ULTRASOUND

and

INFERTILITY

Asim Kurjak

CRC Press
Taylor & Francis Group

Ultrasound and Infertility

Editor

Asim Kurjak, M.D., Ph.D.
Professor and Chairman
Ultrasonic Institute
Collaborative Center for Diagnostic Ultrasound
World Health Organization
Zagreb, Yugoslavia

CRC Press
Taylor & Francis Group
Boca Raton London New York

CRC Press is an imprint of the
Taylor & Francis Group, an informa business

CRC Press
Taylor & Francis Group
6000 Broken Sound Parkway NW, Suite 300
Boca Raton, FL 33487-2742

First issued in paperback 2019

ISBN 13: 978-0-367-45108-0 (pbk)
ISBN 13: 978-0-8493-4766-5 (hbk)

Visit the Taylor & Francis Web site at
http://www.taylorandfrancis.com

and the CRC Press Web site at
http://www.crcpress.com

Library of Congress Card Number 88-14469

Library of Congress Cataloging-in-Publication Data

Ultrasound and infertility /editor, Asim Kurjak.
p. cm.
Includes bibliographies and index.
ISBN 0-8493-4766-1
1. Infertility — Diagnosis. 2. Generative organs — Ultrasonic imaging. I. Kurjak, Asim.
[DNLM: 1. Infertility — diagnosis. 2. Infertility — therapy. 3. Ultrasonic Diagnosis. 4. Ultrasonic Therapy. WP 570 U47]
RC889.U37 1989
618.1'7207543 — dc19
DNLM/DLC 88-14469
for Library of Congress CIP

PREFACE

It has been more than 30 years since the late Professor Ian Donald first started to use ultrasound in obstetrics and gynecology. In a relatively short period of time, ultrasound has improved in what seems to be a logarithmic progression, and it can, with good reason, be said to have changed the way of thinking of modern obstetricians.

However, until recently, the progress was not so rapid in the field of infertility; but, in the last 10 years, ultrasound has become a highly sophisticated scientific tool readily available for the diagnosis and management of infertile couples. Ultrasound has a permanent advantage over other imaging and diagnostic techniques by being rapid, safe, and noninvasive. Therefore, ultrasound is being used with increasing success to determine the time of ovulation, artificial insemination, or *in vitro* fertilization.

In addition, ultrasound appears to be a safe, practical, and noninvasive alternative to laparoscopy in many aspects. It is, therefore, time to produce a book which will serve as a useful guide for the optimal use of ultrasound in an infertility clinic, containing most of the information necessary for practical work, and bringing ultrasonographers and clinicians up to date information on the current knowledge and special problems for the future.

Asim Kurjak

THE EDITOR

Asim Kurjak, M.D., Ph.D., is chairman of the department and professor of Obstetrics and Gynecology at the University of Zagreb, Yugoslavia. He is also head of the Ultrasonic Institute of Zagreb which is the World Health Organization Collaborative Center for diagnostic ultrasound.

Dr. Kurjak obtained his training at the University of Zagreb, receiving his M.D. degree in 1966 and his Ph.D. degree in 1977. He served as an assistant professor at the Department of Obstetrics and Gynecology, University of Zagreb from 1968 to 1980. In 1971, as a British scholar, he was research assistant for 1 year at the Institute of Obstetrics and Gynecology, University of London. From 1983 to 1985 he was external examiner at the University of Liverpool. It was in 1983 that he assumed his present position.

Dr. Kurjak is a member and past president of the Yugoslav Society of University Professors Academy of Croatia. He is member of the advisory board of four international scientific journals. He served as vice president of the European federation for ultrasound in medicine and biology. He has been the recipient of many research grants from the Scientific Council from Yugoslavia and is currently the WHO coordinator for the use of ultrasound in developing countries. Among other awards, he is an honorary member of Ultrasonic Society of Australia, Italia, Egypt, Indonesia and of the Sociation of Obstetrics and Gynecology of Italia, Poland, and Hungary.

Dr. Kurjak has presented over 70 invited lectures at major international meetings and approximately 100 guest lectures at universities and institutes. He has published more than 200 research papers and 14 books in the English language. His current major research interests include ultrasound diagnosis and fetal and Doppler studies of fetoplacental and maternal blood flow.

CONTRIBUTORS

Marinko Biljan, M.D.
Ultrasonic Institute
Medical Faculty
University of Zagreb
Zagreb, Yugoslavia

Menashe Ben-David, Ph.D.
Department of Obstetrics & Gynecology
Hadassah University Hospital
Jerusalem, Israel

Branko Breyer, Ph.D.
Head
Medical Physics Department
Gynecological Cancer Center
University Gynecological Hospital
Zagreb, Yugoslavia

Zeljko Fuchkar, M.D., Ph.D.
Associate Professor
Urology and Kidney Transplants
Clinical Hospital Center
Faculty of Medicine
University "Vladimir Bakaric"
Rijeka, Yugoslavia

Davor Jurkovic, M.D.
Research Fellow
Ultrasonic Institute
Medical Faculty
University of Zagreb
Zagreb, Yugoslavia

Asim Kurjak, M.D., Ph.D.
Professor and Chairman
Ultrasonic Institute
Collaborative Center for Diagnostic Ultrasound
World Health Organization
Zagreb, Yugoslavia

Aby Lewin, M.D.
Department of Obstetrics & Gynecology
Hadassah University Hospital
Jerusalem, Israel

Shraga Rottem, M.D.
Department of Obstetrics & Gynecology
Rambam Medical Center
Medical School
Technion-Institute of Technology
Haifa, Israel

Joseph G. Schenker, M.D.
Professor and Chairman
Department of Obstetrics & Gynecology
Hadassah University Hospital
Jerusalem, Israel

Ilan E. Timor-Tritsch, M.D.
Professor and Director
Department of Obstetrics & Gynecology
College of Physicians & Surgeons
Columbia University
New York, New York

TABLE OF CONTENTS

Chapter 1

BASIC PHYSICS OF ULTRASOUND

Branko Breyer

TABLE OF CONTENTS

I. INTRODUCTION

In this chapter, we describe the basic physical and technological principles of ultrasound diagnostics without mathematical treatment, except for some simple formulas, yet include comments relevant for practical use. Ultrasound diagnostic instruments and procedures are still in fast development, so that mere knowledge of manipulation with the existing instruments is definitely insufficient for sound usage of the existing instruments to come in a few years. The knowledge of underlying principles allows one to understand what is actually new in an instrument, and what are the supposed advantages.

