

Princípio básico

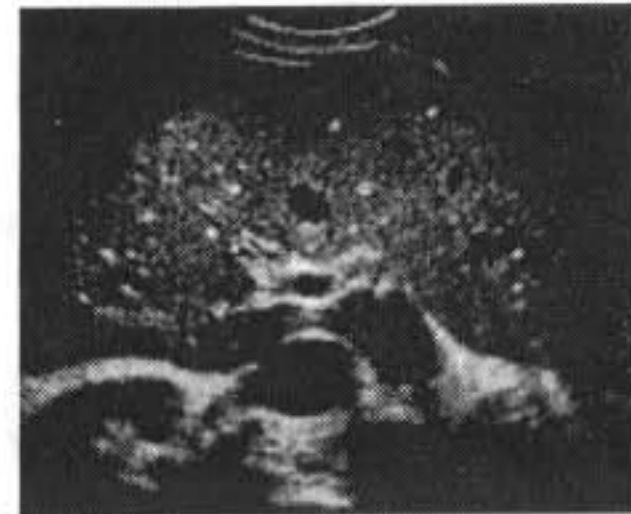
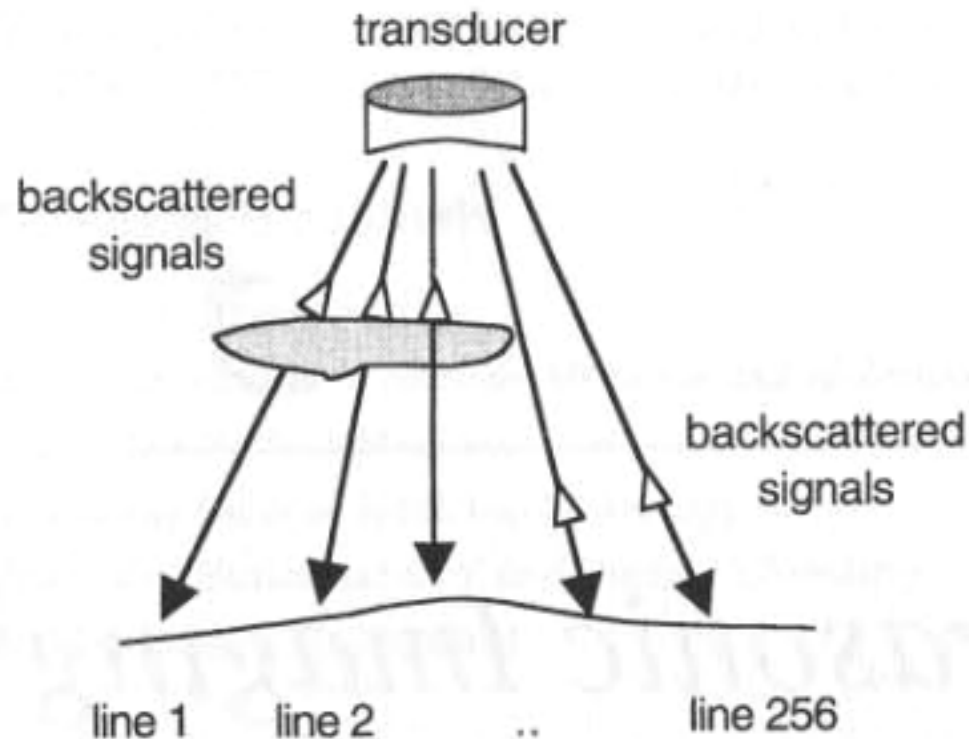
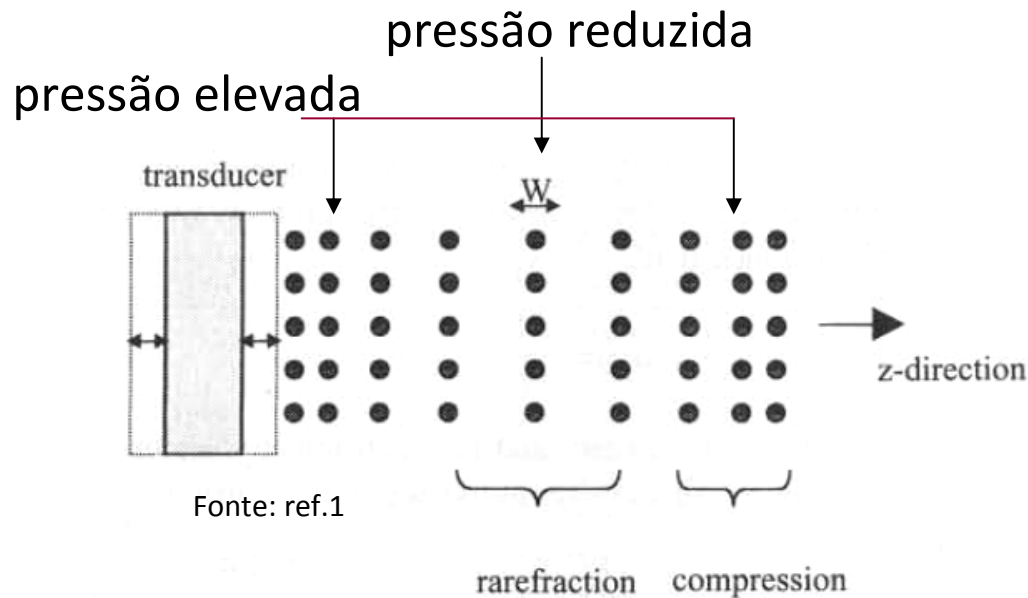


