current changes can be used to present the information on the target as a TV type signal. The *scanning* format of the Diasonograph has been *converted* to a TV picture.

The greyscale capability of the scan converter is due to its non-integrating method of writing the information on the target. On the standard oscillographic display the intensity is an additive process so that several small echoes received from one point scanned from various positions are displayed with the same intensity as one large echo from one point.

The charged silicon target of the scan converter is a non-integrating or non additive process. Several repeated echoes from the same point do not cause over-writing. The 4104G does not only consist of the scan conversion circuits. It is possible to enlarge the image on the target by a zoom facility, the area of the zoom being defined by the 'region of interest' or 'new zoom' circuits. The TV signal can also be inverted to give a negative picture, and this facility is particularly useful for recording results on negative film. The scan conversion process is also carried out in such a way that the target can be scanned and the image viewed at the same time as information is being written on to the target.

Interfacing is supplied to enable the signals to be derived from the Diasonograph in a suitable manner for scan conversion and the TV monitor picture to be displayed (and photographed) on one of the Diasonograph's display tubes.

Accessories

1 NE 4210 Remote Photographic facility for use with the NE 4104G Greyscale Storage Display comprises a 6 inch TV monitor with hinged adaptor to accept a Polaroid or 70mm camera. This TV monitor can be used as a remote viewing monitor and may be placed away from the main ultrasonic scanning area.

2 NE 4108 Video Cartridge Recorder is basically a PAL Colour ½ inch recorder. It is specially modified to suit the particular requirements of recording scans from the greyscale display. Operation is extremely simple, all major operations being pushbutton. The standard ½ inch cartridge tapes supplied have a playing time of 36 minutes and it is possible to produce a simultaneous audio recording using the dubbing facility.

3 NE 4106 Hard Copy Unit produces high quality paper copies of the greyscale information stored on the NE 4104G Greyscale Storage Display video monitors. The copies are large, 216×279mm (8½×11in), and they are produced quickly (approximately 12 seconds).

NE4104G Specification

Description: High resolution unit for storage and display of video and graphic information.

Storage Medium: Princeton Electronic Products PEP 500R Scan Conversion and Image Storage Unit.

Resolution: 1350* to 2100**
*TV lines per diameter at 50% depth of modulation
**TV lines per diameter limiting resolution

Output: Television 1V video signal with composite synch. 625 lines, 50 fields or 525 lines, 60 fields.

Display Unit: Electrohome 28cm monochrome monitor.

Front Panel Controls: Continuously variable.
'Zoom'—Control to magnify the image being viewed.
'X'—Controls the horizontal position of the area being magnified by the zoom.
'Y'—Controls the vertical position of the area being magnified by the zoom.

Pushbutton switch selection:
'Norm'— Normal full-size viewing of the image.
'Set Zoom'—Normal full-size image with 'region of interest' superimposed—this is a rectangular box defining the area to be covered by the zoom. Its position and size are controlled by the X, Y and 'zoom' controls.
'View Zoom'—Enlarges area within 'region of interest' to fill the screen.

Toggle Switches:
'Invert'—Changes the displayed picture from positive to negative. Used for making positive pictures on negative film.
'Write permit'—Enables the circuitry to write images on the storage unit when the probe is moved.
'Erase'—Clears the screen for the next scan.

Preset Controls on the Storage Unit
'Read'—Preset control for optimisation of the image read from the silicon storage target.

continued overleaf

NE4104G Specification (contd)

'Forms'—Focus of image read from the storage target.

Rear Panel Outputs/Inputs
'Footswitch erase'—For remote erase facility. Multiway.
'Video Tape'—1V composite synch video signal for video tape recorder. BNC 75ohm socket.
'TV Monitor'—1V composite synch video signal for TV monitor. BNC 75ohm socket.
'Graph'—Multiway connector to Diasonograph.
'Front Panel'—Multiway connector to front control panel.
'Colour TV'—Multiway connector for colour TV monitor for NE 4104C.
'Mains Input'—Mains power supply.

Dynamic Range of system with NE 4102 36dB

Greyscales Displayed 10.

Power Requirements
220 to 240V, 50Hz
or 110 to 120V, 60Hz } 200VA

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Printed by J. Thomson Colour Printers Glasgow

Bulletin No. 90 July 1976

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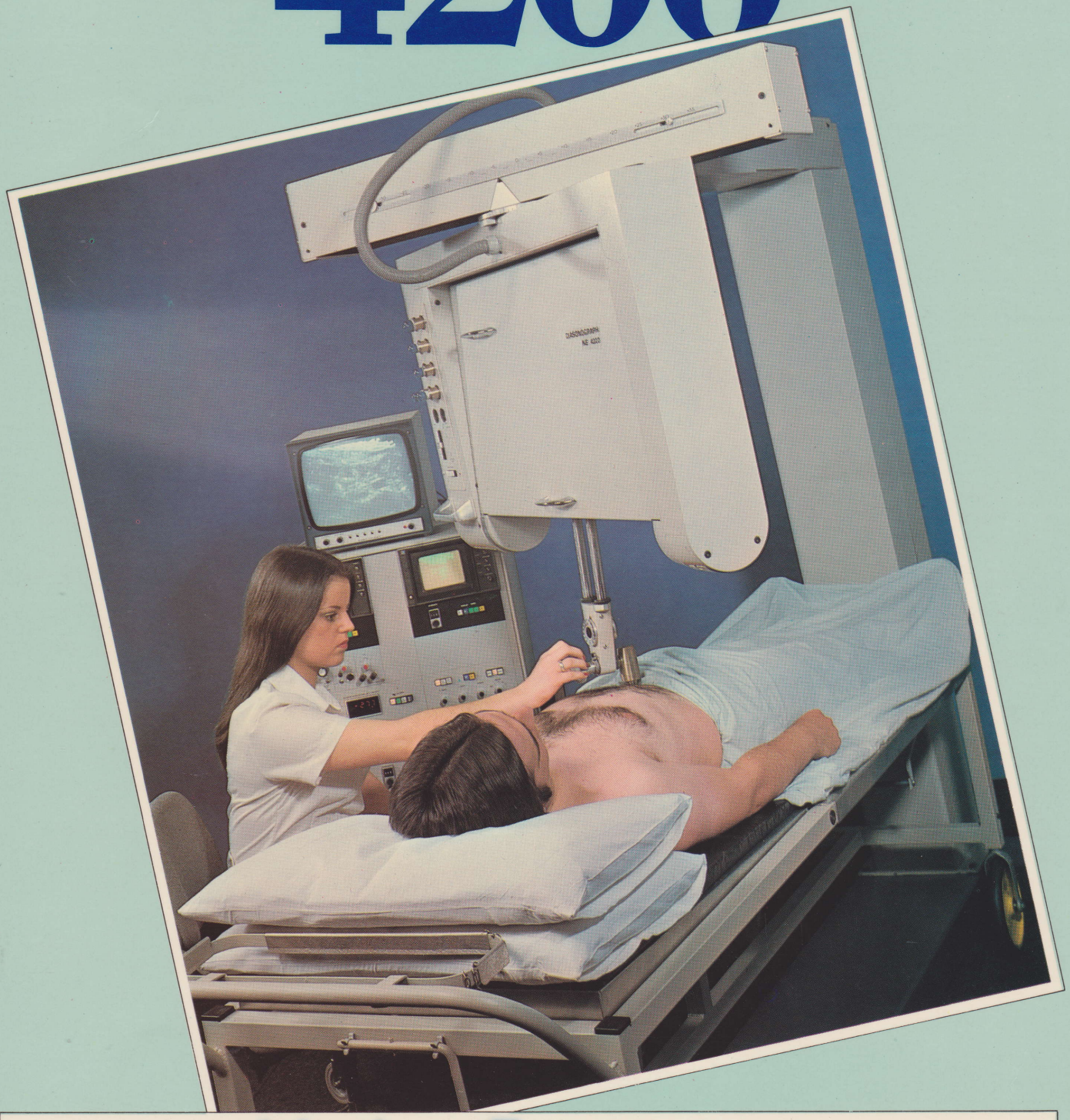
EMI Medical Ltd. Bulletin No. 112 - EMISONIC 4200. October 1977.

EMISONIC 4200



Nuclear Enterprises

EMISONIC 4200
(Formerly Nuclear Enterprises Diasonograph 4200)

Following the merger of Nuclear Enterprises with EMI in October, 1976, Nuclear Enterprises is now a full member of the EMI Group of Companies, world leaders in medical imaging systems.

Nuclear Enterprises pioneered the design and manufacture of advanced Ultrasound Scanners, which have been internationally acknowledged as setting performance standards in this field.

The combined research, development and manufacturing resources now provide a new, and expanded range of advanced Ultrasound Systems. This new range is being marketed under the 'EMISONIC' trade name through Nuclear Enterprises/ EMI Medical and their associated subsidiaries and representatives.

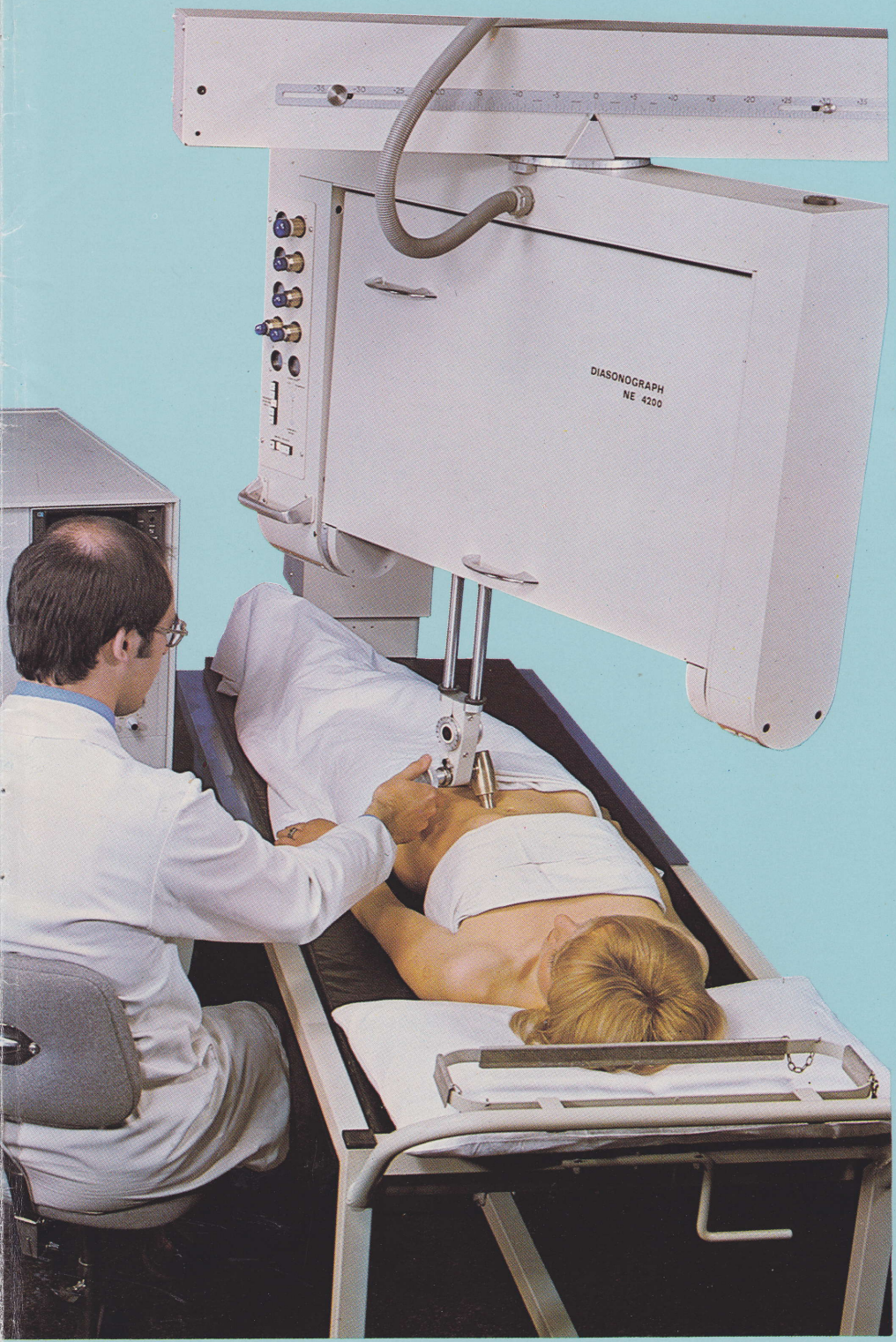
The EMISONIC Ultrasound range of equipment is supported worldwide, in all overseas markets, by specialised teams in marketing, installation and after-sales service.

EMISONIC 4200 with Greyscale Storage Display.

The EMISONIC 4200 is the latest addition to the world-leading range of diagnostic pulse echo, ultrasonic scanners developed by Nuclear Enterprises over the past ten years. It has full greyscale capability and provides a high standard of resolution, outstanding operational capability and reliability. The 4200 single transducer contact scanner incorporates as standard, facilities for displaying A-scan, Cross-Section (B-scan) and Time-Position (M Mode) scans. These facilities enable a comprehensive diagnostic service to be provided in the fields of obstetrics, gynaecology, general abdominal examinations of soft tissue structures, neurology, ophthalmology, endocrinology, gastroenterology and cardiology.

The basic Control Console includes well established high quality displays and the following facilities are provided as standard: display of swept gain waveform; calibrated intensity controls for displays; and amplifier processing mode selection. Careful ergonomic grouping and colour coding of pushbutton controls ensure simplicity of operation. The recording facilities offered include: photographs (Polaroid, 70mm film, X-ray film), strip chart and video.

The 4200 is designed to meet individual clinical requirements as fully and as simply as possible. The system is supplied complete with the 4205G Greyscale Storage Display and the 4103C Cardiac Module is an optional facility. The 4200 can be supplied in its basic form without storage greyscale if required.



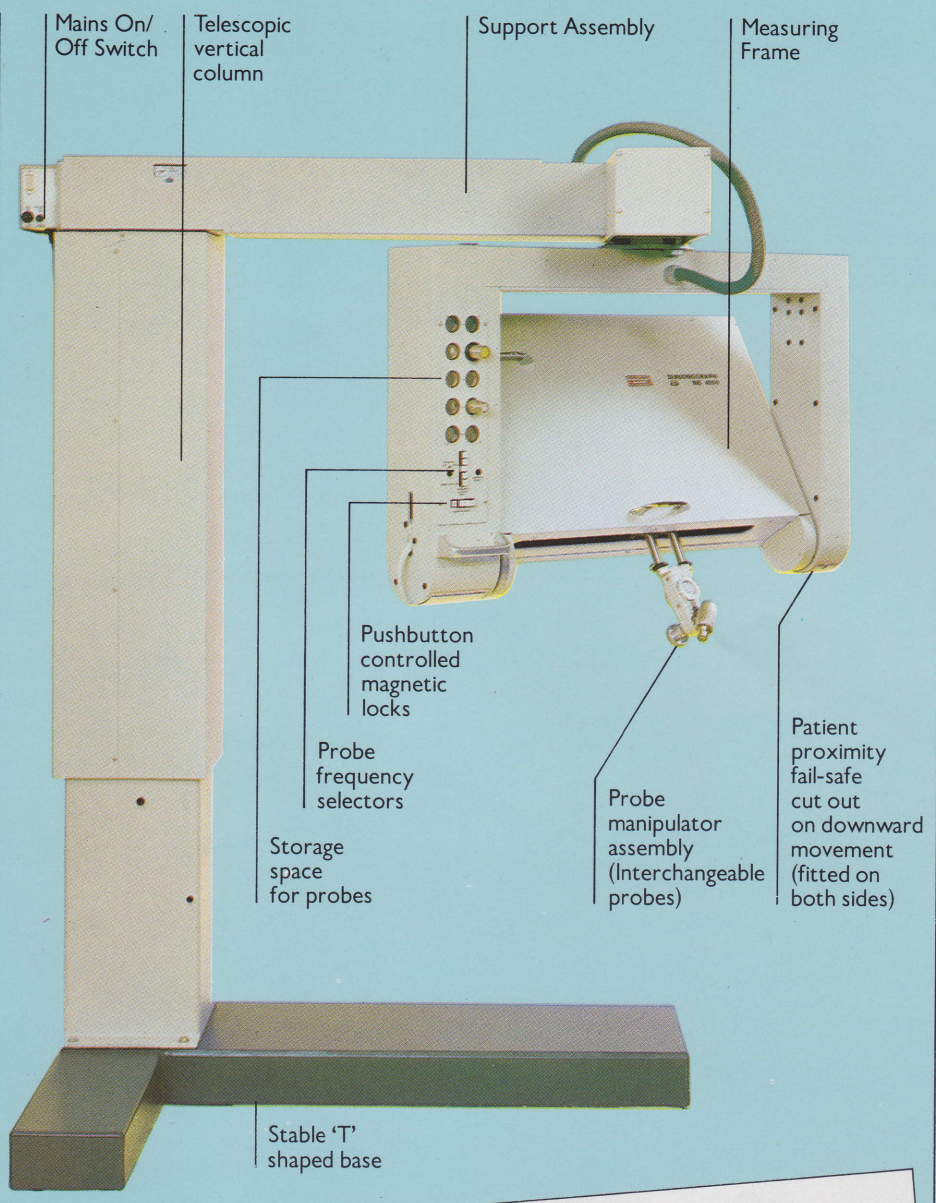
Scanning Assembly with exceptional measuring accuracy.