II. PHYSICAL PRINCIPLES

A. Ultrasound Waves

Ultrasound is, per definition, the sound of a frequency higher than the hearing limit of the human ear, i.e., above 16 to 20 kHz. Bat's definition of ultrasound would be different. In medical diagnostics, one normally uses ultrasound waves of frequencies between 2 and 10 MHz. Basic physical principles are equally valid for audible sound as for ultrasound, only at different scales. Ultrasound is a mechanical wave, i.e., it consists of mechanical vibrations of medium particles through which it propagates. In soft tissues, the medium particles vibrate along the direction of wave propagation creating their densifications and rarefactions in space. Such a wave is called a longitudinal wave. The particles (molecules) oscillate around their (stochastic) balance positions with no net flow of matter, however, the energy flows. At very high energy densities some net flow can be induced, but this does not apply to energies of ultrasound used in diagnostics. Other types of waves like transversal and Raileigh cannot propagate to any appreciable distance in soft tissues. Ultrasound waves are characterized by parameters like frequency, wavelength, propagation speed, intensity, and pressure. Frequency is expressed in hertz (Hz), i.e., cycles per second. The physical dimension is 1/s; 1 Hz = 1 c/s, 1 kHz = 1000 c/s, and 1 MHz = 1 million c/s. The frequency used in diagnostics largely influences their properties. Wavelength is the distance between the same phases of compression of the medium in two consecutive cycles in space and is measured in meters or its subunits like millimeters. The propagation speed depends mainly on the media (tissue) properties through which the wave propagates and is related with frequency and wavelength as follows:

$$\lambda = \frac{c}{f}$$

where c = propagation velocity in meters per second (m/s), f = frequency in hertz (Hz), and λ = wavelength in meters (m).

Strictly speaking, sounds of different frequencies travel at different velocities (frequency velocity dispersion), but these effects are negligible at frequencies, intensities, and circumstances of medical echography. The average propagation speed in soft tissues is 1540 m/s. The speed depends on both density and elastic properties of tissues.

When traveling through tissues, ultrasound causes a variation of the total pressure, i.e., the sonic pressure oscillations are superimposed upon the static pressure (atmospheric pressure). The measure of energy density flowing through a unity area in one unit of time is called intensity and is expressed in watts per square meter or centimeter (W/cm^2 or W/m^2).

When speaking about ultrasound intensity in an ultrasound beam, it is not sufficient to simply state the intensity, one must as well state "where" and "when"; i.e., at what position in space and within which interval. Simply stating the intensity applies only to a plane wave in a nonattenuating medium. In echography, one uses pulsed ultrasound and focused beams

Table 1
ORIENTATIONAL VALUES OF PROPAGATION SPEEDS IN SOFT TISSUES[1,2]

Tissue	Speed (m/s)
Muscle	1570
Kidney	1560
Liver	1560
Brain	1510
Fat	1440

Note: The speed values are for orientation only. Inspection of the available literature shows fairly divergent results. There is a consensus that the average speed for soft tissues should be taken as 1540 m/s.

in attenuating media. After we discuss characteristics of ultrasound beams, we shall turn back to definitions of ultrasound intensities in the beam.

Likewise, when speaking about frequency, one can define a single frequency only as a continuous wave. When speaking about pulses of some frequency, we usually think of frequency obtained by counting the number of zero-level crossings during the pulse. However, a pulse by physical laws contains a mixture of frequencies which we call the frequency spectrum of the pulse, and which is broader — the shorter the pulse. This fact must be born in mind in the practical use of ultrasound echography because, as we shall show, one uses different frequencies for different purposes, so it may sometimes be useful to bear in mind that when using, e.g., a probe declared 3.5 MHz central frequency, one actually transmits and receives a spectrum of frequencies, which may range from below 1 MHz to more than 5 MHz.

B. Ultrasound Propagation in Tissues

Human tissues are not homogenous in respect to ultrasound wave propagation, so that in transversing the tissues, ultrasound waves are being reflected, refracted, scattered, and absorbed. Angle of refraction depends on the respective velocities in the tissues.

The amount of reflected energy depends on characteristic acoustic impedances and angle of incidence upon the boundary of tissues. The tissues we are speaking about are different tissues in terms of elasticity and other physical properties, but can be parts of the same biological tissue. Characteristic acoustic impedance is a property of a medium and is defined as the ratio of the momentary acoustic pressure and particle velocity induced by this pressure.

In different tissues, ultrasound velocities and impedance are generally different (see Table 1). Differences between different soft tissues are much smaller than between soft tissues and bones, respectively, gas. This fact has important consequences in the use of ultrasound in medicine.

1. Refraction and Reflection of Ultrasound Waves

In discussing reflection and refraction of ultrasound, we shall take the simplest case — the plane wave — which, although an idealization, will give us a good idea of what happens when ultrasound waves transverse a boundary of media which is much larger than the wavelength.

As illustrated in Figure 1, a part of the energy is reflected and some of the energy is

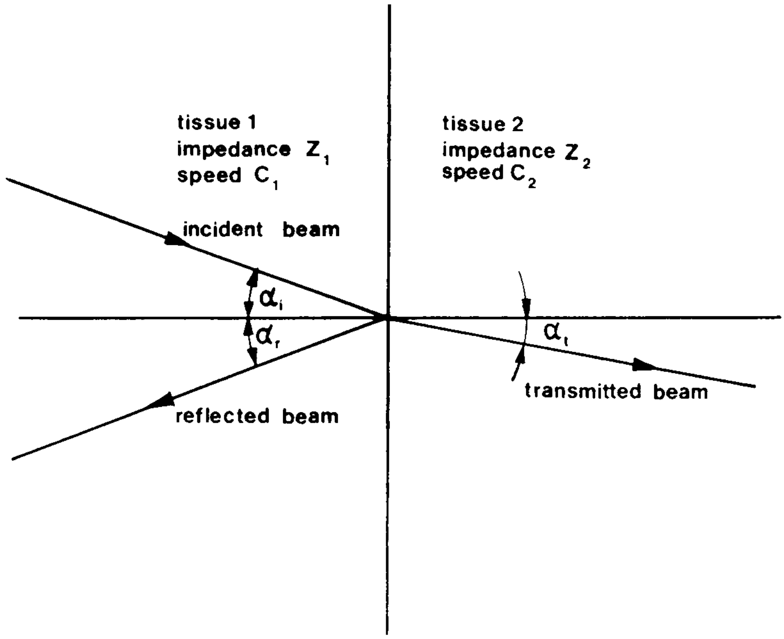

FIGURE 1. Reflection and refraction of ultrasound waves on a media boundary. The two media have different characteristic impedances (Z_1 and Z_2) and different propagation speeds (c_1 and c_2). The boundary dimensions are larger than the wavelength.