FIGURE 3.1. (Left) The basic principles of ultrasonic imaging. An ultrasound transducer transmits mechanical energy into the body. Part of the energy is backscattered from tissue boundaries and small structures and is detected by the transducer. By scanning through the body with a number of adjacent ultrasound beams, an image is formed. (Right) A two-dimensional image of the liver (©2000 ATL Ultrasound). The brightness of each pixel in the image is representative of the amount of energy backscattered at that point.

Fonte: ref.1

Ondas sonoras



u = velocidade do material (1 a 10m/s)

$$u = \frac{dW}{dt}$$

p = alterações de pressão

$$p = Zu = \rho cu$$

Análogo à lei de Ohm

Z = Impedância do meio [$\text{kg m}^{-2}\text{s}^{-1}$]

$$Z = \frac{p}{u} = \rho c$$

W = deslocamento do material relativamente à posição de repouso

Equação de onda plana num meio elástico:

$$\frac{\partial^2 W}{\partial z^2} = \frac{1}{c^2} \frac{\partial^2 W}{\partial t^2}$$

$$c = \sqrt{K / \rho} = \sqrt{1 / (k \rho)}$$

c = velocidade do som

K = módulo de elasticidade [Pa]

$k = 1/K$ = compressibilidade [Pa^{-1}]

ρ = densidade mássica [kg/m^3]

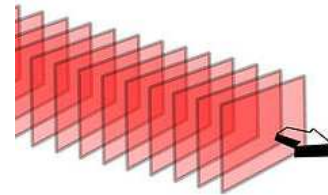
Solução geral

$$W = \underbrace{F(z - ct)}_{\text{onda} \rightarrow} + \underbrace{G(z + ct)}_{\text{onda} \leftarrow}$$