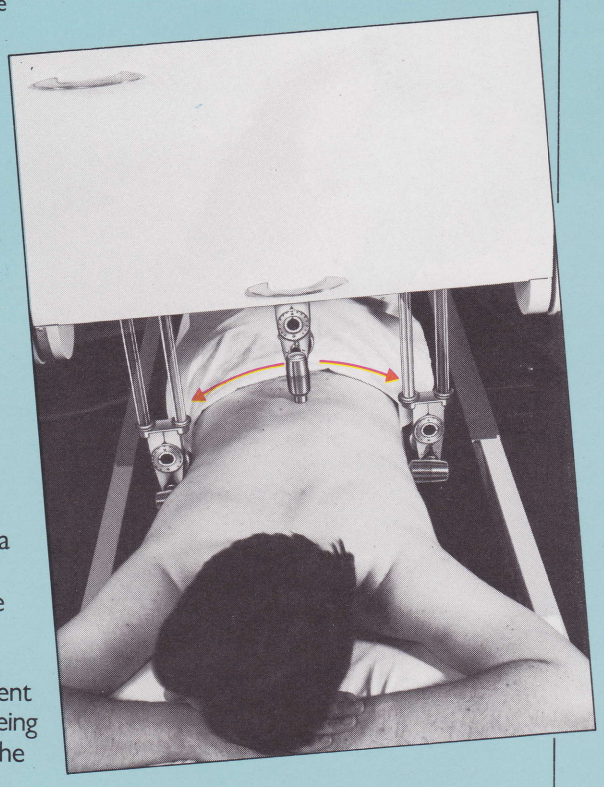
The 4200 measuring system affords an operator maximum freedom in positioning the scanning frame and allows the chosen scanning plane to be easily maintained and reproduced. The one-to-one measuring system coupled to the scanning probe has unequalled measuring accuracy, and all movements are calibrated. Rapid changing from longitudinal to transverse scanning is another important facility offered by the 4200 measuring system.

The 4238 2.5MHz long internal focus probe is supplied as standard with the system but a wide choice of interchangeable bayonet fitting probes for different applications is available. These probes may be exchanged without switching off the equipment, and changed from one side of the assembly to the other to suit the examination. The probe may be locked at a specific angle within its travel both for calibration and operational requirements as when it is used with a 4167 Biopsy Probe.

The **unique probe frequency selection** controls enable the operator to match the amplifier tuning to the transducer being used – thus optimising information production for a particular examination. **1.5MHz, 2.5MHz, 3.5MHz, 5.0MHz and 10MHz optimised frequency tuning is available as standard.**

Patient proximity detectors are provided to ensure that no hazard to a patient occurs from accidental lowering of the measuring frame or probe.

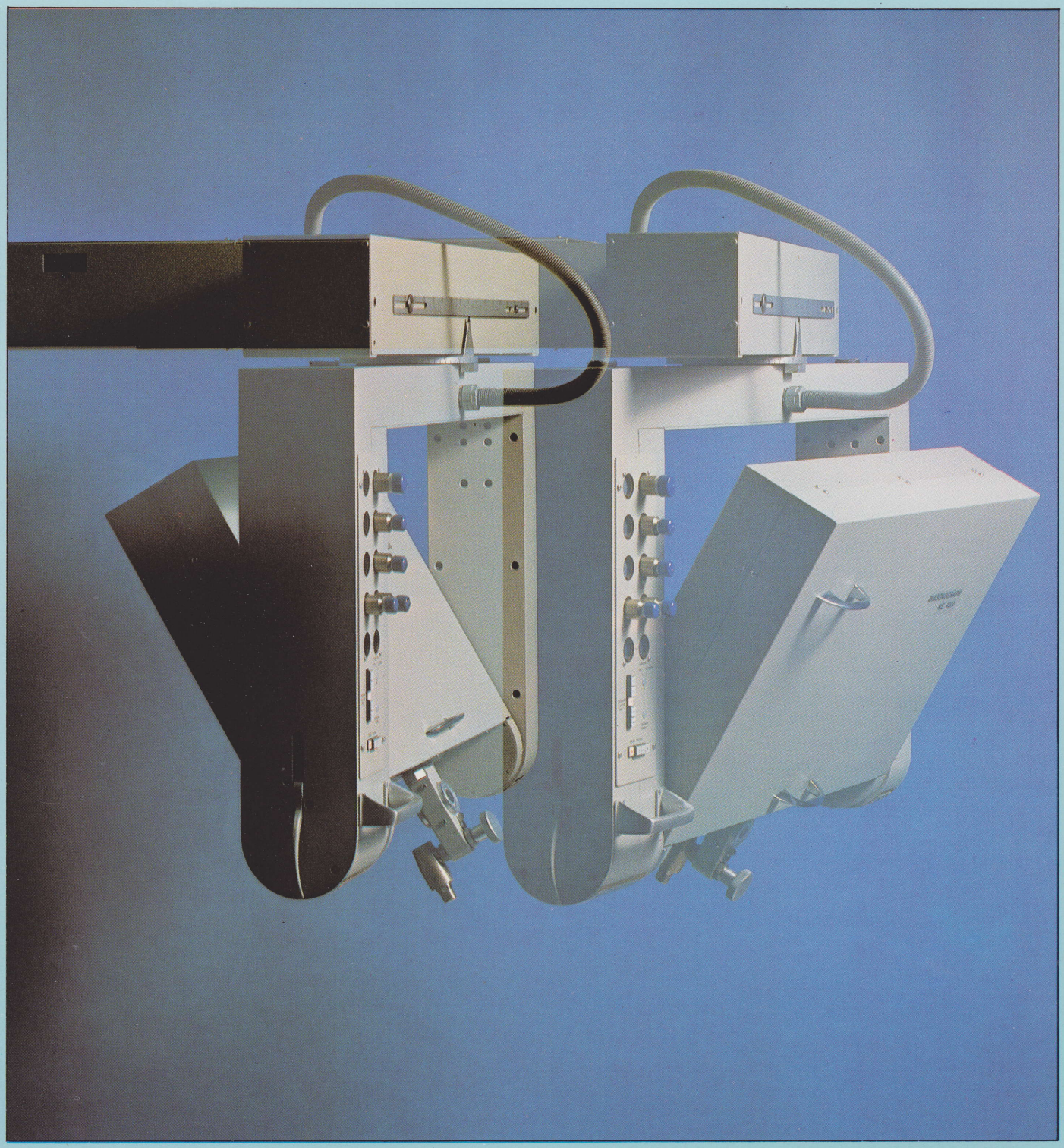




The ultrasonic power output in all Nuclear Enterprises scanners is reduced to the lowest possible level in order to minimise the dose energy to the patient, whilst maximising the use of it. The sensitivity of the equipment is controlled by reducing the transmitted power rather than using high power and reducing the gain of the system.

For most routine investigations a pulse repetition frequency of 600 pulses per second is used. A unique selectable velocity controlled pulse repetition system ensures that ultrasound transmission to the patient only takes place whilst scanning is being carried out at a rate controlled by the speed of scanning.

The transmitter output power and attenuation controls are accurately calibrated in decibels (dB).



An essential requirement for an ultrasonic scanning system is that the cross-sectional plane of interest in the patient is reproduced accurately and reliably. **The exceptional freedom of positioning of the 4200 measuring system allows the operator to change the plane of scan rapidly and minimises user fatigue.**

The majority of clinical investigations require a series of scans to be taken both transversely and longitudinally. Rapid interchangeability between planes is an important facility offered by the 4200. This is especially significant in busy clinics where the patient throughput is high.

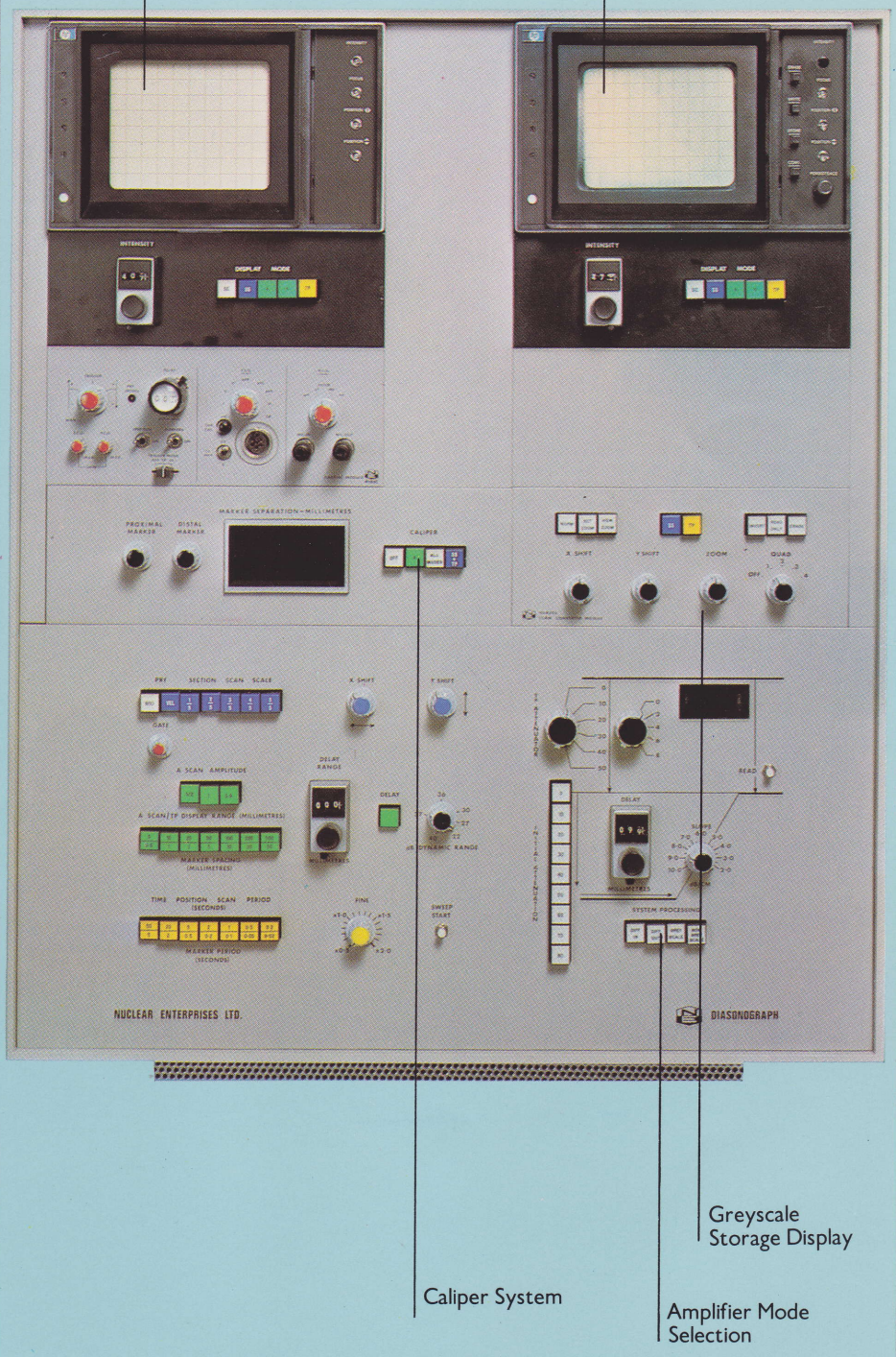
Control Console.

Two high performance display units are included as standard in the trolley-mounted electronics console. One is a short persistence HP 1333 and the other is a variable persistence/storage HP 1335, with foot-operated erase switch. A caliper device, a calibrated swept gain display, calibrated oscilloscope intensity controls, and amplifier processing mode selection are standard facilities in the electronics system. **The controls on the console are simple.** Colour coding is used to indicate function and illumination to indicate choice.

The comprehensive display facilities with the **wide choice of operating modes enables the 4200 to cover a wide range of applications.** It is possible to display section-scans (B-mode), A-scan, inverted A-scan, time-position scan (M-mode), and "open shutter" greyscale scans, including those from the storage greyscale unit, independently on each display.

Accurate measurements to ± 0.1mm of structural dimensions in vivo are possible with the caliper system. This has a large scale illuminated digital readout. The "bright-up" caliper pips can be displayed simultaneously on all modes.

The basic 4200 has a maximum dynamic range of 40dB, which may be reduced to 22dB allowing maximum grey tone range to be obtained of echoes of very similar amplitude arising from relatively homogeneous tissues.



4205G Greyscale Storage Display.

The 4205G Greyscale Storage Display allows the clinician to obtain high resolution greyscale scans, which are viewed and stored on the (300mm) TV monitor.

In greyscale scanning the grey level intensity is determined by the amplitude of the echo. A large echo is recorded as peak white, a low-level as very dark grey and intermediate strength echoes at the appropriate grey scale range. High quality scans showing the full range of greytones simplify the visualisation and interpretation of complex structures. Examination times are minimised and confident diagnoses can be made.

A feature of the 4205G is that it is possible to achieve almost complete freedom from overwriting effects which can seriously degrade a compound scan. The user may, without writing out the picture, "compound" a scan in order to collect as much clinically useful information as possible.

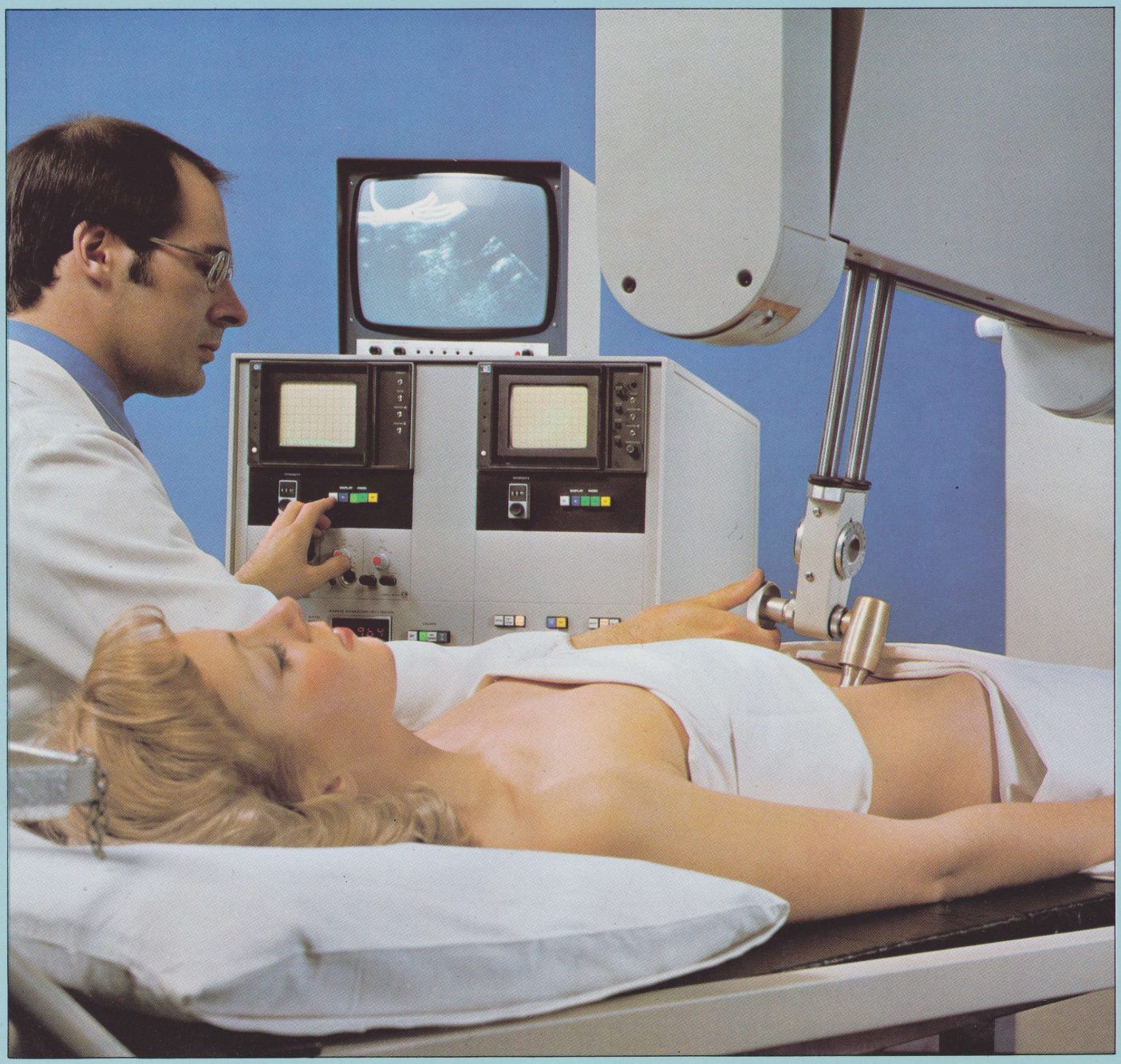
The quad option allows four scans to be viewed simultaneously on the monitor. Each of these scans may be erased independently or viewed at full screen size.

The 4205G time share, read/write facility allows the operator to see the build-up of a picture during the scanning process. Any area of interest may be preselected and, with the use of the zoom control, switched to cover the total area. A magnification 4×life size is possible without significant loss of resolution.

Time-Position and Section Scans may be displayed. By depressing the 'SC' button on the mode selection panel, the stored greyscale picture can be transferred to either console display for photography.