transmitted across the boundary. The angle of refraction depends on the ratio of propagation speeds in the two respective media, as described by Equation 1, while the refraction of energy reflected depends on the difference of characteristic acoustic impedances, as well as on the angle of incidence. (Equation 2a). If the speeds of ultrasound in the first and the second medium are c_1 and c_2, respectively, and the incidence angle is α_i and the transmission angle is α_t, then the refraction angle can be calculated from Equation 1:

$$\frac{\sin\alpha_i}{\sin\alpha_t} = \frac{c_1}{c_2} \tag{1}$$

Equation 2 describes the ratio of the amplitudes of the reflected and the incident wave:

$$\frac{A_r}{A_i} = \frac{Z_2 \cos\alpha_i - Z_1 \cos\alpha_t}{Z_2 \cos\alpha_i + Z_1 \cos\alpha_t} \tag{2a}$$

If the incidence is perpendicular to the media boundary, Equation 2a degenerates to

$$\frac{A_r}{A_i} = \frac{Z_2 - Z_1}{Z_2 + Z_1} \tag{2b}$$

In many practical cases when we consider ultrasound waves, which are reflected back to the transmitting probe, Equation 2b gives a fairly accurate picture of what goes on.

Intensity of ultrasound is proportional to the square of its pressure so that the ratio of the reflected to the incident intensity is at 90°

$$\frac{I_r}{I_i} = \frac{(Z_2 - Z_1)^2}{(Z_2 + Z_1)^2} \tag{3}$$

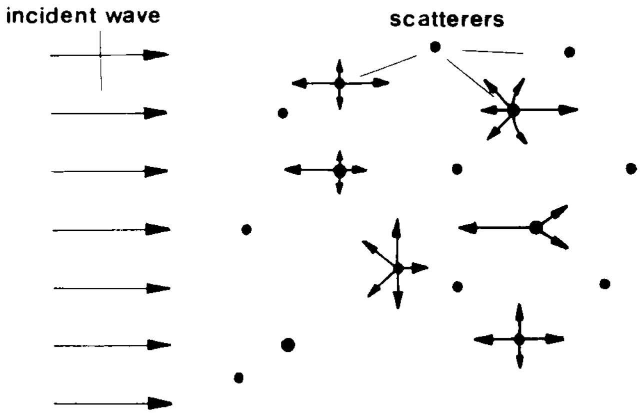

FIGURE 2. The incident waves encounter scatterers stochastically dispersed in the medium. The incident waves induce oscillations of the scatterers, which retransmit wavelets in varied directions with different intensities (illustrated by arrow lengths). The waves that are backscattered add up and interfere, and as such, are detected by the receiving transducer.

If the characteristic acoustic impedances are very different like in the case of soft tissue and gas (e.g., in lungs), from the above equation one can see that virtually all ultrasound energy will be reflected. A consequence of this is that organs containing gas, such as the lungs, cannot be examined by ultrasound, and gas in the stomach and intestines will present serious problems. Another consequence is that some sort of contact gel, oil, or other coupling agent should be applied between the ultrasound transmitter and the body to avoid air bubbles being trapped between the probe and the skin and to allow ultrasound to enter the body. It is interesting to see that ultrasound will be reflected irrespective of whether it enters from a lower or a higher impedance. The absolute value of reflected wave is the same for the same difference in characteristic acoustic impedances. On the other hand, the sign changes indicate a difference in phase.

The energy that is not reflected will be transmitted across the impedance boundary. Thus, in many cases it is the characteristic acoustic impedance that is decisive for the intensity of an echo. Sometimes the acoustic impedance is mixed up with the density by a vague analogy with X-ray imaging. This is wrong because the characteristic impedance depends on the density and propagation speed (c):

$$Z = \rho \cdot c$$

and not only on the density.

So far we have considered the situation illustrated in Figure 1 where the impedance boundary is continuous and much larger than the wavelength. Such a reflector is called specular, i.e., mirror-like. There are many structures in the body which act as specular reflectors, but there are even more structures in which the reflectors are of dimensions similar or smaller than the wavelength. Such small reflectors scatter ultrasound in different directions similar to what happens when light is shined through a sheet of paper. Figure 2 illustrates what happens with ultrasound waves when transversing tissues such as the liver and muscles and many other tissues in the body.

The wavelets from scatterers are retransmitted in all directions and add up in different, only stochastically predictable ways. They interfere among themselves, and the resulting waves that return to the transmitting probe do not exactly represent each of the scatterers. This has important consequences on our interpretation of echographic images (Table 1).

As one can see from Table 1 the speed of ultrasound in different tissues varies, which introduces certain errors in distance measurements. In the majority of cases, this is not very important and one usually settles for an average calibration speed of 1540 m/s for soft tissues. In some instances, this is not good enough and then the problem is circumvented by agreeing upon a standard calibration velocity (e.g., 1540 or 1600 m/s), then measuring the normal dimension of interest using this calibration velocity, and finally producing a graph or table containing normal values with the above supposition. Later on, one can compare values obtained by measurement with the normals, and not with the inaccesible, real values. In this way, one can tell whether or not a dimension is normal (bone length, biparietal diameter, etc.), which is what we really need, but one can be by some percentage wrong when comparing the actual, inaccessible dimension, which is not really important.

Absorption of ultrasound in tissues reduces ultrasound intensity as it passes through the tissues. Attenuation is further increased by the scattering of ultrasound waves. Both phenomena are frequency dependent and are expressed more at higher frequencies. Therefore, higher frequency ultrasound beams are more readily attenuated than the lower frequency beams and so, when we need to scan deeper structures, we must apply lower frequency ultrasound. At present, one typically uses approximately 3 MHz for general abdominal scanning in adults (liver, pancreas, advanced pregnancy); and about 5 MHz for children scanning, early pregnancy, neck, and breast; and 7 MHz for shallow scanning. A general rule is that the highest frequency will still satisfy penetration; i.e., the shortest applicable wavelength should be used.