F e G , funções arbitrárias

Ondas harmónicas planas

Caso particular alternado sinusoidal



$$u = u_0 e^{ik(z \pm ct)} = u_0 e^{i(kz \pm \omega t)}$$

$$p = p_0 e^{i(kz \pm \omega t)}$$

$$W = \frac{u_0}{\pm i\omega} e^{i(kz \pm \omega t)}$$

$\lambda = cT$ comprimento de onda

$T = 1/f$ período

$k = 2\pi/\lambda$ número de onda

$\omega = kc$ frequência angular

$f = \omega/2\pi$ frequência

$i^2 = -1$



Intensidade da onda sonora

$$I = \frac{1}{2} p_0 u_0 = \frac{p_0^2}{2Z} \quad [W / m^2]$$

$$DB = 10 \log(I_{rms} / I_{ref})$$

$$= 20 \log(p_{rms} / p_{ref})$$

$$I_{ref} = 1 pW / m^2$$

$$p_{ref} = 1 \mu Pa$$

Propriedades acústicas dos materiais biológicos

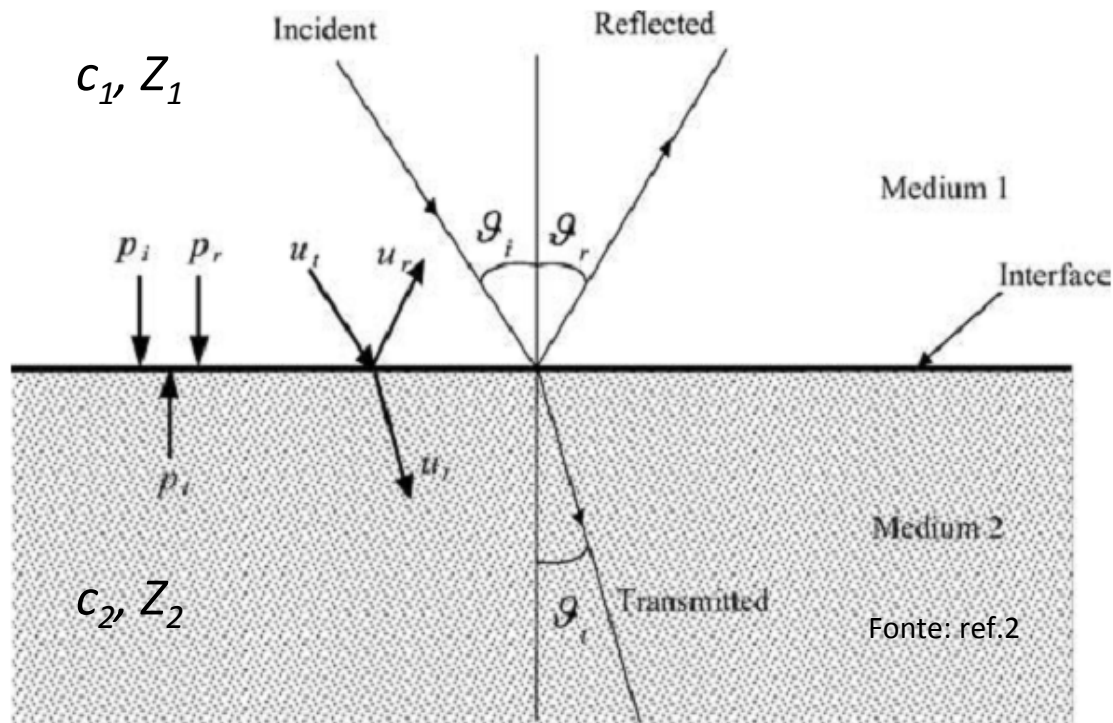
TABLE 3.1. Acoustic Properties of Biological Tissues

Fonte: ref.1

	Characteristic Acoustic Impedance $\times 10^5 (\text{g cm}^{-2} \text{s}^{-1})$	Speed of Sound (m s^{-1})
Air	0.0004	330
Blood	1.61	1550
Bone	7.8	3500
Fat	1.38	1450
Brain	1.58	1540
Muscle	1.7	1580
Vitreous humor (eye)	1.52	1520
Liver	1.65	1570
Kidney	1.62	1560

Assume-se $c \approx 1540 \text{ m/s}$ nos tecidos moles

Reflexão e transmissão através de uma descontinuidade



Leis de Snell

$$\theta_i = \theta_r$$

$$\frac{\sin(\theta_i)}{c_1} = \frac{\sin(\theta_t)}{c_2}$$

ou

$$\lambda_1 \sin(\theta_i) = \lambda_2 \sin(\theta_t)$$



Coeficientes de reflexão e transmissão

Pressão

$$R_p = \frac{p_r}{p_i} = \frac{Z_2 \cos(\theta_i) - Z_1 \cos(\theta_t)}{Z_2 \cos(\theta_i) + Z_1 \cos(\theta_t)}$$

$$T_p = \frac{p_t}{p_i} = \frac{2Z_2 \cos(\theta_i)}{Z_2 \cos(\theta_i) + Z_1 \cos(\theta_t)}$$

Intensidade

$$R_I = \frac{I_r}{I_i} = \left(\frac{Z_2 \cos(\theta_i) - Z_1 \cos(\theta_t)}{Z_2 \cos(\theta_i) + Z_1 \cos(\theta_t)} \right)^2$$

$$T_I = \frac{I_t}{I_i} = \frac{4Z_1 Z_2 \cos^2(\theta_i)}{(Z_2 \cos(\theta_i) + Z_1 \cos(\theta_t))^2}$$

$$R_I + T_I = 1 \quad (\text{conserv. da energia})$$

Incidência perpendicular

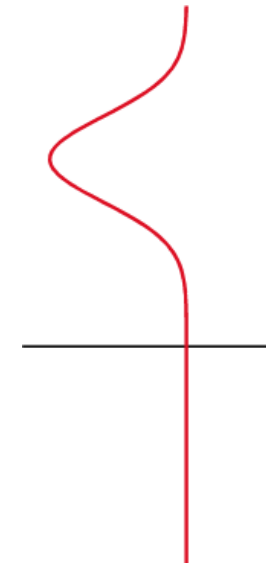
$$R_{p\perp} = \frac{p_r}{p_i} = \frac{Z_2 - Z_1}{Z_2 + Z_1}$$