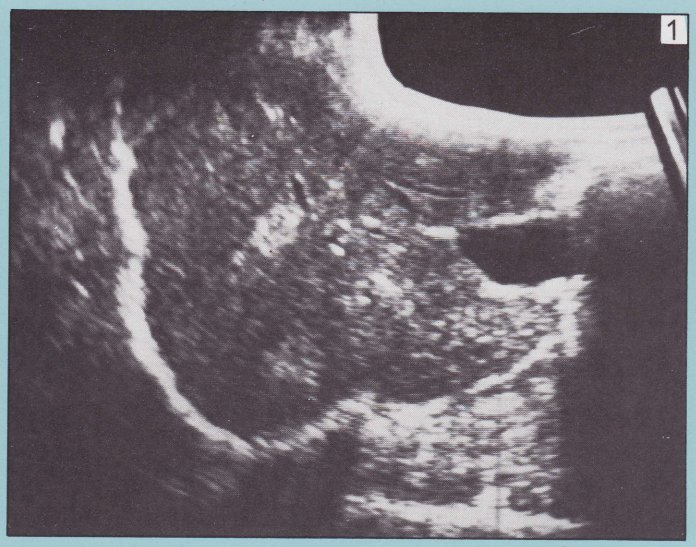
Selection of the "invert" pushbutton changes the displayed picture from positive to negative to facilitate negative film photography.

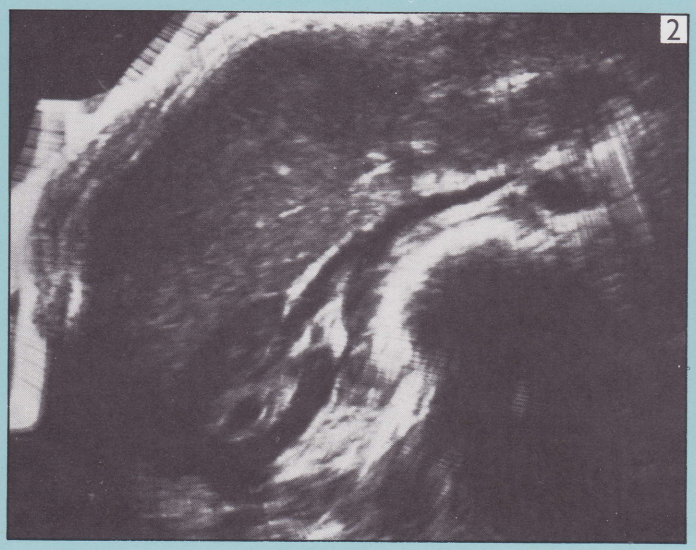
Permanent records may be obtained by conventional Polaroid photography. Alternatively, 70mm or X-ray film recordings can be made. As electronic signals are now available in standard TV form, slave or remote TV monitors may be used for display, and video tape recorders employed to store and replay information. Thus video monitors may be sited in lecture theatres or in consulting rooms.

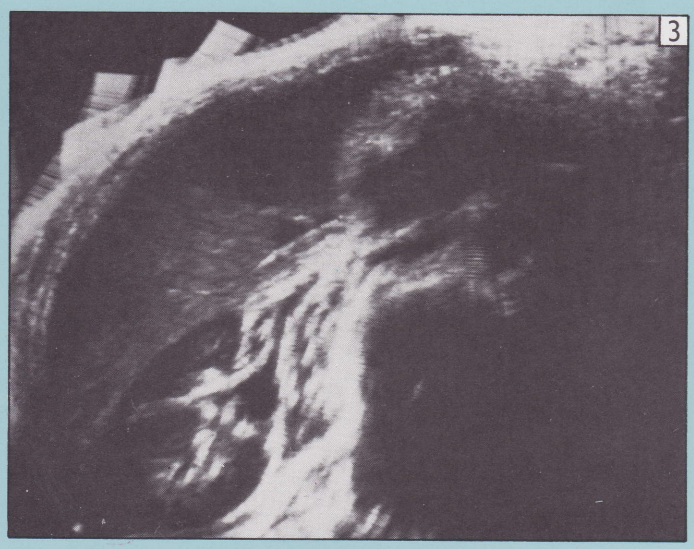


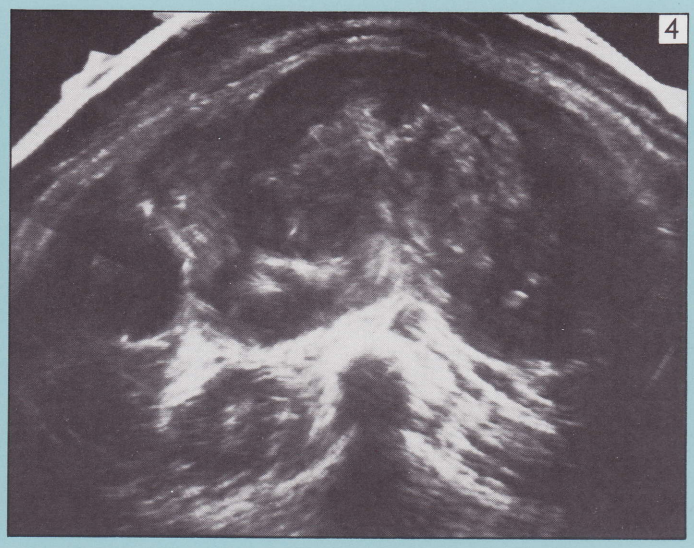
Clinical results.

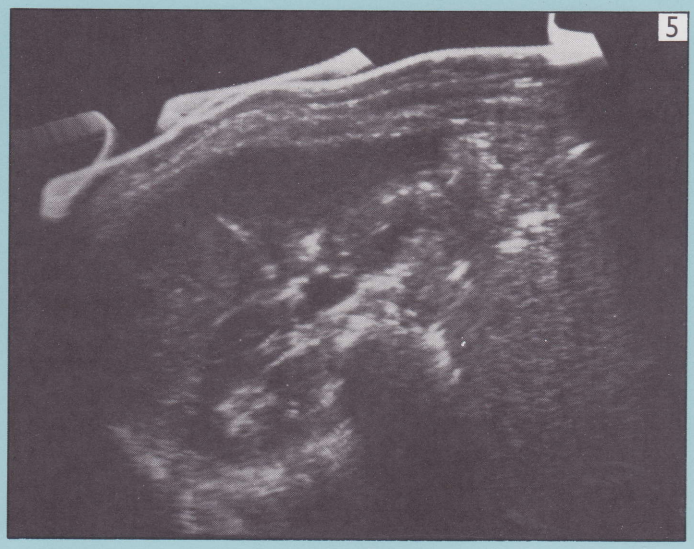
All result photographs on these pages are reproduced by courtesy of the Clinical Research Centre and Northwick Park Hospital, Harrow, Middlesex.

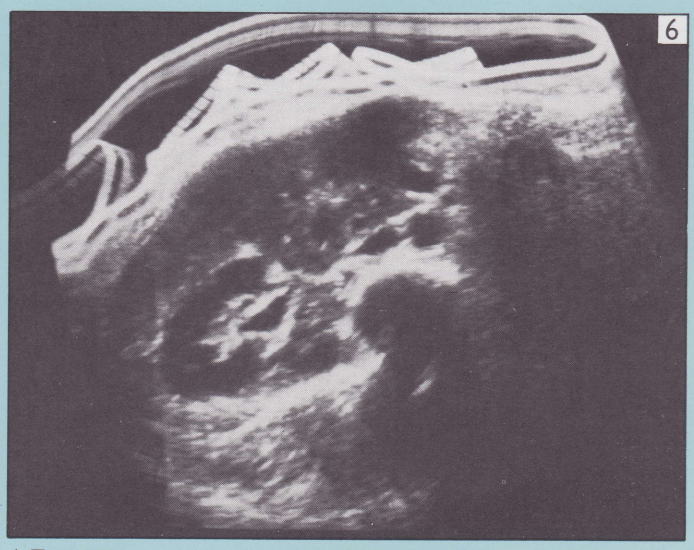








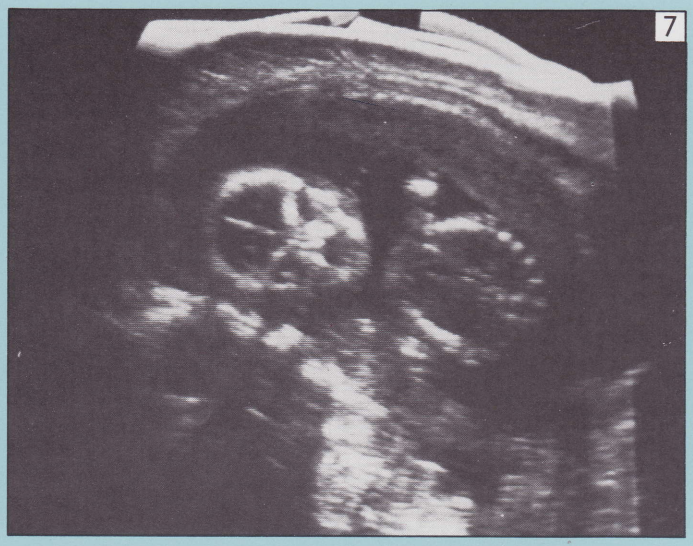


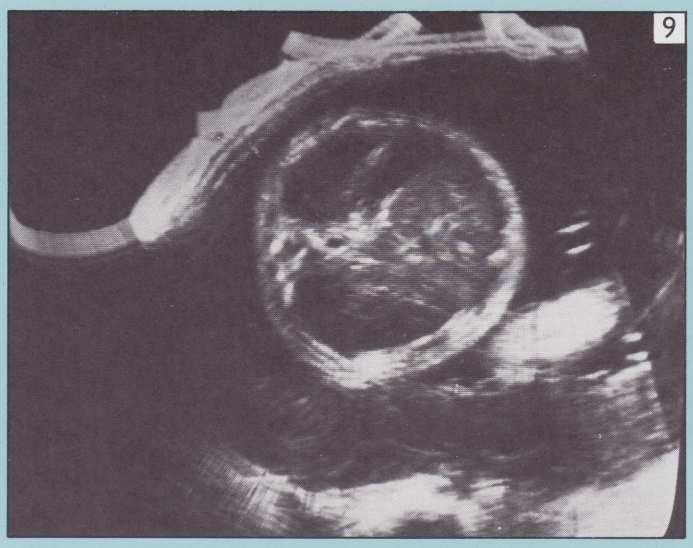


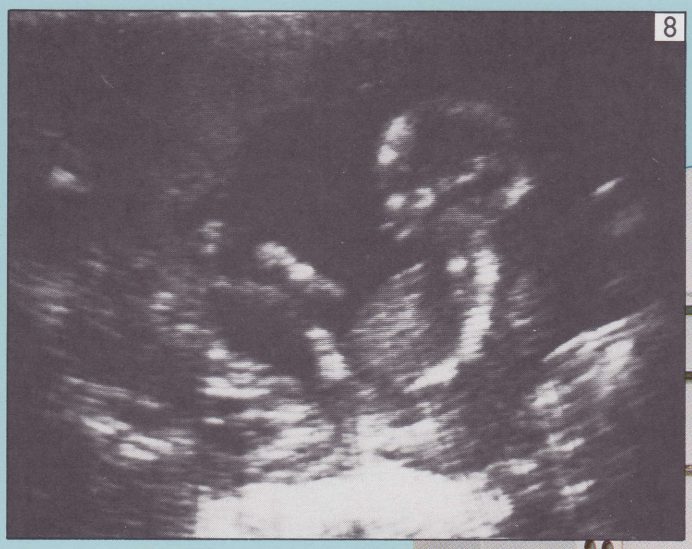
1. Longitudinal scan of right lobe of liver with metastases.
2. Transverse scan showing normal kidney and renal vein.
3. Transverse scan showing pelvis of kidney.

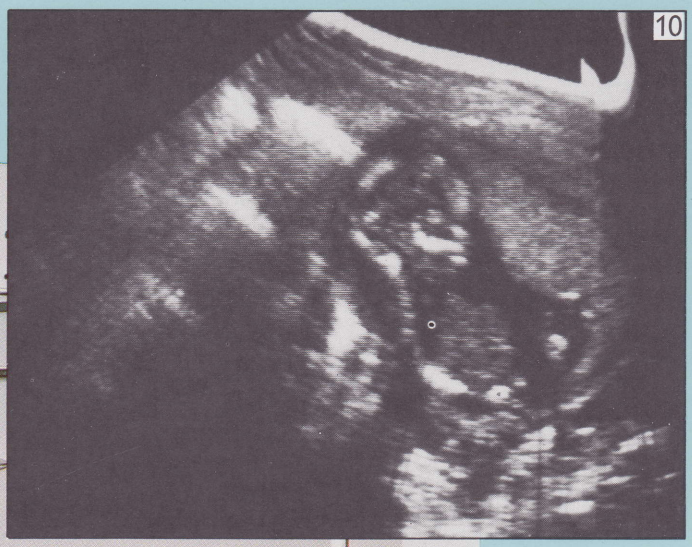
4. Transverse scan showing enlarged pancreas of patient with acute pancreatic necrosis.
5. Transverse scan showing head and body of normal pancreas.
6. Transverse scan showing slightly enlarged head of pancreas with dilated ducts within the head.

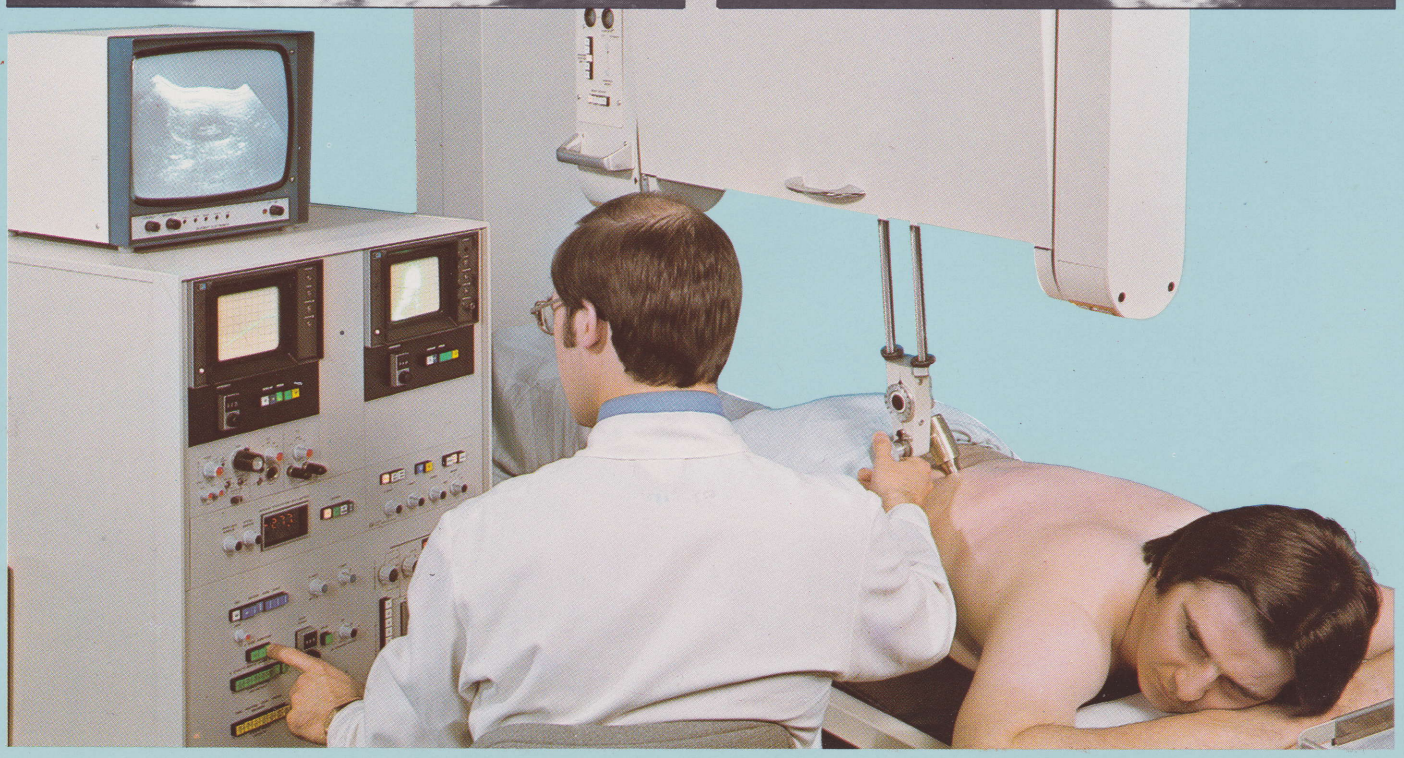
7. Longitudinal scan (anterior view) of fetus showing both orbits.
8. Longitudinal scan of fetus at 14 weeks.
9. Cross section of fetal abdomen at 36 weeks.
10. Longitudinal scan of fetus at 13 weeks showing detail of face.











Accessories.