Apart from this, scattering, absorption, and velocity are characteristic of tissues and are potential means of quantitatively characterizing tissues. Many experiments are under way along these lines with a fair probability that they will lead to quantitative ways of describing and diagnosing diffuse lesions of parenchymatous organs and differentiating malignant from benign lesions or at least stating a heavy suspicion of malignancy.

III. ECHOSCOPIC SYSTEMS

Ultrasound waves can be used, for medical diagnostics, in many different ways, but the most common method at present is ultrasonic echography. The principle is very simple. One sends out into the body pulses of ultrasound (about 1 μs long) and the echoes from different reflectors and scatterers in the body are detected by the same probe, which was used for transmission. Knowing the speed of ultrasound and measuring the time necessary for the echoes to return, the distance to the reflectors could be calculated. The direction in which the ultrasound pulses have been transmitted is known too, so that one can determine the position of a reflector in two dimensions. The echoes are usually shown on a screen in B, A, or M mode. A and B modes are shown in Figure 3. In A mode (amplitude mode), the echoes returning from along the line of sight of the transmitting receiving probe are shown as peaks proportional to the intensity of reflected ultrasound pulses at their respective depths. While this method of display is nowadays rarely applied, we shall describe it in more detail, because this helps to understand the more complicated and more often used B-mode display. The procedure is as follows: approximately 1000 times per second short pulses are transmitted into the body. Synchronously with the ultrasound pulse transmission, a line begins to be drawn on a cathode-ray tube (CRT) screen. Since the transmitted pulse is only 1 μs long, we have in principle 999 μs at our disposal for registration of echoes. Whenever an echo returns, it is received by the same probe, amplified, and then taken to the CRT tube where the line is deflected proportionally to the returned echo intensity, so that the peak is shown on the screen. The distance between peaks in the screen corresponds to some degree to the actual distance between the reflectors in the body. If the writing velocity at the screen is made exactly half of the speed of ultrasound in the body, the scale will be 1:1,

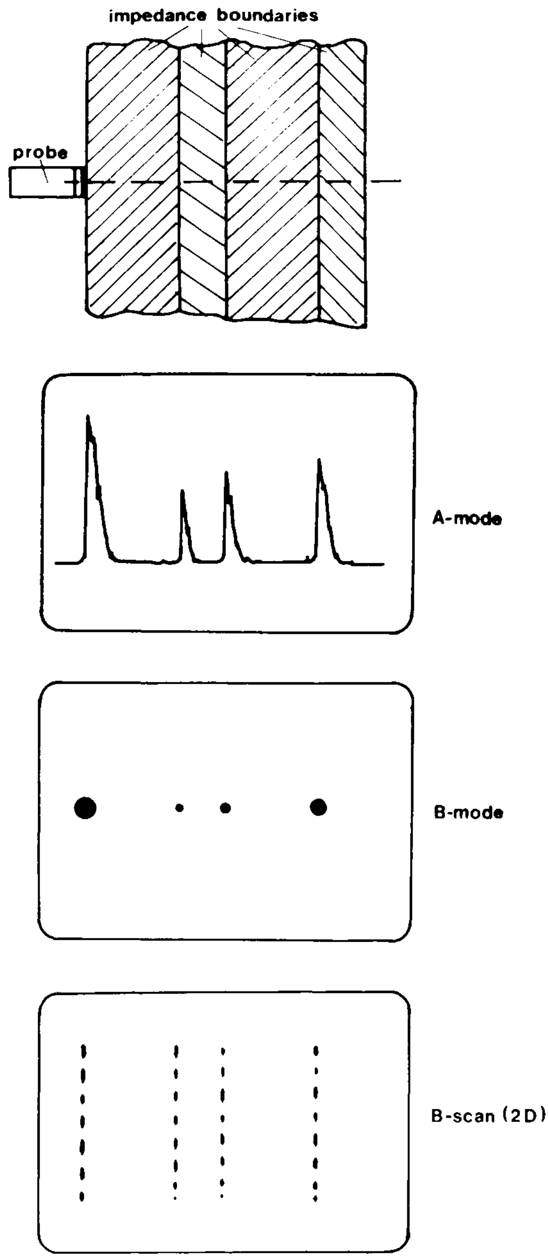

FIGURE 3. An illustration of A- and B-mode displays. Ultrasound pulses reflect from the tissue boundaries and are displayed as peaks or bright dots on the echoscope screen. If the transducer scans the body interior, the resulting bright dots in the B-mode form into a tomographic image of the scanned area.

because ultrasound makes a round trip in the body. This basic method of obtaining an A-mode display in many instances at present has been substituted by more indirect methods, although the principle is the same. The A mode is still used in neurology, ophthalmology, and indirectly in tissue characterization.

In the B-mode display (brightness modulation), the returned echoes are shown as bright

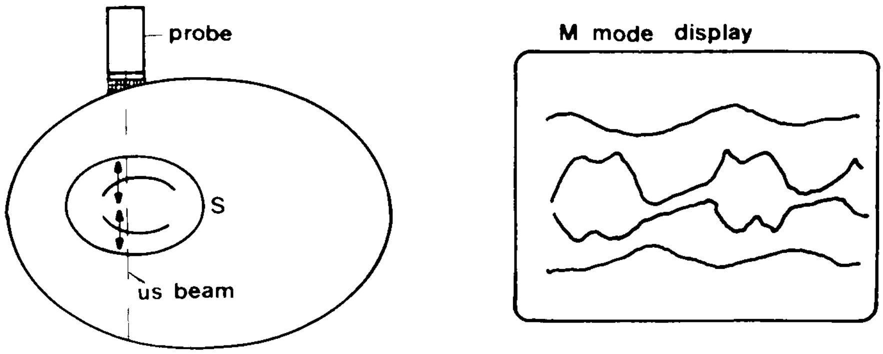

FIGURE 4. In M-mode display, the still structures are shown as straight lines and the moving ones as wavy lines. The ordinate is the depth and the abscissa is the running time.

dots on the screen. The position of bright dots on the screen corresponds to the position of corresponding reflectors due to the electronic system that measures the time necessary for echoes to return (the round trip time) and the position and angulation of the transmitting-receiving transducer. In real-time systems, which comprise a vast majority of all scanners, the transmitting-receiving probe is automatically moved or angulated, scanning in this way the interior of the body. There are some systems that electronically mimic this movement with the same result of scanning the body. In real-time systems, this ultrasound beam steering is done fast enough to produce a live image. The bright dots (Figure 3) are arrayed in the processor memory and then processed (smoothed, interpolated, etc.) and displayed. The image is a tomographic (section) image of the interior of the body in front of the ultrasound scanning probe. It consists basically of data along the lines corresponding to probe lines of sight. In practice, the section thickness depends on the probe, which we describe in detail later on, and scanner setting too. The thickness in practice range is approximately between 2 and 10 mm.

M mode (movement) is another way of representing ultrasonic echoes, in particular the moving structures. Figure 4 illustrates obtaining an M-mode display. Here too, the echoes are represented by bright dots on the screen, but this time the display is not linked to the system or continuous measurement and determination of transmitting transducer angle, but the transducer is continuously aimed in the same, fixed direction of the moving structure of interest (e.g., heart). The system measures the depths of the moving structures and displays these (changing) depths as they change in time. One axis, usually the abscissa, thus is the running time; the other, usually the ordinate, is the reflector depth. In this way, it is possible to show quantitatively the movement characteristics of, e.g., heart valves, and measure their opening and closing velocities. M-mode display plays an important role in echocardiography, and elaborate methods of detailed processing of data obtained in this way have been developed.

A. Main Blocks of an Echoscope

An echoscope uses short ultrasound pulses to obtain two-dimensional (2D) B-mode images in roughly the following way: short pulses are transmitted from a transmitter-receiver probe into the body. The echoes returning from reflective structures within the body are picked up with the same probe and then electronically processed to obtain a section image on a television (TV) screen. This can be accomplished because, in measuring the time necessary for echoes to return, one knows the velocity of ultrasound along the line of sight or the probe. Echoes that traveled a longer way, return later and are more attenuated. Therefore, the later-coming echoes (echoes coming from deeper structures) must be more amplified in

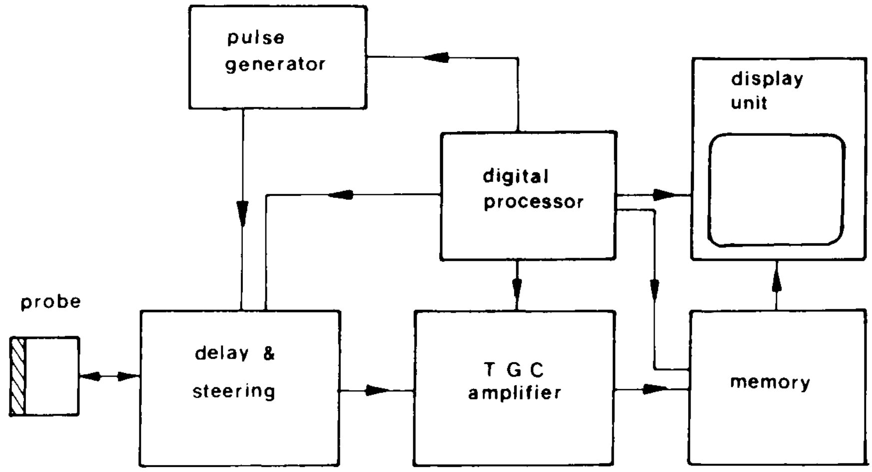

FIGURE 5. Block diagram of an echoscope. A digital processor directs the pulses from a pulse generator via a delay and steering unit into the probe. Signals from the probe (echoes) are processed in the delay unit, taken to the TGC system, amplified, memorized, and prepared for display. The whole process is synchronized and coordinated by the digital processor (computer) and its software.

the echoscope in order to compensate for this attenuation. Such amplified, compensated signals are taken to the digitizing system, which converts the data into a form suitable for digital storage. The procedure is controlled by at least one microprocessor. The content of the microcomputer memory is then displayed on a TV monitor. A simplified block diagram of a system capable of doing this is shown in Figure 5.

The probe is the crucial element of an echoscope and its properties dictate many of the system's characteristics. The probe contains one or more ultrasound transducers. A transducer is a device capable of converting one kind of energy or signal into another kind; in our case, the transducers convert electrical pulses into ultrasonic (mechanical) pulses and vice versa. The same transducers act as transmitters and receivers of ultrasound at different times.

In the beginning of a transmission cycle, the pulse generator generates a very short electrical pulse (tens to some hundreds of volts). This impulse is taken to the transducers in the probe via a control and delay unit, which takes the pulse from the pulse generator to the appropriate transducers in the probe at the appropriate times. After the transmission of the short ultrasound pulse is completed, the system waits for the echoes from the body to return. As the echoes return back to the probe, the transducers pick them up and generate electrical signals, which are now taken to the complicated amplifier, the time gain compensation (TGC) amplifier, which amplifies the signals. The signals that come later are amplified more in order to compensate for attenuation of ultrasound in the body. The amplified signals, which represent reflectors within the body, are stored in a digital memory together with data on the depth and direction they returned from. The depth is inferred from the time interval between the transmission of the pulse and the reception of an echo (taking the average ultrasound propagation speed of 1540 m/s into account). The direction is known from the probe steering data, which are stored in the microprocessor, which controls the scanning procedure. The data from the memory are taken to the display unit and displayed. For various reasons, one cannot simply display all the data as they come from the body, so they are processed before being shown on the screen. The processing always includes some sort of compression of data because the range of ultrasound intensities of interest which return from the body is much larger than the range that can be shown linearly as different brightnesses on a TV screen. On the other hand, different echo intensities are encoded as

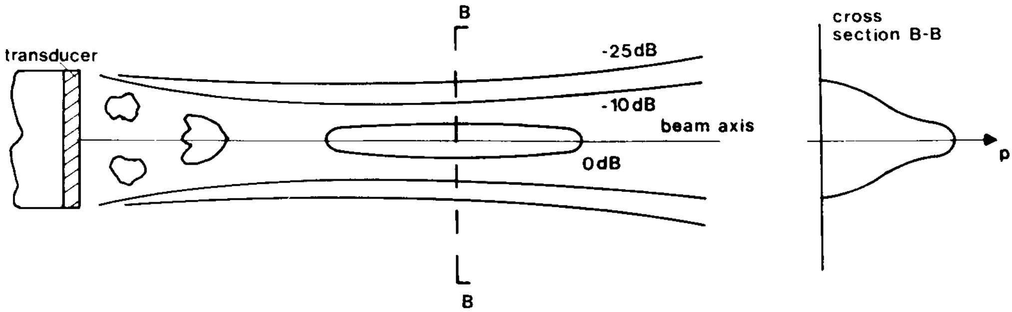

FIGURE 6. An ultrasonic beam intensity is not uniform or sharply cut off. It can be described with isointensity or isopressure curves. In the focal region it is narrowed. The intensity near the transducer is irregular.

different shades of grey on the screen. This transfer function is therefore nonlinear and can be changed with controls. Apart from this, in modern scanners there regularly is an additional interpolation program, which makes the image better adjusted to human perception characteristics, and in some more sophisticated systems, it mathematically improves the resolution of the details by taking into account some characteristics of the probe and ultrasound propagation through the body.