1. 4211 Patient Trolley with paper sheet dispenser and facility for raising or lowering the patient's head to allow adjustment of posture for comfort and/or ease of investigation.
2. 4141 Specially designed Water Bath for immersion scanning techniques.
3. 4110 Echo Generator for system calibration checks.
4. 4126 Calibration Jig for system registration checks.
5. 4210 Photographic Facility for use with the 4205G Greyscale Storage Display comprising a 6 inch (152mm) TV monitor, hinged camera adaptor and a Shackman Super Seven Polaroid Camera. This TV monitor can be used as a remote viewing monitor and may be placed away from the main scanning area.
6. 4232 Shackman Super Seven Polaroid Camera as used in the 4210 except that it is supplied with an AS20 adaptor allowing photographs to be taken from the HP 1335 and HP 1333 oscilloscopes.
7. 4235 70mm Linhof Film Holder and M4 International Module for the Shackman Super Seven Camera as an alternative to the Polaroid Back.
8. 4143 Multi-Format X-ray Film Holder for the Shackman Super Seven Camera giving six photographs, each 63×78mm, on one sheet of 203×254mm X-ray film. (Bulletin No. 103.)
9. 4108 Video Cartridge Recorder is basically a PAL Colour half inch (12.7mm) magnetic tape recorder. It is specially modified for recording high resolution scans from the 4205G Greyscale Storage Display. Operation is extremely simple, all major controls being operated by pushbuttons. The standard half inch (12.7mm) cartridge tapes supplied have a playing time of 36 minutes and it is possible to produce a simultaneous audio recording using the dubbing facility. (Bulletin No. 99.)
10. 4103C Cardiac Module extends the use of 4200 systems in the field of cardiology. This module allows simultaneous presentation of the Time-Position Scan (M-mode) ECG and PCG traces. In addition, the display of ECG triggered cross-section scans of moving structures is possible. The pulse repetition rate is raised to 1,200 pulses per second and synchronisation of the Time-Position Scan from the ECG waveform can be achieved. (Bulletin No. 95.)
11. 4105 Fibre Optic Recording Oscilloscope provides a flexible, non-integrating display-record facility. In addition to continuous chart recording (M-mode) conventionally used in cardiological investigation, the 4105 can provide permanent hard copy records of section scans, time-position scans (M-mode) and A scans. All 4200 systems are manufactured with a fibre optic recorder interface socket. Details of other recorders available on request. (Bulletin No. 91.)
12. 4144 Video Character Generator enables the operator to type information onto the greyscale scan displayed on the video monitor. Patient details, date and areas of anatomical interest may be marked and recorded. This facility is of particular importance if negative film recording methods are used. (Bulletin No. 101.)

Ultrasonic Transducers.

A 2.5MHz flat faced, 13mm diameter, long internal focus probe is supplied as standard with every 4200 system. This is suitable for most routine diagnostic imaging. However, large improvements in resolution and lesion detection can be expected in a limited area of interest by selection of a specialised transducer from the wide range available for both scanning and hand-held applications.

Each transducer is marked with a reference number, frequency, diameter and focal length. Metal-based bayonet fitting transducers ensure good grounding and freedom from external electrical interference.

Full details of the complete range are available on request. The following list details some of the most commonly used transducers.

Recommended Scanning Transducers

1.	4238	Long Internal Focus Bayonet Fitting Probe 2.5MHz, 13mm active element diameter.	Supplied as standard
2.	4310	Non-focused Bayonet Fitting Probe. 1.5MHz, 19mm active element diameter. Good angular resolution in far range.	Deep penetration, obese patients posterior placenta
3.	4311	Long Internal Focus Bayonet Fitting Probe. 1.5MHz, 19mm active element diameter. Angular resolution improved in mid range.	
4.	4325	Non-focused Bayonet Fitting Probe. 2.5MHz, 13mm active element diameter. Optimum angular resolution in far range	General purpose liver and kidney, and obstetric applications.
5.	4323	Long Internal Focus Bayonet Fitting Probe. 2.5MHz, 19mm active element diameter.	
6.	4383	Medium Internal Focus Bayonet Fitting Probe. 3.5MHz, 13mm active element diameter. Lower penetration, high angular and range resolution in mid range.	Pancreas, thyroid, breast and scanning of older children.
7.	4388	Long Internal Focus Bayonet Fitting Probe. 3.5MHz 19mm active element diameter. High angular and range resolution at mid range.	
8.	4351	Non-focused Bayonet Fitting Probe. 5.0MHz, 13mm active element diameter. High range resolution penetration limited.	Paediatric, ophthalmology and thyroid.
9.	4354	Long Internal Focus Bayonet Fitting Probe. 5.0MHz, 13mm active element diameter. Improved angular resolution in mid range.	
10.	4167	2.5MHz Biopsy Probe with a central aperture allowing use of needles up to 1.96mm in in diameter.	Designed primarily for amniocentesis but also useful for cyst aspiration.
11.	4364	Long Internal Focus Probe. 1.5MHz, 19mm active element diameter.	Echoencephalography
12.	4367	Medium Internal Focus Probe. 2.5MHz, 13mm active element diameter.	General purpose cardiac
13.	4394	Medium Internal Focus Probe. 3.5MHz, 13mm active element diameter.	Adult and paediatric cardiac
14.	4343	Non-focused Probe. 5.0MHz, 13mm active element diameter.	Neonate and paediatric cardiac
15.	4340	Non-focused Probe. 5.0MHz, 6mm active element diameter.	Ophthalmic
16.		Aspiration/Biopsy Transducer. Non-focused, 2.5MHz, 13mm active element diameter with 2.4mm central aperture (aperture accommodates 14 gauge aspiration or biopsy needle).	

EMISONIC 4200 Specification.

Operating controls are distributed in ergonomic groupings to assist operators without specialised knowledge of physics or electronics. Controls are calibrated in tissue where appropriate (see Control Panel right).

Sensitivity Control by Transmitter Attenuator:
0 to −50dB in 10dB steps, +0 to −8dB in 2dB steps

Swept Gain (Depth Compensation): Initial reduction:
0 to 80dB in 10dB steps
Initial Delay: 0 to 250mm continuously variable.
Slope: 1.5 to 10.5dB/cm continuously variable.

Picture Scales: Section Scan only:
1/5, 2/5, 3/5, 4/5, 5/5 of full-scale.
Scanned Area:
Horizontal: 500mm nominal
Vertical: 250mm nominal
Probe Rotation: ±135° from vertical in the plane of scan.

A-Scan Display:
Half screen or whole screen or Swept Gain. Inverted V on half or whole screen.

A-Scan Range: 5,10, 20, 50, 100, 200, 500mm switched steps.

A-Scan Delay: 0 to 500mm continuously variable. A delay set/use switch allows the operator to preselect the region to be examined.

Time-Position Display: Intensity modulation of horizontal sweep.

Vertical Sweep (Time-Position only):
Single vertical sweeps triggered by push button. Scan period continuously variable between ×0.5 and ×2 of the following ranges:–
50, 20, 5, 2, 1, 0.5, 0.2 in seconds.
Horizontal Range (Time-Position Display): 5, 10, 20, 50, 100, 200, 500mm switched steps (same control as A-Scan range).

Operating Frequencies: 1.5, 2.5, 3.5, 5.0, 10MHz

Energy: Maximum available ultrasonic intensity is a function of the probe in use. Using the standard range of probes, the following figures are typical.
Normal Mean
Acoustic Power: 1.5MHz 0.70 microwatt
2.5MHz 6.92 microwatt
5MHz 84.0 microwatt
Maximum Available Mean Acoustic
Power: 1.5MHz 7.0 milliwatt
2.5MHz 6.92 milliwatt
5MHz 2.7 milliwatt

NOTE: Maximum operating sensitivity of the equipment is controlled by adjustment of the output power.

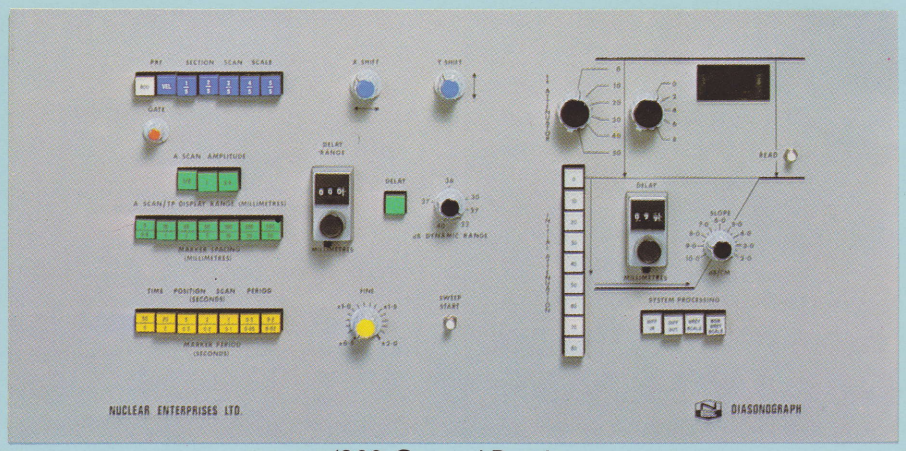
Transmitter Pulse Repetition Rate:
(a) Fixed 600 pulses/second.
(b) Velocity controlled (section scan only) 0 to 1800 pulses/second.
(c) With cardiac facility (time position scan) 1200 pulses/second.
(d) Triggered section scan with cardiac facility 10 to 100 pulses/cardiac cycle.

Horizontal Delay (Time-Position Display):
0 to 500mm continuously variable (same control as A-Scan range).

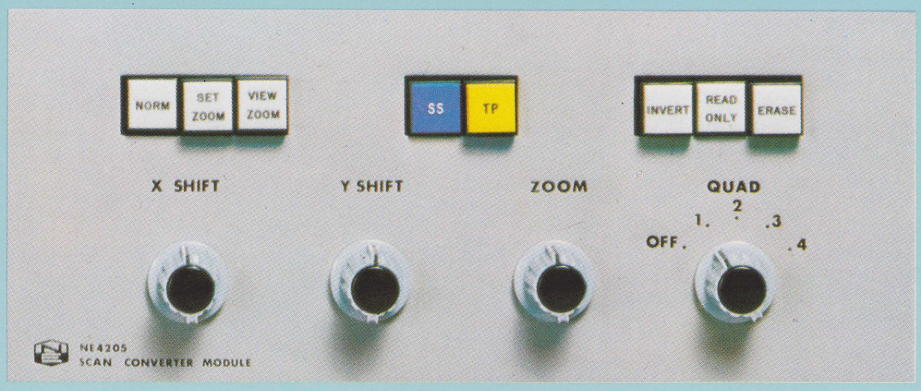
Dynamic Range of Echoes: 40 to 22dB continuously variable.

Ultrasonic Caliper: Marker separation 0 to 199.9mm in 0.1mm steps over entire A-Scan range.

Power Requirements: 100 to 120/200 to 240V ac. 50/60Hz 800VA.



4200 Control Panel



4205G Control Panel

4205G Greyscale Storage Display Specification.

Description: High resolution unit for storage and display of video and graphic information.

Storage Medium: Princeton Electronic Products Scan Conversion and Image Storage Unit. (Special Version).

Resolution: 1350* to 2100**
* TV lines per diameter at 50% depth of modulation.
** TV lines per diameter limiting resolution.

Output: Television IV video signal with composite synch.
625 lines, 50 fields or 525 lines, 60 fields.

Display Unit: 300mm monochrome monitor.

Front Panel Controls:
(Continuously variable)
"Zoom"–magnifies the image being viewed.
"X"–controls horizontal position of the area being magnified by zoom.
"Y"– controls vertical position of the area being magnified by zoom.

Pushbutton switch selection:
"Norm"–normal full-size viewing of the image.
"Set-Zoom"–normal full-size image with "region of interest" superimposed. This is a rectangular box defining the area to be covered by the zoom. Its position and size are controlled by the X, Y and "Zoom" controls.
"View Zoom"–enlarges area within "region of interest" to fill the screen.
"Invert"–changes the displayed picture from positive to negative. Used for making positive pictures on negative film.
"Read only"–stops the writing of images on the storage unit when the probe is moved.
"Erase"–clears the screen for the next scan.
"SS"–selects section scan picture storage.
"TP"–selects time-position scan picture storage.
"Quad Option"–allows 4 scans to be viewed simultaneously and erased independently.

Rear Panel Outputs/Inputs: "Video Out 1 and 2" –Two independent IV composite synch video signals for TV monitor or video tape recorders.

Dynamic Range of system with 4205G 36dB

Greyscales Displayed: 10

Space saving

The ergonomic design of the EMISONIC 4200 system ensures maximum use of the available floor space and affords considerable freedom of choice in the layout of the examination room.

Control Console Cabinet:

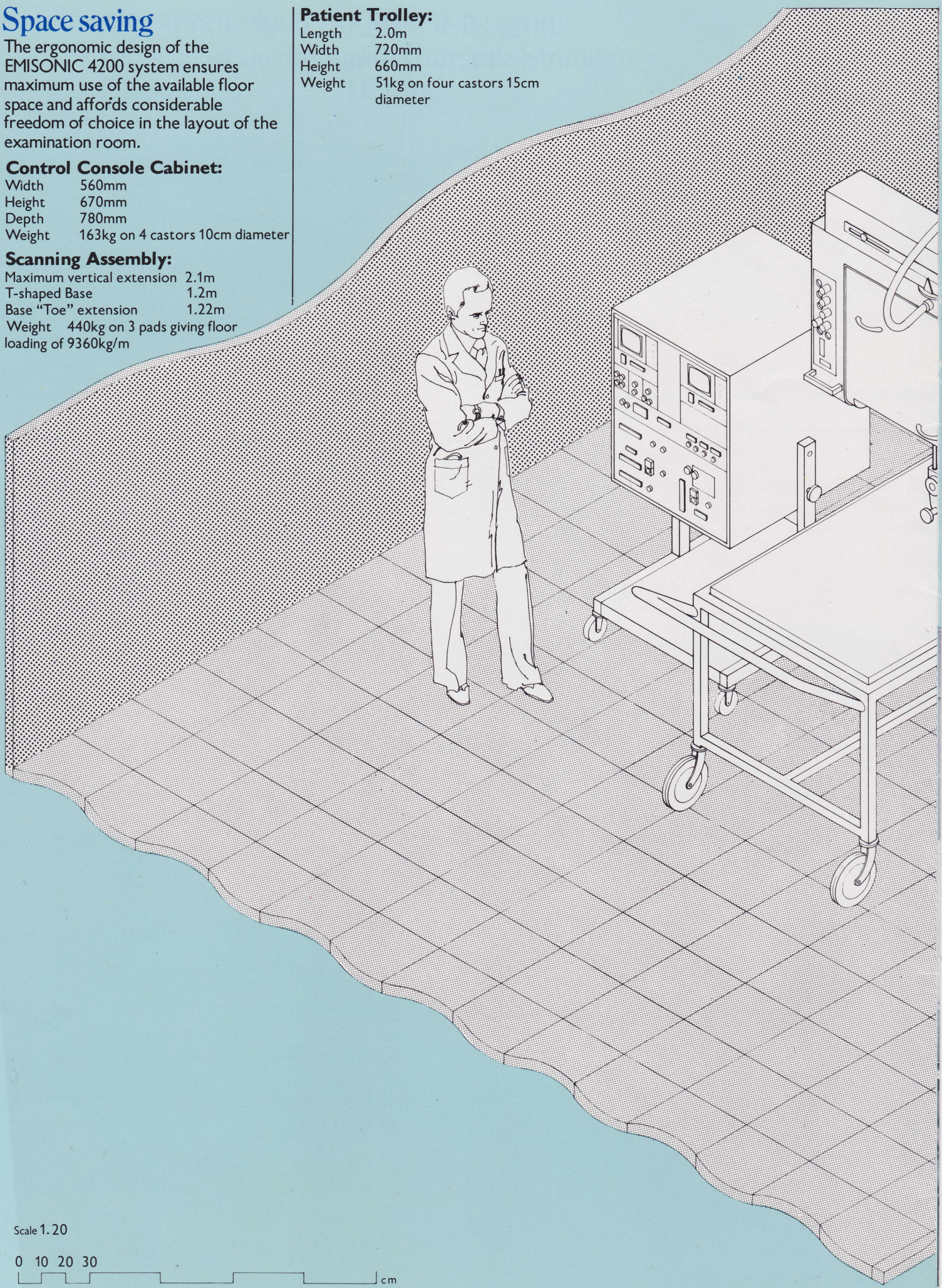
Width	560mm
Height	670mm
Depth	780mm
Weight	163kg on 4 castors 10cm diameter

Scanning Assembly:

Maximum vertical extension	2.1m
T-shaped Base	1.2m
Base "Toe" extension	1.22m
Weight	440kg on 3 pads giving floor loading of 9360kg/m

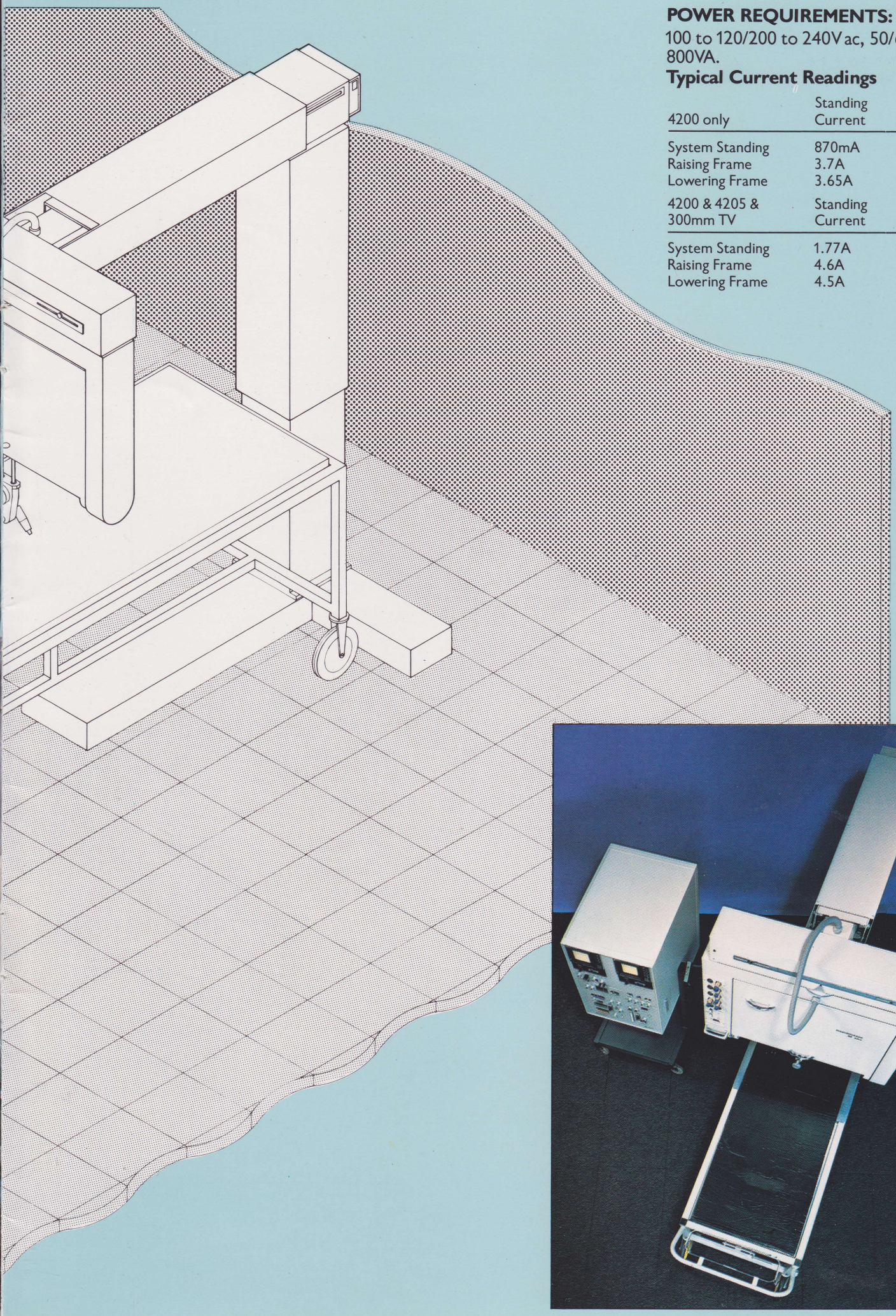
Patient Trolley:

Length	2.0m
Width	720mm
Height	660mm
Weight	51kg on four castors 15cm diameter



Scale 1.20

0 10 20 30 cm



POWER REQUIREMENTS:

100 to 120/200 to 240V ac, 50/60Hz, 800VA.