In the history of ultrasound echography (i.e., only 10 years ago), there was one other type of scanner — the static scanner — which played an important role in obtaining good quality images. These systems, now rare, used a single probe, which was hand moved by the operator with the consequence that the operator had a freedom of choice of the scanning format. The imaging was not real time, but the images were of good quality and with a fixed geometrical relation to the patient couch. The angle and position of the probe were measured by an electromechanical device from which the probe was hanging. At present, these systems are rarely manufactured, so we shall not describe them in detail.

B. Transducer and the Ultrasound Beam

As mentioned above, the transducer converts ultrasound signals into electrical signals and vice versa. The active element of a transducer is a piezoelectric ceramic element or a piezoelectric plastic foil. The property of a piezoelectric material is to deform under the influence of mechanical force (stress), and to generate electrical charge on its sides if it is subjected to mechanical stress. In echography, electrical pulses are taken onto the sides of piezoelectric transducers inducing vibrations in them, which are transmitted into the body as ultrasonic vibrations. Conversely, the echoes from the body induce minute deformations of the transducers so that electrical charge appears at their sides covered with conductive layers from where it is taken to amplifiers in the echoscope.

We must pay more attention to the volume in the medium (water, human body, etc.) which is occupied by this ultrasonic energy and which we call an ultrasonic beam. Figure 6 shows schematically the ultrasonic beam in front of a transducer. It is represented by curves that connect points of equal ultrasound intensity — the isointensity curves. A beam like this can be obtained only in nonattenuating media (degassed water is near enough to this for our purposes). In attenuating media, the actual intensity is modified by attenuation, but the echoscope will "see" an equivalent beam like the one shown in Figure 6 if the TGC amplifier is correctly adjusted.

The intensity is not sharply cut off, but gradually decreases to zero on the sides. If the transducer is focused, then the beam has a narrowing around the focus. Near the transducer, ultrasound intensity is irregular due to interference of ultrasound waves from different parts of the transducer face. Far from the transducer, ultrasound intensity decreases regularly and

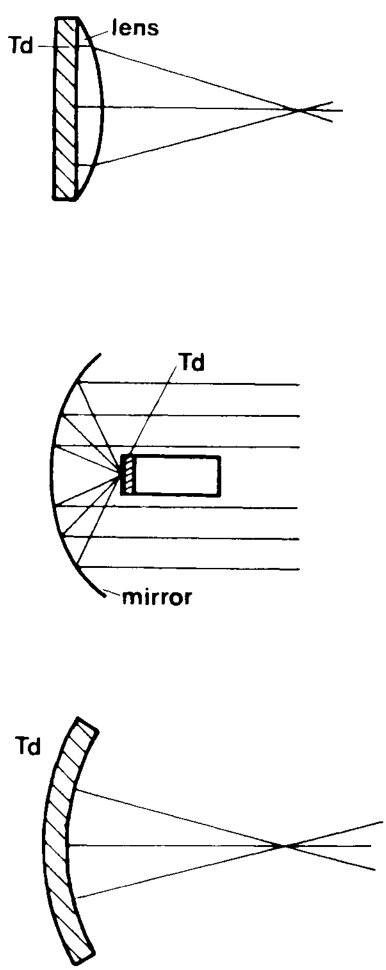

FIGURE 7. Ultrasonic waves can be focused with lenses, mirrors, or curved transducers and combinations thereof.

continuously. The two zones can be mathematically defined. The near zone is called the Fresnel zone, and the far zone is the Fraunhofer zone. In pulse operation, the near zone is less irregular than in the continuous wave (CW) operation.

Ultrasound waves can be focused with lenses, mirrors, curved transducers, and electronically. Figure 7 illustrates the methods of mechanical, fixed focusing. The lenses are built of plastic materials, and the shape of a focusing lens can be convex or concave, depending on ultrasound propagation speed in the lens material. Ultrasound mirrors are built as thin structures containing air in order to obtain nearly total reflection. During the production stage, a curved transducer is formed into a focusing shape (e.g., paraboloid). All these focusing methods have a focusing distance determined by their shape, which cannot be changed. In practice, this means that there should be as many different probes as many different foci are needed. Since the focusing zones are fairly long, the number of probes at one frequency would not be more than three, usually two. Electronic focusing is illustrated in Figure 8. One can electronically focus a beam only if the transducer is composed of separate elements, which can be separately activated, and which can separately receive signals from the body. In our example, the composite transducer consists of three rings,

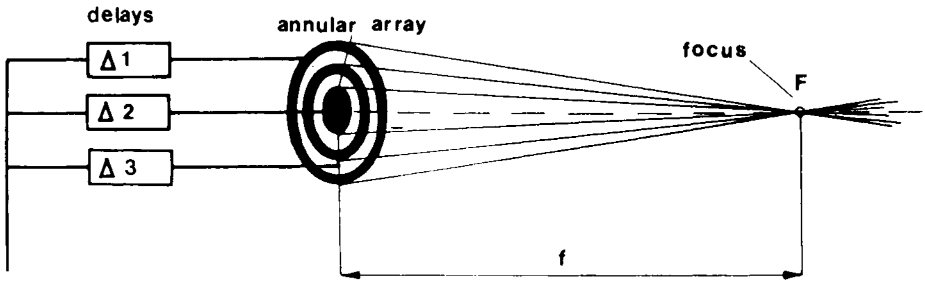

FIGURE 8. An anular array consists of anuli of piezoelectric material that can be activated separately. The focus of such a composite transducer can be adjusted by changing the delays. Larger delays between activations yield a nearer focus.

one within another. If we define a focus depth (f) along the transducer axis, we can see that the distances from different rings to the focus point (F) are different, so that ultrasound pulses need different times to reach F. If we now transmit from the outermost ring first, then from the second ring, and after another delay from the central transducer, with the delays in activation adjusted so that they compensate for the different travel times, then the ultrasound pulses from all the three rings meet at the same time at point F. This in fact is focusing. The distance f depends on the delays and can be chosen by outside controls. Therefore, such a focusing system is flexible with a broad choice of focusing depths.

So far, we have spoken about focusing transmission. Reception focusing follows basically the same rules. In fixed focusing transducers, the transmission and the reception characteristics are equal and the round trip focusing is a result of both. In electronic focusing, these two can be made different, in fact, one can make the receiving focus follow the ultrasound pulse as it transverses the body, thus making a nearly continuous "dynamic" focus.

Lately, a mathematical method for improving the resolving power of echoscopes, previously used only in military applications, has been implemented in medical echoscopes. This computed echography is based on the fact that if the response of a system on the simplest reflector is known and sufficient computer power is provided, one can untangle the effects of the beam shape from the real characteristics of the imaged object.

Properties of the ultrasound beam are very complicated; we mentioned only the most important and those that are of greater practical importance to the user of modern scanners. Other characteristics of the beam such as sidelobes (auxiliary beams) and quantitative evaluation of beam properties can be found in more detailed and dedicated literature.

So far we have spoken only of the distribution in space, but the time distribution, the pulses, deserve mention too. The pulses sent out from the transducer are short, usually 2 to 4 cycles long. This means that for higher frequencies, the pulses can be made shorter, with a consequence of a better depth (axial) resolving power. The actual length of such pulses is about 1 μs (1 millionth of a second). This means that when defining the intensity of ultrasound, one must clearly state whether it is the average intensity or the intensity during only the pulse itself. The difference is large and of utmost importance when considering possible hazards of ultrasound. The time average intensity can be calculated by multiplying the intensity during the pulse with the ratio of "on" time (actual transmission time) to total time. If there are 1000 pulses per second and each of them has a duration of 1 μs, the ratio is 1/1000, i.e., the time average is 1 thousandth of the intensity during the pulse duration. Of course, when considering possible hazards, one must also take into account the space concentration (focusing) by taking the ratio of the surface area of the transducer and the area of the cross section at the focus. Therefore, when looking at some quoted intensity data, one must be aware of whether it is the time-average-space-average (SATA) or space-peak-time-peak (SPTP) value. Examining the matter in more detail, it turns out to be more complicated, but the above outline is the basic logic.

C. Echoscope Probes and Scanning Systems

The echoscope probe is the device containing one or more ultrasound transducers used for transmitting and receiving ultrasound. At present, the probes are built in such a manner as to automatically scan the interior of the body with which they are put in contact via some coupling agent (oil, gel). They usually operate fast enough to give a real-time image, (i.e., about 20 images per second). Occasionally, the images are composed slower with an added quality in detail resolution. The type of probe determines to a great extent the scanner properties and field of application.

Figure 9 illustrates a number of probes presently in routine use. Probe 9a is called a linear array. It contains a number of narrow, ribbon-like transducers, all of them separately connected to a cable and connector. There are usually about 64 such transducers in a probe of between 5 and 12 cm in length. Each of the transducers can have additional grooves etched to improve some of the directivity characteristics. The width of the ribbon-like transducers is too small to make a good beam, so one must activate more than one transducer at a time to obtain a composite transducer much larger than wavelength. A way to do this is to activate groups of transducers, e.g., first group 1-10, then group 2-11, then group 3-12, and so on all the way to group 55-64. In this way, one obtains a shift equal to the width of a single transducer with an apparent composite transducer ten single widths wide. Furthermore, the shifting is the actual scanning of the area in front of the probe, without any mechanical movement. In addition to the shifting-scanning, one adds delays to activation of single transducers in a group obtaining electronic focusing as previously explained for the composite ring transducer. Of course this electronic focusing applies to only one plane. In the other plane, there is a fixed focus lens. This type of transducer is used mainly in obstetrics, breast, and thyroid scanning.

Figure 9d and c illustrates the most commonly used mechanical sector scanners. They produce an approximately triangular image. In both types, the transducers are mechanically moved, thereby scanning the area in front of them. The probe illustrated in Figure 9d is a rocking probe, i.e., a transducer that is made to rock in front of an acoustical window in order to scan whatever is in front of the window. The rocking movement is not smooth but rather stepwise, and the probe spends at each increment angle the time needed for the transmission-reception sequence.

The probe illustrated in Figure 9c contains three to five transducers mounted on a turning wheel. Each of the transducers is activated and used only at the time when it traverses in front of the acoustic window. This movement is stepwise too, and the data are gathered along predefined lines with each of the transducers. In fact, the images obtained with the different transducers overlap. The property of the sector transducers is that they have a small acoustic window and can look to the side. Therefore, they are applied for gynecology, the upper abdomen, and cardiology.

The probe illustrated in Figure 9e uses an anular array for focusing and an ultrasonic mirror for scanning. The ultrasound pulses are transmitted onto the mirror and the echoes return to the anular array transducer via the mirror. The mirror itself is tilting, thereby sending the beam in different directions within the body.

The probe illustrated in Figure 9b is called a curvilinear probe (sometimes called a convex probe) and is basically built and activated like the linear array, but yielding an image format between the rectangular (like the linear array) and sector format. The image looks like a trapezoid.

The phased array illustrated in Figure 9f has the transducers mounted in an array like the linear array, but it is much shorter (1 to 2 cm). In this case, both the focusing and the beam