Typical Current Readings

4200 only	Standing Current	Initial Surge
System Standing	870mA	
Raising Frame	3.7A	4.2A
Lowering Frame	3.65A	4.2A
4200 & 4205 & 300mm TV	**Standing Current**	**Initial Surge**
System Standing	1.77A	
Raising Frame	4.6A	6A
Lowering Frame	4.5A	5A

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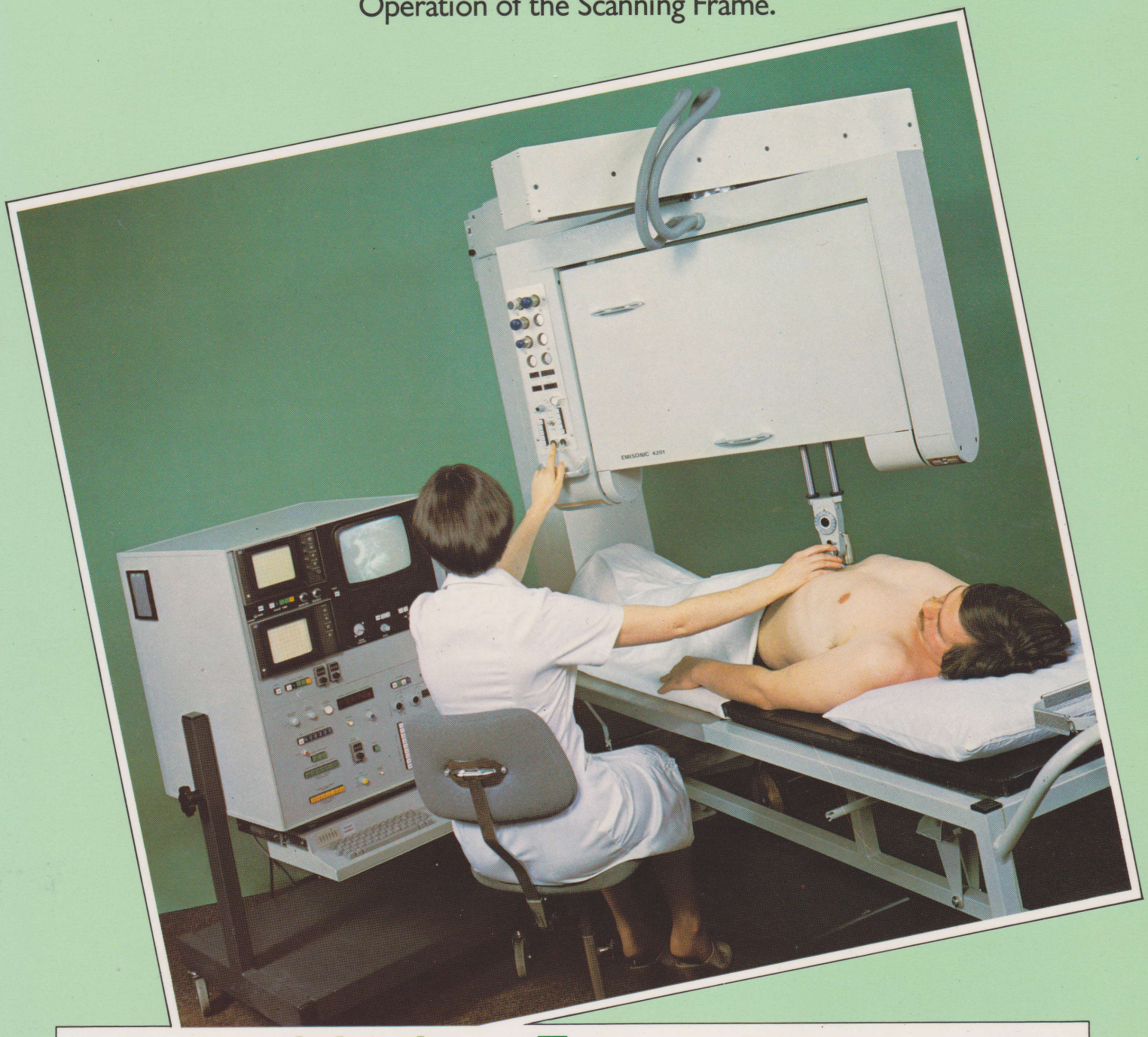
EMISONIC 4201

A Compact Ultrasound Scanner with Automatic Entry of Scanning Parameters and Motorised Operation of the Scanning Frame.



Nuclear Enterprises

EMISONIC 4201

The EMISONIC 4201 extends the EMISONIC Range of Ultrasound Scanners. In addition to providing the high clinical performance and reliability of the EMISONIC 4200, the 4201 provides a number of supplementary feature benefits.

Compact 3-Display Control Console

This incorporates an 'A' MODE DISPLAY for TGC curve and depth recognition, a video greyscale PHOTOGRAPHIC DISPLAY and a large screen VIEWING DISPLAY. The viewing display and the photographic display have the capability of independent video invert, which allows the operator to view the scans in one format and simultaneously take photographs of the clinical scans in another.

Automatic Patient Data Entry with Unique Data Security Interlocks

Patient details and scanning parameters are entered automatically into viewing and display monitors. Special interlock circuitry ensures complete protection of patient data.

Motorised or Manual Operation of the Scanning Frame

Operator fatigue is minimised by the use of the motorised drive of the scanning frame between scans.

Incremental Power Assisted Drive with 2mm to 40mm Range

Precisely separated parallel scans in increments of 2mm, 5mm, 10mm, 20mm, 30mm or 40mm can be activated by finger tip controls in either the longitudinal or transverse direction. In addition, accurate digital readout of the scanning frame position is displayed on the scanning frame control panel for convenient viewing.

Unique Resolution Enhancement Facility

This allows clinical details and information on the high resolution greyscale storage display to be maximised.

Pushbutton Auto-Centering of Image

Plus an independent 'Y' position control.

Simultaneous Viewing 'A' Scan and TGC Waveforms

Electronic Joy Stick Control of Region of Interest Box

Subsequent depression of the 'view zoom' magnifies this region to full screen size.

Pushbutton Tuned Amplifier Frequency Selection

1.5, 2.5, 3.5, 5.0 and 10MHz operation.

Keyboard Control of Alpha-Numeric Data

Information in addition to that automatically displayed on the scan may be added using the manual keyboard. If required, the digital reading of the caliper separation may also be incorporated into the data displayed by simple pushbutton operation.

High Resolution Transducers

Two specially selected transducers are provided as standard. One 2.5MHz 19mm diameter long internal focus 4323 and one 3.5MHz 19mm diameter long internal focus 4388. A standard range of transducers and optional accessories is available with all EMISONIC systems.

Adjunctive Real-Time Option Available

Request details of the EMISONIC Spinner System (Bulletin No 118)

EMISONIC Range

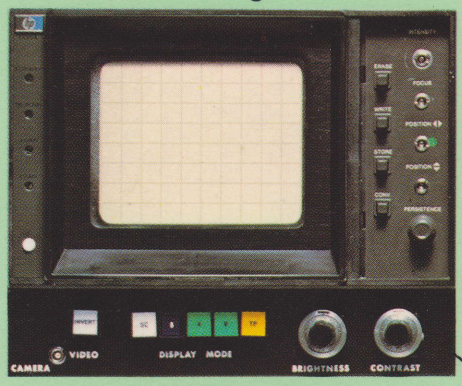
Following the merger of Nuclear Enterprises with EMI in October 1976, Nuclear Enterprises is now a full member of the EMI Group of Companies, world leaders in medical imaging systems.

Nuclear Enterprises pioneered the design and manufacture of advanced Diasonograph Ultrasound Scanners, which have been internationally acknowledged as setting standards in this field.

The combined research, development and manufacturing resources now provide a new, and expanded range of advanced Ultrasound Systems. This new range is being marketed under the 'EMISONIC' trade name through Nuclear Enterprises, EMI Medical and their associated subsidiaries and representatives.

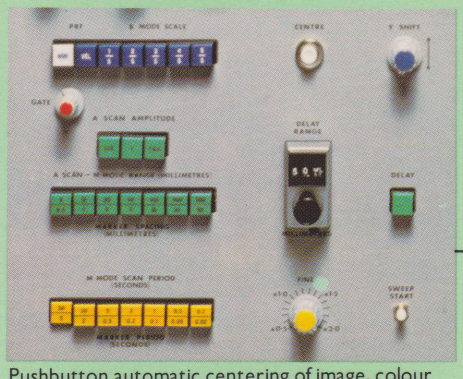
The EMISONIC Ultrasound range of equipment is supported world-wide, in all overseas markets, by specialised teams in marketing, installation and after-sales service.

The EMISONIC 4201 has been designed with the needs of the clinician in mind. Minimum patient discomfort and trauma, high patient throughput, trouble free system operation and high resolution clinical data provided by the 4201 make a major contribution to a sophisticated ultrasound scanning service.

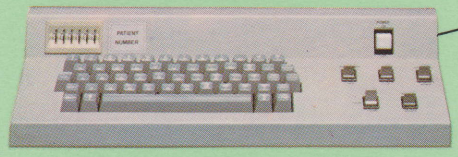


High definition photographic display.

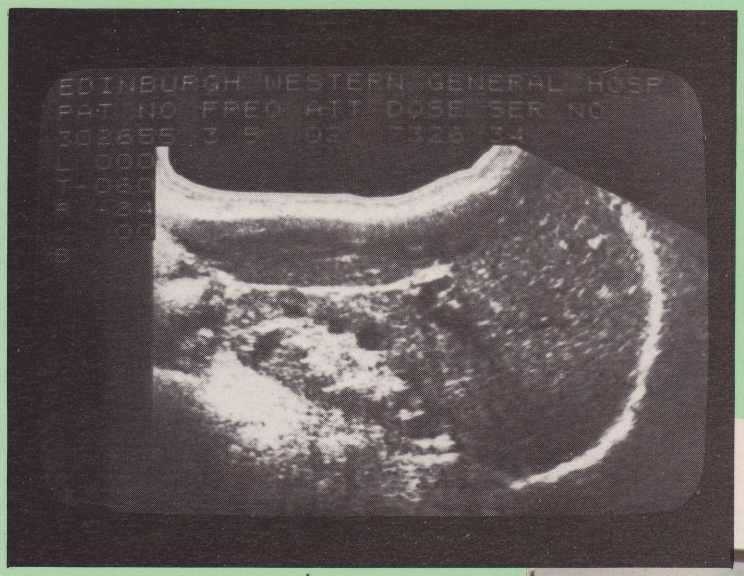
0.1% accurate digital caliper. Pushbutton readout onto display.



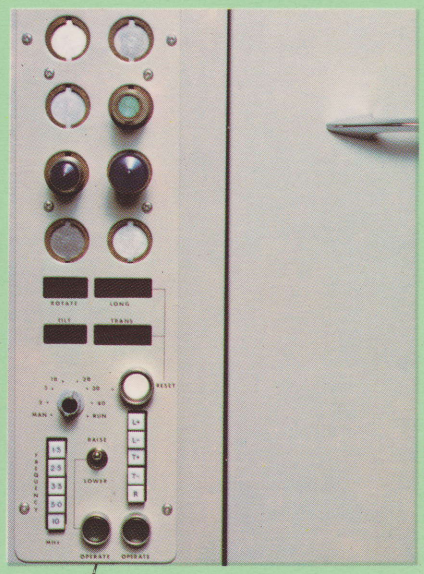
Pushbutton automatic centering of image, colour coded controls of 'B' Scan, 'A' Scan and 'M' mode operations.



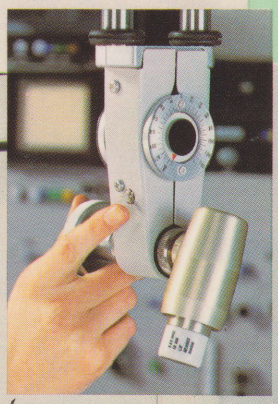
Keyboard entry of patient data to displays.



Automatically recorded scan data on clinical record.



Probe storage– digital read-out of scanning frame position and angulation–stepping motor control– tuned amplifier frequency selector.



Finger tip two-directional control of stepping motor.

EMISONIC 4201 Condensed Specification

The EMISONIC 4200 specification (see Bulletin 112) is also applicable to the EMISONIC 4201. In addition, EMISONIC 4201 includes the following features:

Motorised Movement of Scanning Frame Assembly (3 Modes)

Manual:	Functions as EMISONIC 4200
Step:	Selectable in 2mm, 5mm, 10mm, 20mm, 30mm or 40mm discrete steps. Actuation is by pushbutton on scanning frame control panel or by probe assembly pushbuttons. Direction of travel preselected for L+, L−, T+, T−. (L = Longitudinal, T = Transverse.)
Run:	Continuous motorised movement along L ± or T ± axes dependent on axis selected. Actuation as for Step.

Alpha Numeric Data Displayed on Scanning Frame

The position of the scanning plane with reference to a predetermined anatomical landmark is shown continuously on a digital display (±1mm accuracy). The angles of tilt and rotation of the scanning plane are also displayed.

Keyboard Data Entry: 2 pages of display available

Page 1: Automatic Fixed Data Entry

The following protected and interlocked alpha numeric data can be displayed on all monitors.

Title	Preprogrammed to customer's requirements, e.g. hospital name
Patient Number	PAT NO +6 digits
Frequency	FREQ +2 digits with decimal point as selected on scanning frame.
Attenuation	ATT +2 digits as selected on electronic console transmitter setting.
Dose	DOSE +4 digits ×100. This count is the number of transmitter pulses received by the patient. »» Indicates a dose of greater than 10^6 pulses. The internal dose counter is reset when the PAT NO is changed.
Scan Serial Number	SER NO +2 digits. Option 1: Updates by one for each combination of photograph and storage greyscale erase. Option 2: Updates by one for every storage greyscale erase. Options 1 and 2 are switch selectable; on both SER NO is automatically reset to 0 when PAT NO is changed.
Longitudinal Movement	'L' 3 digits with sign, either + or −. Units mm.
Transverse Movement	'T' 3 digits with sign, either + or −. Units mm.
Rotational Movement	'R' 2 digits with sign, either + or −. Units degrees.
Tilt Movement	/ 2 digits with sign, either + or −. Units degrees.
Marker Separation	'B' 4 digits with decimal point. Entered onto display by depressing "ENTER READING" pushbutton on console front panel. May be updated at any time.
Interlock	If any of the above parameters, with the exception of marker separation, are changed during a scan further writing of the picture will be prevented, thus ensuring integrity of alpha numeric data.

Manual Alpha Numeric Data Entry

Data may be entered in the free area of the display by use of the video character generator keyboard.

Page 2. Unrestricted Data Entry

No interlock arrangements. Data may be entered as required by use of the video character generator keyboard.

Nuclear Enterprises Limited
* Sighthill, Edinburgh EH11 4EY Scotland
Telephone 031-443 4060
Cables: Nuclear, Edinburgh Telex 72333
*Registered office. Registration No 31256 Scotland.

North America
EMI Medical Inc.
3605 Woodhead Drive,
Northbrook, Illinois 60062
Telephone (312) 291 4444
Telex 289450

Bulletin No. 116
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A member of the EMI group
International Leaders in music, electronics and leisure.

Design Forth Studios Edinburgh. Photography Inglis Stevens Edinburgh. Printed in England by the Oxley Printing Group Ltd

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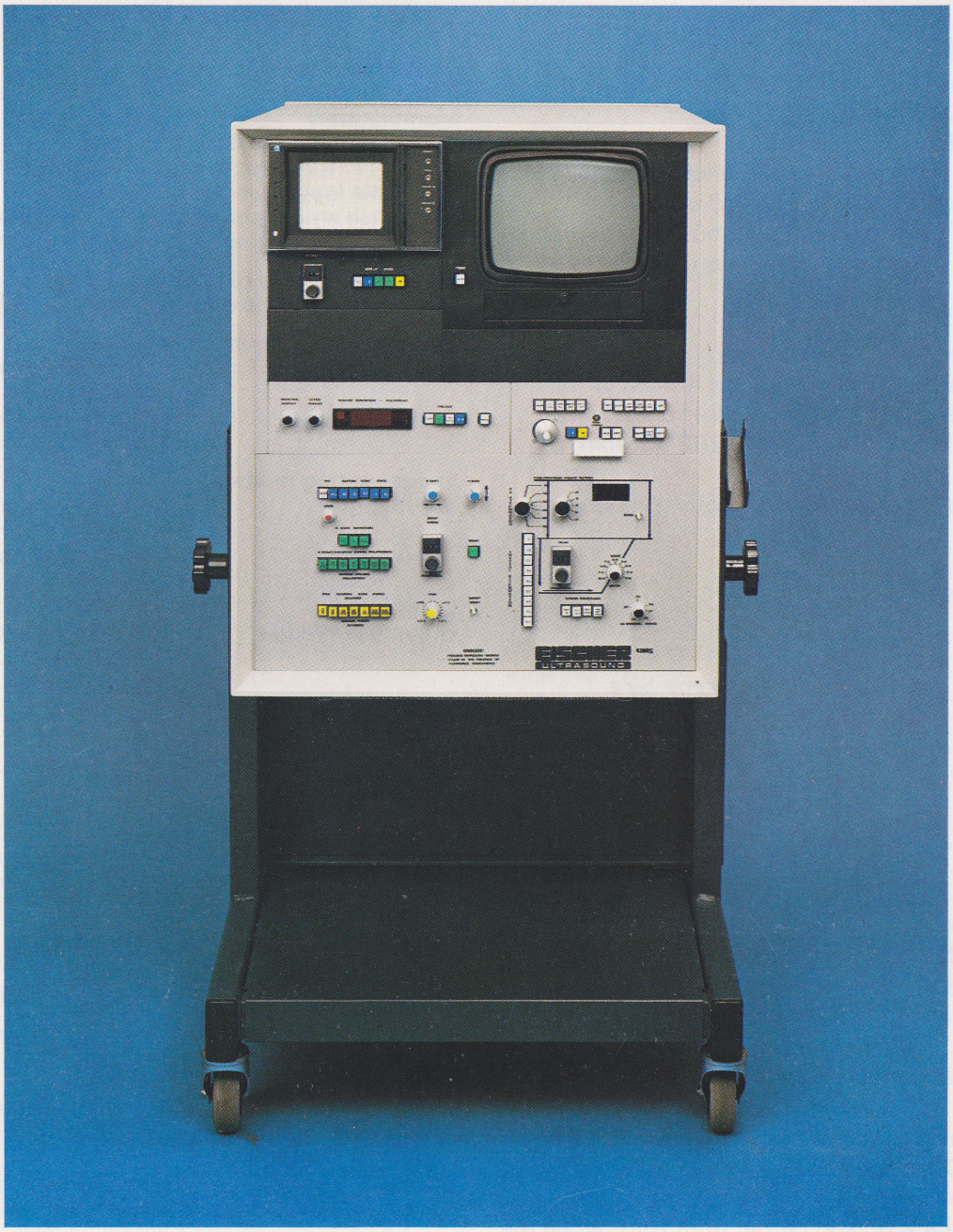
Fischer Ultrasound Ltd. 4200S Console. High Resolution Imaging Combined with Maximum Operating Convenience

FISCHER ULTRASOUND

4200S CONSOLE

HIGH RESOLUTION IMAGING COMBINED WITH MAXIMUM OPERATING CONVENIENCE



One of the comprehensive range of modular diagnostic ultrasound units designed by Fischer Ultrasound to meet individual clinical requirements as simply and cost effectively as is possible.

Connected with the appropriate probes or scanning arms, A mode, M mode and B mode (real time with the addition of the 4260 module) high resolution images can be displayed in greyscale and manipulated using the Digital Memory.

The 4200S console which has been designed for standing or seated operation is mounted on a wheeled trolley and in addition to incorporating a Digital Memory Matrix of over 1 million bits, features high quality displays and an ultrasonic caliper facility. Careful ergonomic grouping and colour coding of push button controls ensure simplicity of operation.

FISCHER ULTRASOUND LTD.

Bankhead Crossway South, Edinburgh EH11 4EY, Scotland Telephone 031-443 4166 Telex 727045 Fischr G

TECHNICAL DATA

Displays
Hewlett Packard 1333A short persistence display unit.
(White phosphor 1332A optional)
9-inch monochrome TV monitor.

Dynamic Range of Echoes
40 to 22dB continuously variable. Dynamic range of system not less than 36 dB

Operating Frequencies
2.5, 3.5, 5, 7.5, 10MHz and Wide Band.

Operating Controls
Sensitivity controlled by transmitter attenuator.
0 to −50 dB in 10 dB steps
0 to −8 dB in 2 dB steps
Swept gain (TGC time gain compensation) controlled by

Initial attenuation	0 to 80 dB in 10 dB steps
Initial delay	0 to 250mm continuously variable
Slope	1.5 to 10 dB/cm continuously variable

A Mode
Displayed on short persistence display.
Displays available are A Scan, Inverted A Scan or superimposed A Scan and TGC waveform.

A Scan ranges:	5, 10, 20, 50, 100, 200 500mm switched ranges
A Scan delay:	0 to 500mm continuously variable

B Mode
Displayed on both displays.

Scanned Area

Range Selected	*Image Size*	
10cm	10cm (X axis)	8cm (Y axis)
20cm	20cm ,,	16cm ,,
30cm	30cm ,,	24cm ,,
40cm	40cm ,,	32cm ,,

M Mode
Scrolling on both displays.
3 ranges 100, 200 and 500mm selected on console.
7 speeds 0.5, 0.8, 1.0, 1.5, 1.8, 3 and 5 seconds, selected on console.
Marker dots written every 0.5 sec along X axis and every 1cm along Y axis. (Each dot has a black and white area to enable identification of markers for all picture grey levels.)

Transmitter Pulse Repetition Rate
a) Fixed PRF is 800 pulses per second
b) Variable rate 'Velocity'. Proportional to the rate of movement of the scanning probe, from zero (when the probe is stationary) to a maximum of 800 pulses/second.

Ultrasonic Caliper
Caliper separation: 0 to 199.0mm may be displayed independently along vector on short persistence display on A Mode, B Mode or M Mode displays.

System velocity Caliper velocity Caliper separation	May be displayed digitally on a panel meter, selected by a rotational control in cabinet access door on the right hand side of the console.

Omnidirectional caliper with digital readout may be displayed on B Mode displays utilising the Digital Memory. Two moveable reference crosses positioned by joystick control. Marker crosses separated by 1cm marker dots. Digital readout accuracy ±1mm on all ranges.

Digital Memory

Image matrix size:	512 × 512 pixels
Memory size:	1,048,576 bits
Number of grey levels:	16

Tissue Texture Processing—four operator-selectable greyscale input transfer curves.
Greyscale emphasis—four operator-selectable post storage (display) grey level enhancement programmes.

Live Vector: Line of dots separated by 1cm (displayed along direction of ultrasound vector)

Image magnification:

Read Zoom	2 × area to be magnified is selected by a moveable box cursor.
Write Magnify	2 × and 4 × area to be rescanned is selected by a moveable box cursor.

Image writing modes:

Survey	Overwrite in which the most recent pixel data is written into the memory while simultaneously erasing old data.
Compound	Peak writing mode.
Greyscale Wedge	16-level reference greyscale on left hand side of image.

Dimensions
Width: 68cm
Height: 129cm
Depth: 84cm

Power Requirements

Voltage	240 V +5%−10% 100 V +5%−10% 220 V +5%−10% 120 V +5%−10%
Frequency	47 to 63Hz
Power	800 VA

Options for use with 4200S Consoles

8200	Flexible Articulated B Scan Arm
OR	
4200	Rectilinear B Scan Frame
4260	Real Time Sector Scan Facilities
	Photographic and Alphanumeric Data Entry Systems
	Fischer's range of probes developed to suit specific applications

For further information on the above and other Fischer Ultrasound products please contact your local Fischer representative or write directly to:

FISCHER ULTRASOUND

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TELEX: 433-0258 HGFI UI

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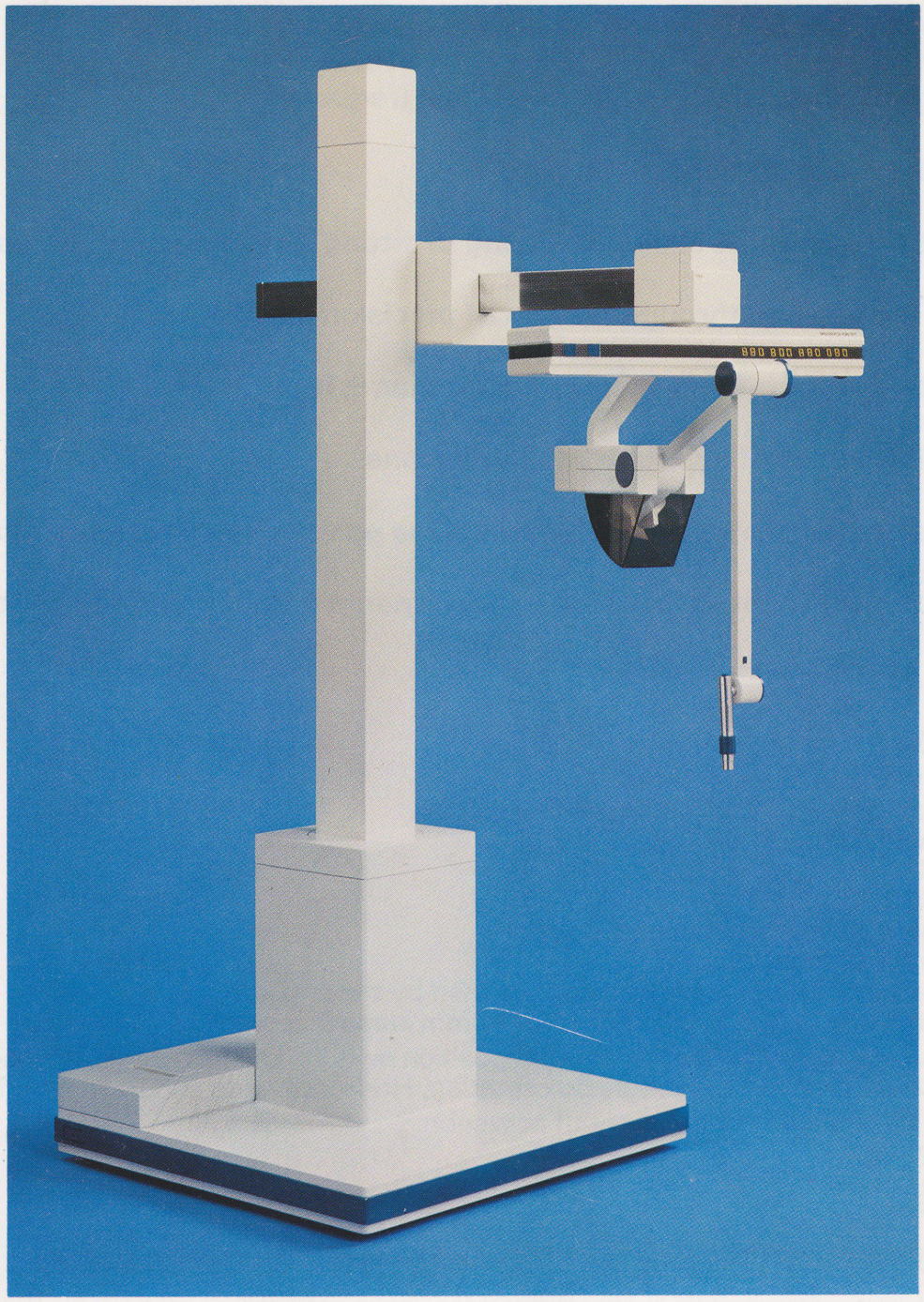
Fischer Ultrasound Ltd. Articulated Scan Arm

FISCHER ULTRASOUND

ARTICULATED SCAN ARM



ACCURATE — EASY TO USE

- Offers smooth, simple and repeatable positioning of the transducer for high resolution scanning.
- Readily adjusted for transverse, sagittal and oblique scanning, improving the efficiency of scanning procedures.
- Ease of movement and finger-touch controls minimise operator fatigue.
- Provides consistently accurate measurement capability.
- Compatible with Fischer B-scanner console range.

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Readily interchangeable frequency encoded bayonet fitting transducers.

Microprocessor control of traverse movements and angular detection.

Display of positional information and operational frequency.

Electrically activated locks secure stability.

Flexible range of movements allows scanning of sitting, standing or prone patients.

Accurate scan registration through the use of precision electronics.

A.I.U.M. phantom registration accuracy better than ±1mm.

Rigidity for maintenance of a precise scanning plane.

Motor drive assistance allows accurate scan plane positioning in selected increments.

Controls are optimally located for range of movements required during a scanning procedure.

SPECIFICATION

Position accuracy of transducer ±1mm.
Absolute accuracy of scan arm position readout ±1mm.
Probe angulation ±170°.
Scan arm tilt ±110°.
Scan arm rotation ±360°.
Traverse movement ±20cm from central position.
Vertical movement available 120cm.
The diameter of a semicircle capable of being scanned by the arm is 70cm.
Preset Automatic Traverse—2, 5, 10, 20, 30 or 40mm steps (accuracy ±0.3mm).
All electrical locks are controlled from a single 3-position switch. The following movements may be activated—Column rotation, Traverse Box rotation, Horizontal slide and Scan arm tilt.
Transducers can be changed with or without power on.

Height	234cm	
Base width	100cm	
Base depth	95cm	
Electrical requirements	Voltage	240V±10% 120V±10% 220V±10% 110V±10%
	Frequency	40 to 63Hz
	Power	200 VA maximum

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Fischer Ultrasound Ltd. MARTI

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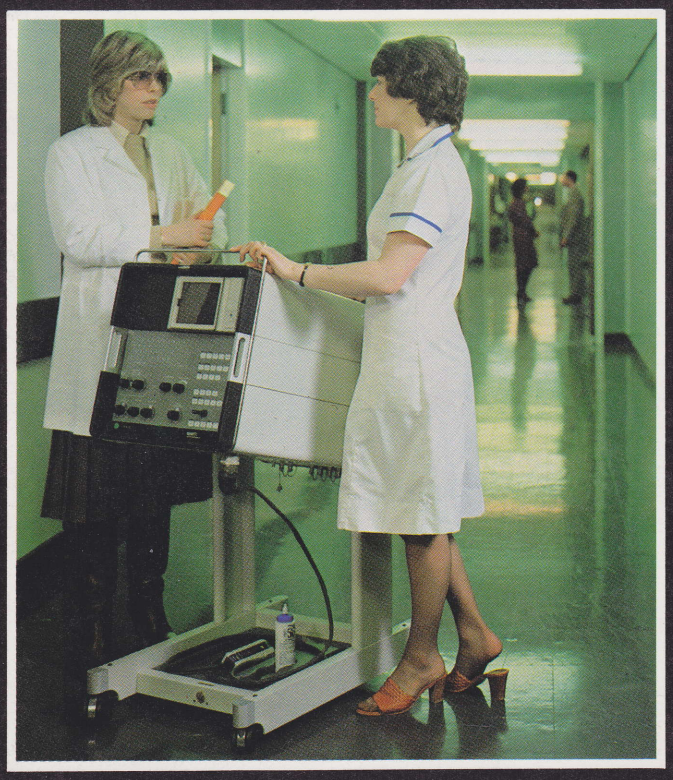


MARTI

FISCHER ULTRASOUND

MARTI

MARTI is a compact and transportable real time sector scanner. Advanced digital scan conversion techniques combined with rotating transducer technology provide a solution to real time imaging problems in Radiology, Obstetrics and Internal Medicine.

MARTI provides A-mode, B-mode real time sector scan and M-mode facilities in a single unit. A variable image frame rate is available for optimum visualisation of rapidly or slowly moving structures. The high resolution real time images are displayed with 64 greylevels and may be frozen without loss of resolution for detailed viewing and recording. Pre and Post storage image processing makes it possible to improve delineation of subtle tissue differences.

Complete patient identification, dual electronic calipers, automatic Area, Perimeter, BPD and CRL computation are standard. All measured data appears alphanumerically on the monitor screen. Permanent recording may be made on Polaroid film, multiformat imagers or video tape recorders.

"Freeze" Of Images
Two methods of obtaining a static image are available to allow detailed study or permanent recordings. "Frame freeze" provides instantaneous freezing of real time images. "Frame grab" allows frozen images to be obtained at low rotational speeds which ensures the highest possible resolution of the stored image.

Advanced Digital Scan Converter
The large capacity digital memory allows greyscale B-mode sector scan images with 64 greylevels to be stored and displayed.

TV Compatibility
Outputs are available for direct interfacing to remote monitors, multiformat imagers and video recorders.

Pre and Post Data Storage Processing

The greyscale emphasis of the image may be allocated both prior to obtaining a stored scan and after the image has been stored. Four different functions are available for both pre and post processing.

Automatic TGC

Time gain compensation (TGC) may be manually adjusted in a conventional manner or the new automatic optimisation feature may be selected.

Scale Change and Zoom

The size of the scanned image may be varied and areas of interest magnified ×2 for detailed examination.