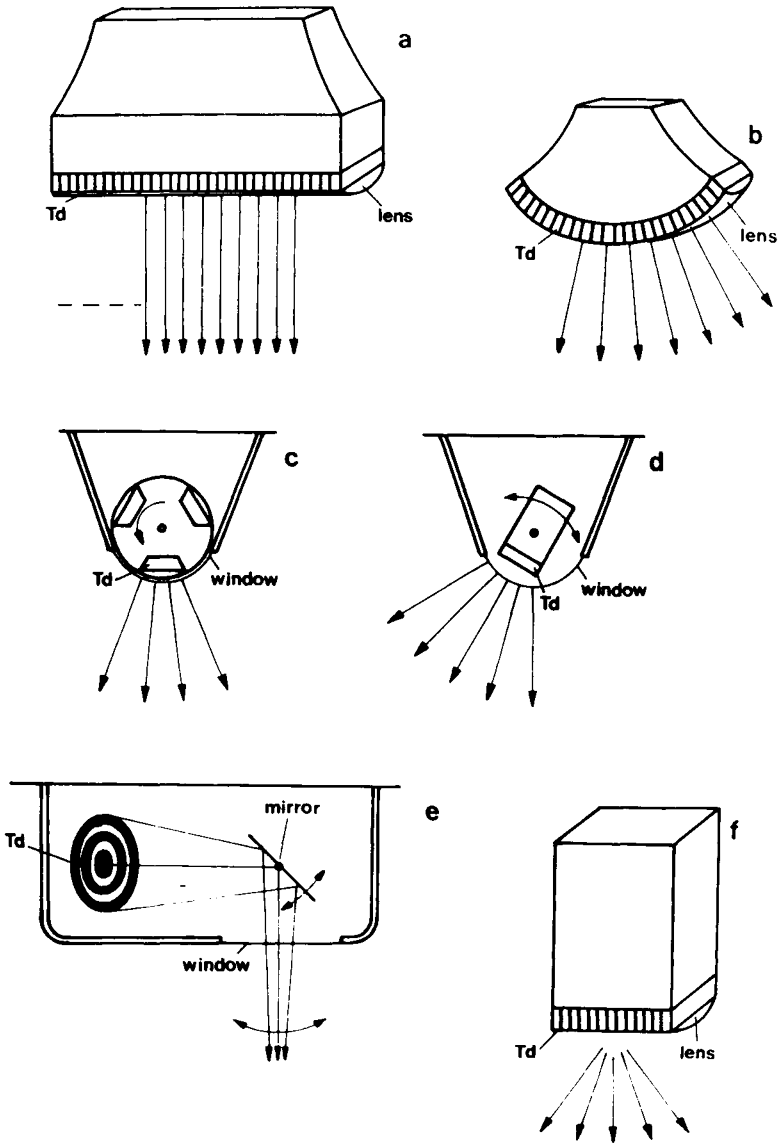

FIGURE 9. Real-time probe types. (a) A linear array of transducers (td). Square image format. (b) A curvilenear probe. A trapezoid image and an acoustic window of dimensions between the linear and sector probes. (c) A rotating mechanical sector. Sector image and small acoustic window. (d) A rocking mechanical sector probe. Sector image and small acoustic window.(e) An anular array for focusing and an ultrasonic mirror for scanning mounted in a housing yield high-quality sector images. (f) A phased array uses electronic delays for both steering and focusing of the beam.

steering (change of its angle) are achieved with different delays in activation of the arrayed transducers. The probe produces a sector image without the use of moving parts. For a long time this system was very esteemed in echocardiography, but was less used in the upper abdomen due to its problems with beam sidelobes. At present, the problem of sidelobes has been solved to a great extent in the higher class instruments.

The probes as described are the most common types used in transcutaneous contact scanning. There is a large variety of ways of combining these principles and adjustments for specific uses.

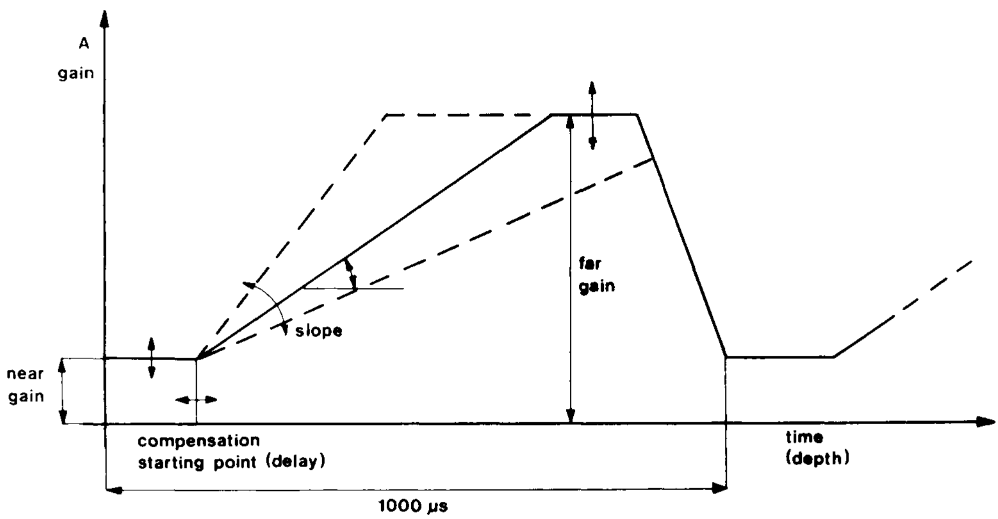

FIGURE 10. An illustration of the TGC amplifier gain variation. The near, the far, and the overall gain can be changed. The compensation slope can be adjusted and so can the compensation starting point (delay).

For ophthalmology and breast scanning, one often uses built-in waterbath probes and water transducers. Special probes for intraoperative use, for transvaginal, transvesical, or transrectal scanning have been developed. Transesophageal probes for echocardiography have been developed. Special attachments or holes in the probes for puncture needle guidance have been developed in order to keep the needle within the scanning plane under a predefined angle. All these are adaptations of the above mentioned principles to specific and special needs. To describe them all would require many printed pages, but they can all be understood if the basic principles are clear.

D. Attenuation Compensation: TGC

We have already mentioned the TGC system that is used for compensation of ultrasound attenuation in tissues. Echoes that return from deeper reflectors must be amplified more because they have been more absorbed and scattered under way. The echoes from deeper structures come later and so the TGC system compensates the attenuation by amplifying more the echoes that come later. The gain changes from minimum to maximum during each transmission-reception cycle (i.e., 1000 times per second or so). Figure 10 is a graphical representation of gain change during one cycle. The actual attenuation follows an exponential law and so does the TGC amplifier — with the opposite sign. The graphical representation in Figure 10 shows the increase of gain as linear for simplicity. Basically, one can by separate controls change the initial gain, the gain slope, the far gain, and the compensation starting point. In more attenuating tissues, the compensation slope must be steeper. The TGC system thus changes the relative amplifications at different depths. One can, with a separate control, change the overall gain, i.e., push up or lower the whole gain curve in Figure 10. In Figure 11, some examples of adjustment and misadjustment of the TGC system are shown.

There are some systems that have a present TGC function and then a set of sliding controls for modification of the TGC curve to particular needs and at different depths.

E. Dynamic Range

When speaking about the dynamic range of bright dots seen on a TV screen, we mean the ratio of brightnesses of the brightest and the least bright spot still visible on the screen. If we speak of the dynamic range of echo intensities of interest, we mean the ratio of the