Reverse Polarity Image

A simple rotational control provides black on white or white on black presentation.

Touch Sensitive Keyboard

Provides full alphanumeric patient identification. Time, date and operating frequency are automatically displayed.

Mechanical Rotating Transducer Probe

Four transducers are mounted on a rotating drum housed within the real time probe. The small patient contact area allows access through small acoustic windows eg. intercostal spaces. The probes are vibration free and patient discomfort is minimal.

Electronic Calipers

Two simultaneous measurements of distance may be made using the dual caliper markers. These markers may be moved in all directions over the image (either B-mode sector or M-mode).

Area/Perimeter

In additional to linear measurements, Area and Perimeter measurements may be made. Alphanumeric readout of the measured data appears on the display.

Automatic Display of Gestational Period

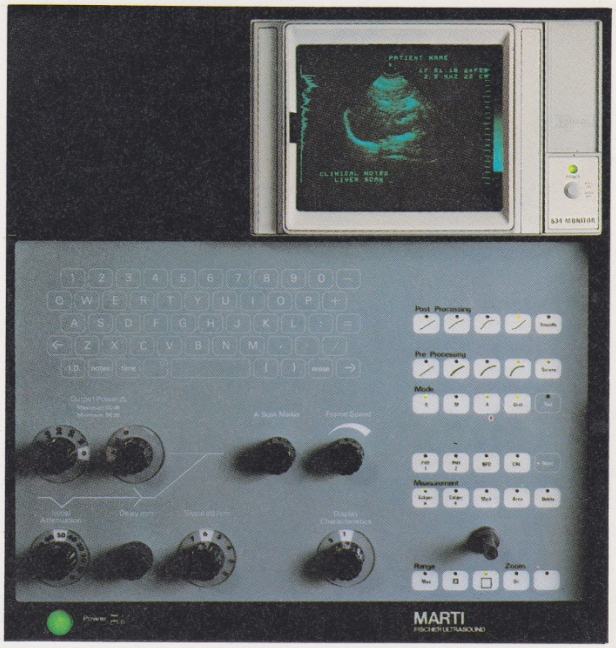
Following either a crown rump length (CRL) or biparietal diameter (BPD) measurement, a display in weeks and days of the gestational period may be obtained.

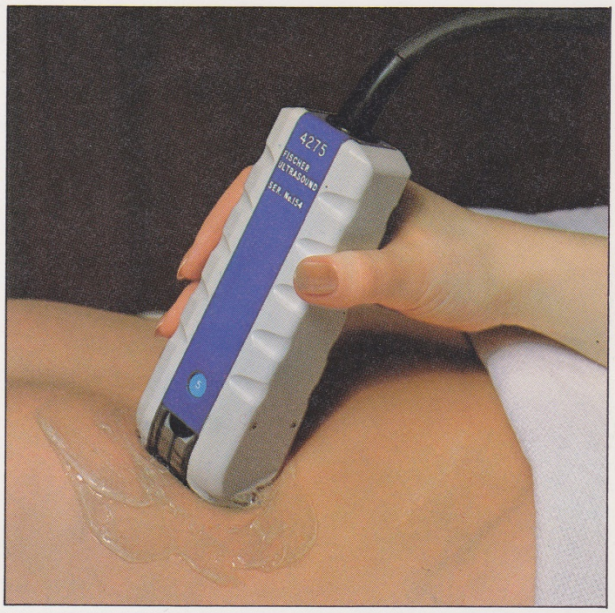
Display

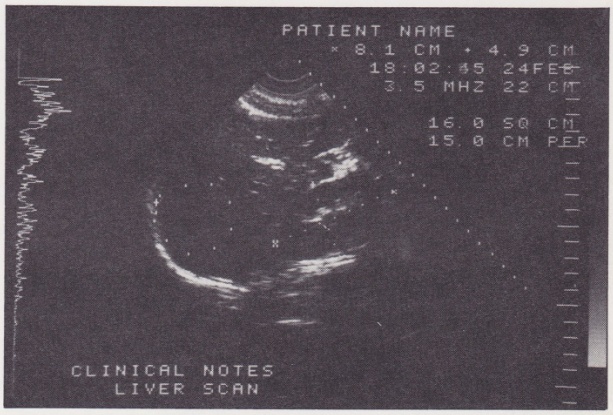
A-mode. When selected a bright marker line appears which can be steered within the 90° sector. The A-mode along this vector is shown on the left hand side of the display.

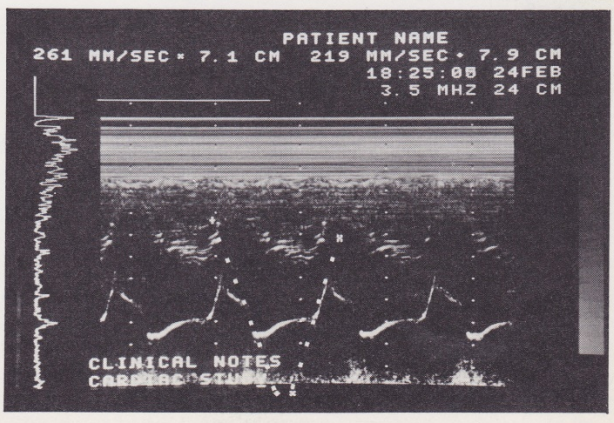
B-mode sector. 90° sector scan images are displayed with 64 greylevels on the high brightness white phospher display.

M-mode. Selection displays an M-mode which scrolls horizontally across the screen. This may be frozen by use of the footswitch. The electronic calipers may be used to provide depth and slope information. In M-mode the rotation of the transducers within the probe is stopped and the probe can be used in the conventional manner to provide M-mode information.









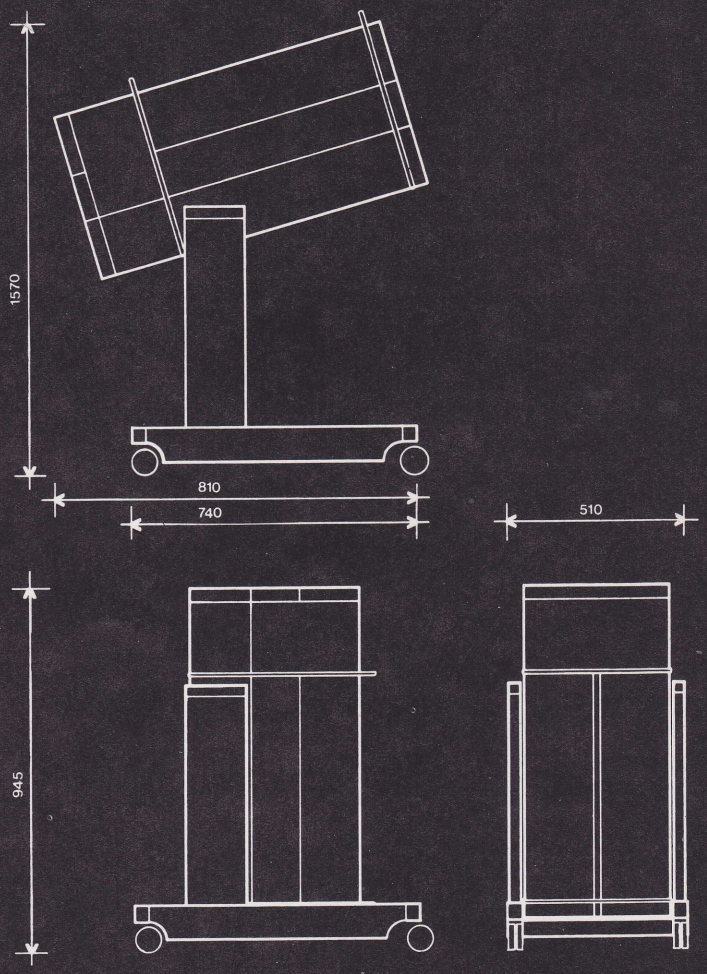
SYSTEM SPECIFICATIONS

Sector Display:	90° sector image, 64 greylevel wedge, A-mode, A-mode marker, range scale marks and alphanumeric annotation on high brightness, white phospher TV raster display.
Rotating Probe:	3.5 MHz (standard) other frequencies available on request.
Operating Modes:	A-mode, B-mode sector or M-mode.
Digital Memory:	480 pixels × 640 pixels × 6 bits, Scan Converter. 64 greylevels displayed on B-mode. M-mode 400 pixels × 640 pixels × 4 bits. 16 greylevels displayed on M-mode.
Frame Rate:	Variable. Maximum 24 frames/sec. Minimum 6 frames/sec.
Maximum Depth of Scan:	24 cm from apex.
Display Ranges:	8cm, 10cm, 15cm, 22cm, 24cm, plus Zoom × 2.
Dynamic Range:	40dB.
Sensitivity Control:	By transmitter attenuator 0 to -50dB in 10dB steps. 0 to -8 dB in 2 dB steps.
Time Gain Compensation:	Initial gain 0 to 80 dB in 10 dB steps. Initial delay 0 to 250mm continuously variable.
Pre-processing:	4 selectable modes.
Post-processing:	4 selectable modes.
Omnidirectional Electronic Calipers:	Distance — between any two points on the image in B and M-modes. Dual calipers allow two independent measurements of distance. Readout in cms.
M-mode Markers:	Markers at ½ sec apart in time and 1 cm apart in depth.
Alphanumeric Keyboard:	Touch sensitive keyboard allows entry of: — 1 line of 20 characters for patient identification. 1 line of 18 characters plus 1 line of 16 characters for patient notes.
Recording:	Polaroid photography standard. Options: — Multiformat camera, video tape recorder.

Computation Facilities:		
	Area	Automatic readout in sq. cms.
	Perimeter	Automatic readout in cms.
	Slope: (for each caliper selected)	Automatic readout on M-mode in mm/sec
	Depth: (for each caliper selected)	Automatic readout on M-mode in cm.
	Crown Rump Length (CRL)	Display of gestation period in weeks and days.
	Biparietal Diameter (BPD)	Display of gestation period in weeks and days.

Mains Voltage:	240V +5% -10% 120V +5% -10% 220V +5% -10% 110V +5% -10%
Power:	400VA
Net Weight:	90kg

DIMENSIONS Units mm



Fischer Ultrasound is the Fischer Imaging Corporation company specialising in Diagnostic Ultrasound.

An extensive range of Ultrasound scanners have been developed and assembled at our Edinburgh, Scotland headquarters by experts in this technology.

If needed, rapid service assistance can be provided by fully trained engineers to ensure the optimum performance of your system.

For further information on MARTI and other Fischer Ultrasound products please contact your local Fischer respresentative or write directly to Fischer Ultrasound.

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Fischer Ultrasound Ltd. LINUS

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LINUS

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LINUS

The LINUS real time system is a compact, portable linear array from Fischer Ultrasound. In the Fischer tradition Linus's image quality maintains the highest standards. Simplicity of operation, portability, reliability are all featured in an economical package.

Advanced ultrasound technology provides high quality real time images by means of electronic and lens focusing. The lightweight transducer is easy to position and an image reversal facility helps to maintain correct anatomical orientation.

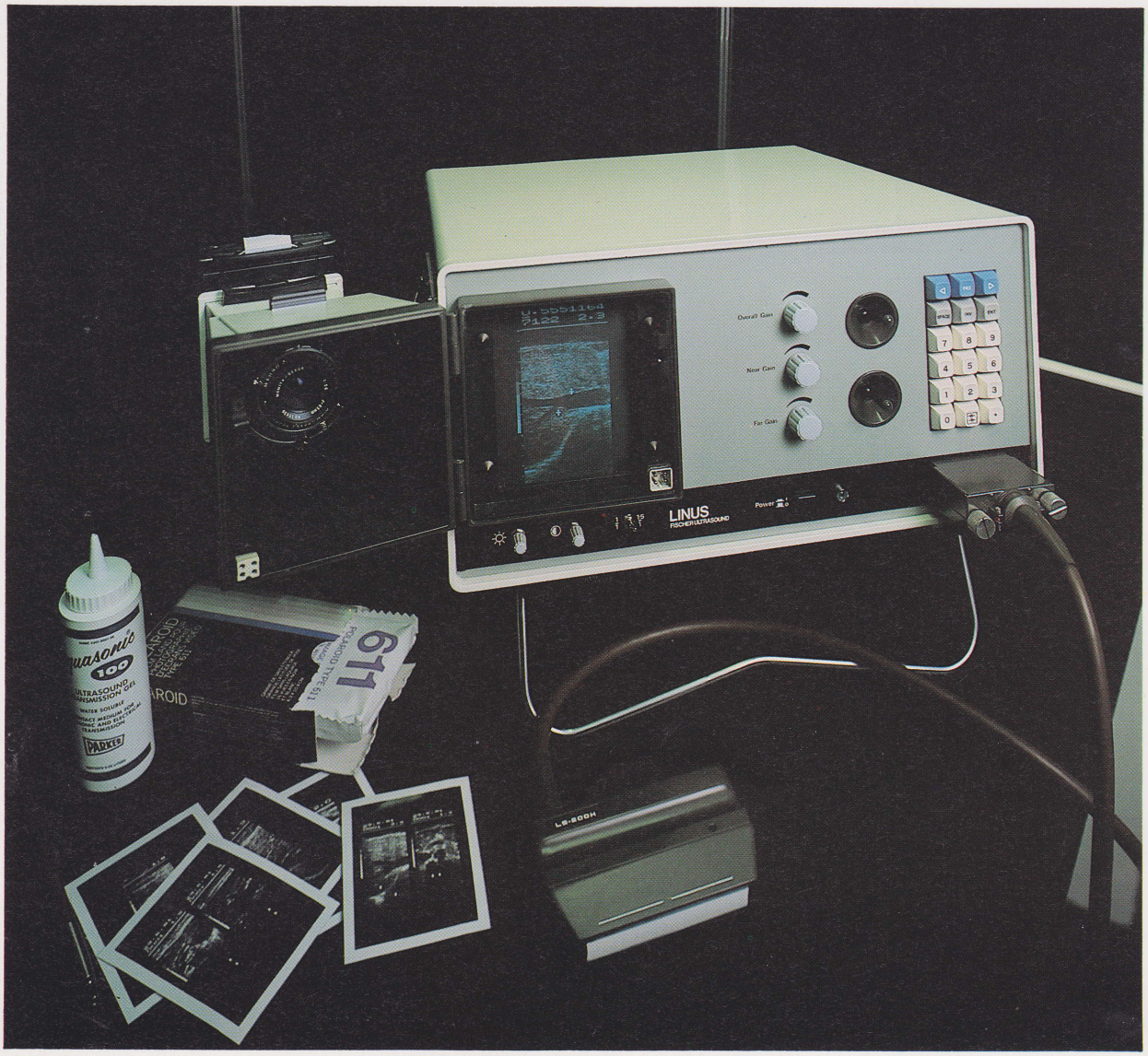
The freeze frame facility enables dynamic images to be frozen for photography. One ¾ scale or two ½ scale images may be photographed with the Polaroid camera. Brightness and focus are automatically adjusted. A greyscale wedge is displayed on the CRT.

Numeric keys are available for patient identification and date. Omnidirectional calipers with digital readout allow accurate measurements to be made.

Simple three control Time Gain Compensation (TGC) and adjustable scale size of ½ or ¾, allows the operator to adjust the field of view to suit the examination.

Optional transducers, camera and display systems are available for expanded capability.

LINUS has the in-built flexibility to guarantee a system with continuing viability for further applications.



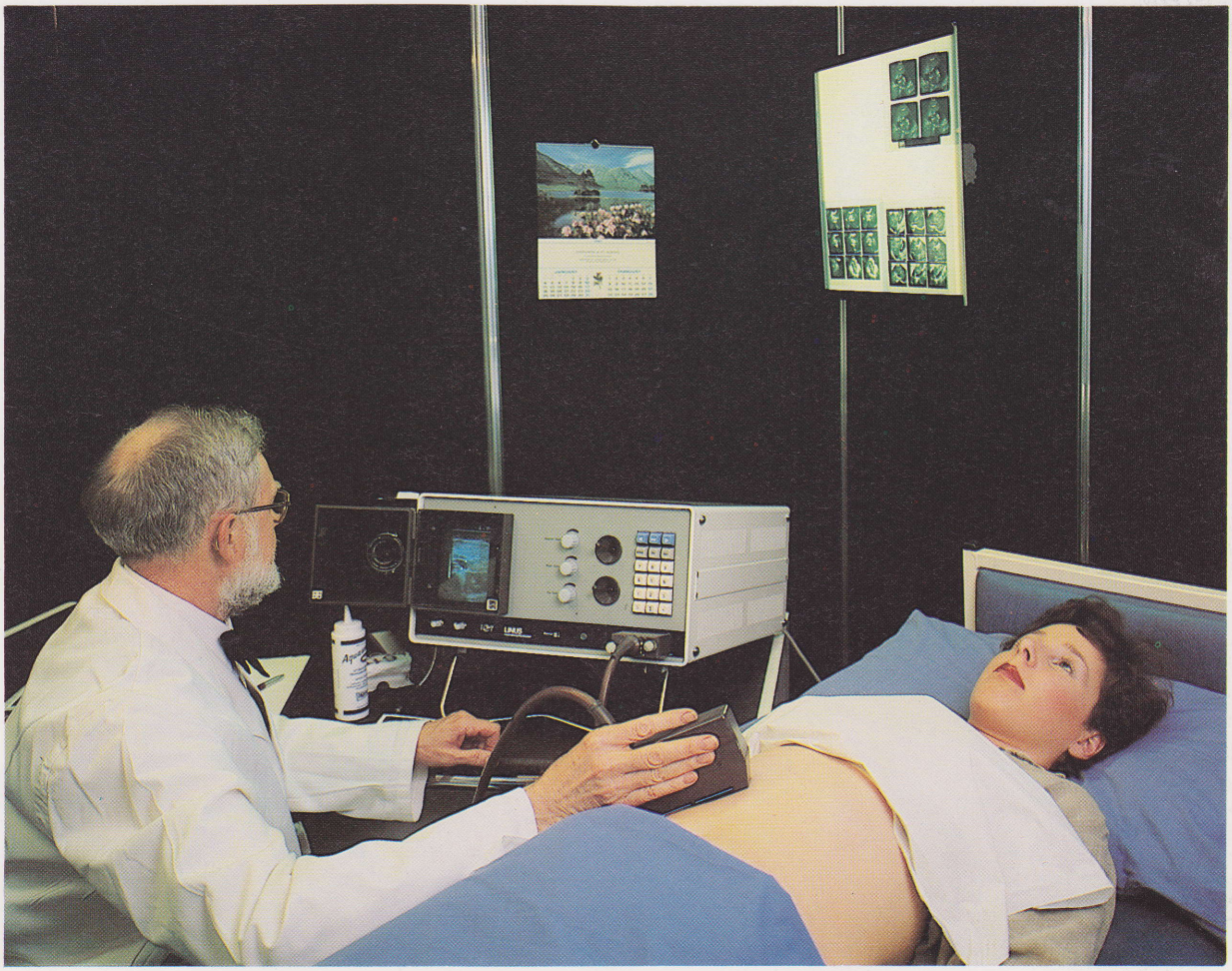
LINUS – A COMPACT PORTABLE UNIT

Clear, simple control unit with CRT display

The lightweight transducer

High quality real time images may be frozen for photography

FISCHER ULTRASOUND



LINUS – HIGH PERFORMANCE YOU CAN DEPEND ON

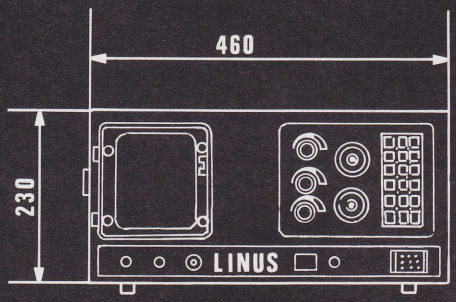
SYSTEM SPECIFICATIONS

General Scanning method:	Linear electronic scanning with electronic focusing and lens focusing
Frequency:	2.25, 3.5, 5 MHz
Focal length:	50 mm approx.
Field of View:	85 mm wide, 20 mm deep (with standard 3.5 MHz Transducer)
Dynamic Range:	40 dB
Number of scanning lines:	114
Frame Rate:	24 per second
TGC:	Near Gain 0 to –50 dB Far Gain 0 to 30 dB Overall Gain 0 to ± 40 dB
Grey Scale:	8 Shades of Grey, Grey Bar Display
Caliper:	Dual joystick control 3 Integer display including one decimal unit Accuracy of ±0.1 cm, ±1 mm
Annotation:	13 digit display numerals 0 to 9, decimal point and space.
Power:	Input power – 100V, 117V, 220V, 240V, AC±10%; 200V A. Input frequency: 50 or 60 Hz
Environment:	Ambient operating temperature 15°C to 35°C
Weight:	Transducer 330 gm, Main Unit 25 Kg.

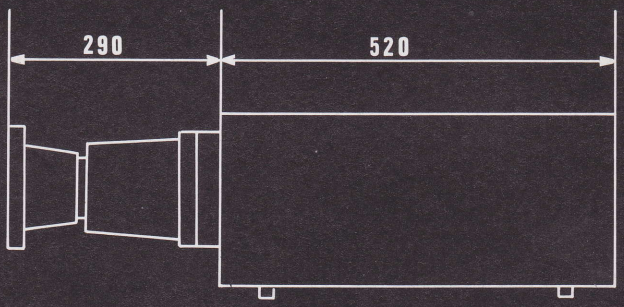
STANDARD CONFIGURATION

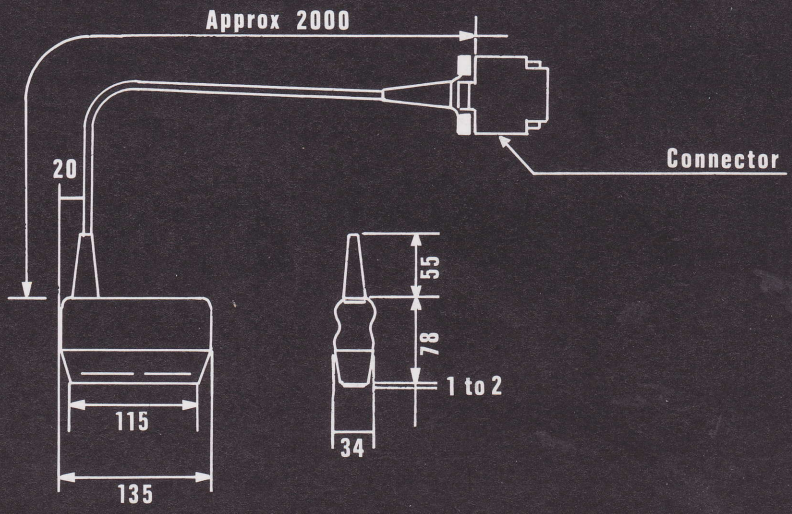
Main Unit incorporating 8 inch display, 3.5 MHz transducer and Polaroid camera.

DIMENSIONS Units mm



Main Unit





Standard 3.5 MHz Transducer

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Locations of archived Diasonographs in the UK

Date, approx	Location	Equipment	Origin	Manufacturer
c. 1957	Hunterian Museum Glasgow	The first contact scanner	Queen Mother's Hospital Glasgow	Operational prototype
c. 1960	Glasgow Museums Resource Centre	The first automatic contact scanner	Queen Mother's Hospital Glasgow	Operational prototype
c. 1965	Science Museum store	Diasonograph	Queen Charlotte's Hospital London	Nuclear Enterprises Ltd
c. 1970	Glasgow Museums Resource Centre	Diasonograph	Queen Mother's Hospital Glasgow	Nuclear Enterprises Ltd
c. 1972	Glasgow Museums Resource Centre	NE 4102	Queen Mother's Hospital Glasgow	Nuclear Enterprises Ltd
c. 1980	National Museum of Scotland Edinburgh	NE4200	Newcastle General Hospital	Nuclear Enterprises Ltd
c. 1986	Science Museum store	NE4200S	Hillingdon Hospital, London	Fischer Ultrasound
c. 1989	Science Museum display	Emisonic 4264 "spinner"	Queen Charlotte's Hospital	Nuclear Enterprises/EMI

List compiled by Francis Duck

INFORMATION FOR AUTHORS

PUBLICATION OF DOCTORAL THESIS AND DISSERTATION ABSTRACTS

A special feature of Medical Physics International (on line at www.mpijournal.org) is the publication of thesis and dissertation abstracts for recent graduates, specifically those receiving doctoral degrees in medical physics or closely related fields in 2010 or later. This is an opportunity for recent graduates to inform the global medical physics community about their research and special interests.

Abstracts should be submitted by the author along with a letter/message requesting and giving permission for publication, stating the field of study, the degree that was received, and the date of graduation. The abstracts must be in English and no longer than 2 pages (using the MPI manuscript template) and can include color images and illustrations. The abstract document should contain the thesis title, author's name, and the institution granting the degree.

Complete information on manuscript preparation is available in the INSTRUCTIONS FOR AUTHORS section of the online journal: www.mpijournal.org.

For publication in the next edition abstracts must be submitted not later than Април 1, 2019.

INSTRUCTIONS FOR AUTHORS

The goal of the new IOMP Journal Medical Physics International (http://mpijournal.org) is to publish manuscripts that will enhance medical physics education and professional development on a global basis. There is a special emphasis on general review articles, reports on specific educational methods, programs, and resources. In general, this will be limited to resources that are available at no cost to medical physicists and related professionals in all countries of the world. Information on commercial educational products and services can be published as paid advertisements. Research reports are not published unless the subject is educational methodology or activities relating to professional development. High-quality review articles that are comprehensive and describe significant developments in medical physics and related technology are encouraged. These will become part of a series providing a record of the history and heritage of the medical physics profession.

A special feature of the IOMP MPI Journal will be the publication of thesis and dissertation abstracts for will be the publication of thesis and dissertation abstracts for recent doctoral graduates, specifically those receiving their doctoral degrees in medical physics (or closely related fields) in 2010 or later.

MANUSCRIPT STYLE

Manuscripts shall be in English and submitted in WORD. Either American or British spelling can be used but it must be the same throughout the manuscript. Authors for whom English is not their first language are encouraged to have their manuscripts edited and checked for appropriate grammar and spelling. Manuscripts can be up to 10 journal pages (approximately 8000 words reduced by the space occupied by tables and illustrations) and should include an unstructured abstract of no more than 100 words.

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Illustrations can be inserted into the manuscript for the review process but must be submitted as individual files when a manuscript is accepted for publication.

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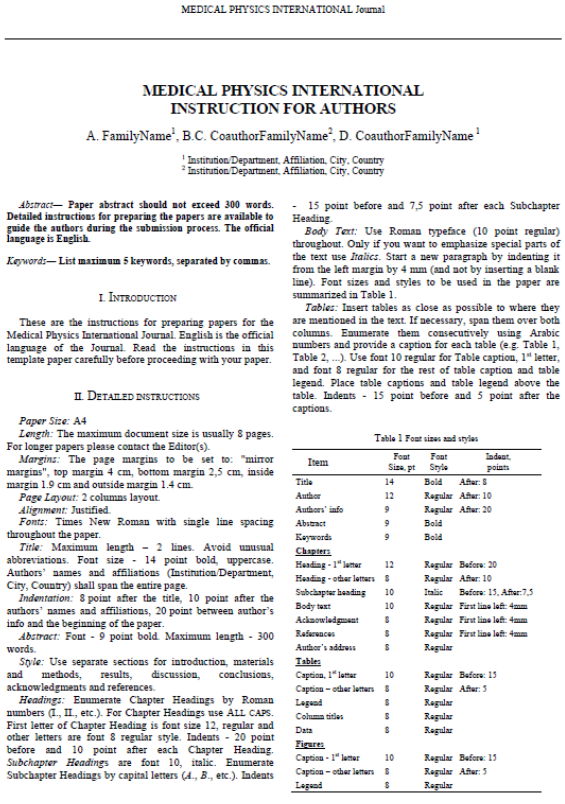
Authors considering the development of a manuscript for a Review Article can first submit a brief proposal to the editors. This should include the title, list of authors, an abstract, and other supporting information that is appropriate. After review of the proposal the editors will consider issuing an invitation for a manuscript. When the manuscript is received it will go through the usual peer-review process.



MEDICAL PHYSICS INTERNATIONAL Journal

MEDICAL PHYSICS INTERNATIONAL INSTRUCTION FOR AUTHORS

A. FamilyName[1], B.C. CoauthorFamilyName[2], D. CoauthorFamilyName [1]

[1] Institution/Department, Affiliation, City, Country
[2] Institution/Department, Affiliation, City, Country

Abstract— **Paper abstract should not exceed 300 words. Detailed instructions for preparing the papers are available to guide the authors during the submission process. The official language is English.**

Keywords— **List maximum 5 keywords, separated by commas.**

I. INTRODUCTION

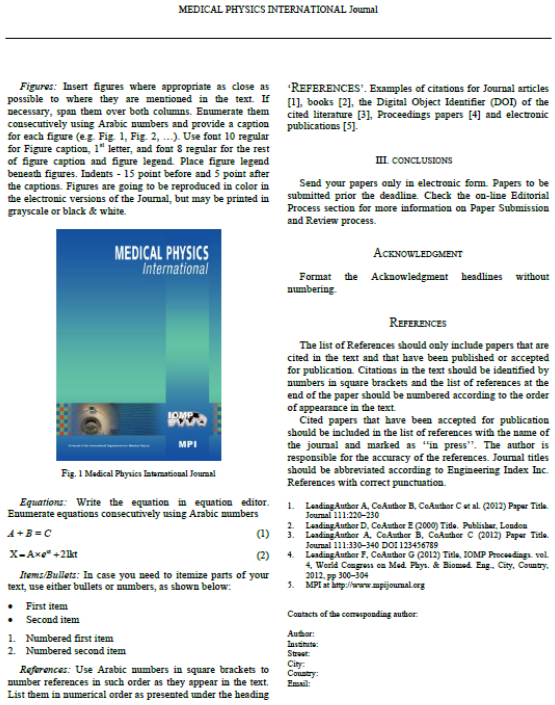
These are the instructions for preparing papers for the Medical Physics International Journal. English is the official language of the Journal. Read the instructions in this template paper carefully before proceeding with your paper.

II. DETAILED INSTRUCTIONS

Paper Size: A4
Length: The maximum document size is usually 8 pages. For longer papers please contact the Editor(s).
Margins: The page margins to be set to: "mirror margins", top margin 4 cm, bottom margin 2,5 cm, inside margin 1.9 cm and outside margin 1.4 cm.
Page Layout: 2 columns layout.
Alignment: Justified.
Fonts: Times New Roman with single line spacing throughout the paper.
Title: Maximum length – 2 lines. Avoid unusual abbreviations. Font size - 14 point bold, uppercase. Authors' names and affiliations (Institution/Department, City, Country) shall span the entire page.
Indentation: 8 point after the title, 10 point after the authors' names and affiliations, 20 point between author's info and the beginning of the paper.
Abstract: Font - 9 point bold. Maximum length - 300 words.
Style: Use separate sections for introduction, materials and methods, results, discussion, conclusions, acknowledgments and references.
Headings: Enumerate Chapter Headings by Roman numbers (I., II., etc.). For Chapter Headings use ALL CAPS. First letter of Chapter Heading is font size 12, regular and other letters are font 8 regular style. Indents - 20 point before and 10 point after each Chapter Heading. *Subchapter Headings* are font 10, italic. Enumerate Subchapter Headings by capital letters (*A.*, *B.*, etc.). Indents - 15 point before and 7,5 point after each Subchapter Heading.
Body Text: Use Roman typeface (10 point regular) throughout. Only if you want to emphasize special parts of the text use *Italics*. Start a new paragraph by indenting it from the left margin by 4 mm (and not by inserting a blank line). Font sizes and styles to be used in the paper are summarized in Table 1.
Tables: Insert tables as close as possible to where they are mentioned in the text. If necessary, span them over both columns. Enumerate them consecutively using Arabic numbers and provide a caption for each table (e.g. Table 1, Table 2, ...). Use font 10 regular for Table caption, 1st letter, and font 8 regular for the rest of table caption and table legend. Place table captions and table legend above the table. Indents - 15 point before and 5 point after the captions.

Table 1 Font sizes and styles

Item	Font Size, pt	Font Style	Indent, points
Title	14	Bold	After: 8
Author	12	Regular	After: 10
Authors' info	9	Regular	After: 20
Abstract	9	Bold	
Keywords	9	Bold	
Chapters			
Heading - 1st letter	12	Regular	Before: 20
Heading - other letters	8	Regular	After: 10
Subchapter heading	10	Italic	Before: 15, After: 7.5
Body text	10	Regular	First line left: 4mm
Acknowledgment	8	Regular	First line left: 4mm
References	8	Regular	First line left: 4mm
Author's address	8	Regular	
Tables			
Caption, 1st letter	10	Regular	Before: 15
Caption – other letters	8	Regular	After: 5
Legend	8	Regular	
Column titles	8	Regular	
Data	8	Regular	
Figures			
Caption - 1st letter	10	Regular	Before: 15
Caption – other letters	8	Regular	After: 5
Legend	8	Regular	



MEDICAL PHYSICS INTERNATIONAL Journal

Figures: Insert figures where appropriate as close as possible to where they are mentioned in the text. If necessary, span them over both columns. Enumerate them consecutively using Arabic numbers and provide a caption for each figure (e.g. Fig. 1, Fig. 2, ...). Use font 10 regular for Figure caption, 1st letter, and font 8 regular for the rest of figure caption and figure legend. Place figure legend beneath figures. Indents - 15 point before and 5 point after the captions. Figures are going to be reproduced in color in the electronic versions of the Journal, but may be printed in grayscale or black & white.

Fig. 1 Medical Physics International Journal

Equations: Write the equation in equation editor. Enumerate equations consecutively using Arabic numbers

$$A + B = C \quad (1)$$

$$X = A \times e^{at} + 2lkt \quad (2)$$

Items/Bullets: In case you need to itemize parts of your text, use either bullets or numbers, as shown below:

- First item
- Second item

1. Numbered first item
2. Numbered second item

References: Use Arabic numbers in square brackets to number references in such order as they appear in the text. List them in numerical order as presented under the heading 'REFERENCES'. Examples of citations for Journal articles [1], books [2], the Digital Object Identifier (DOI) of the cited literature [3], Proceedings papers [4] and electronic publications [5].

III. CONCLUSIONS

Send your papers only in electronic form. Papers to be submitted prior the deadline. Check the on-line Editorial Process section for more information on Paper Submission and Review process.

ACKNOWLEDGMENT

Format the Acknowledgment headlines without numbering.

REFERENCES

The list of References should only include papers that are cited in the text and that have been published or accepted for publication. Citations in the text should be identified by numbers in square brackets and the list of references at the end of the paper should be numbered according to the order of appearance in the text.
Cited papers that have been accepted for publication should be included in the list of references with the name of the journal and marked as ''in press''. The author is responsible for the accuracy of the references. Journal titles should be abbreviated according to Engineering Index Inc. References with correct punctuation.

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5. MPI at http://www.mpijournal.org

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