$$T_{p\perp} = \frac{p_t}{p_i} = \frac{2Z_2}{Z_2 + Z_1}$$

$$R_{I\perp} = \frac{I_r}{I_i} = \left(\frac{Z_2 - Z_1}{Z_2 + Z_1} \right)^2$$

$$T_{I\perp} = \frac{I_t}{I_i} = \frac{4Z_1 Z_2}{(Z_2 + Z_1)^2}$$

Igual impedância \Rightarrow reflexão nula e transmissão perfeita



Coeficientes de reflexão para materiais biológicos

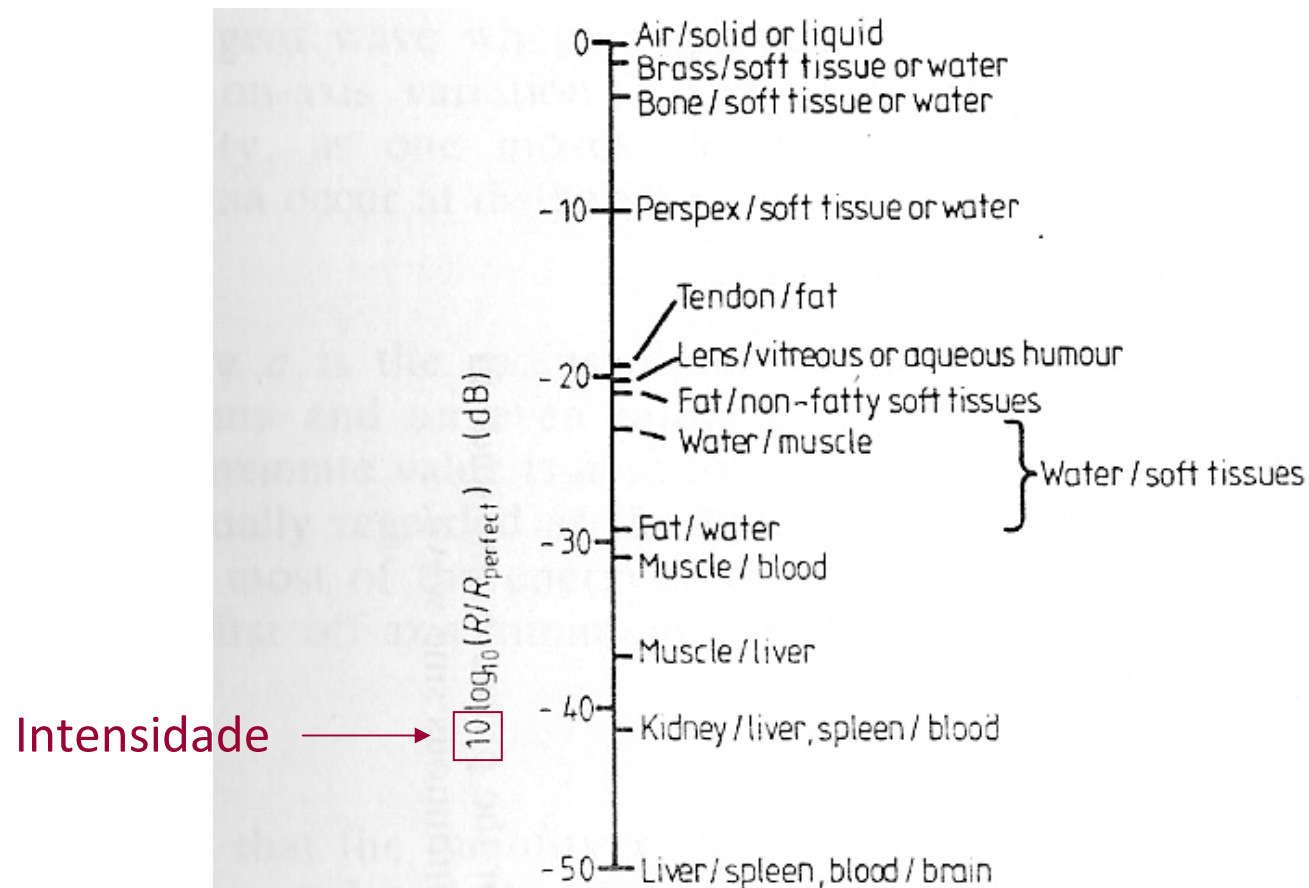
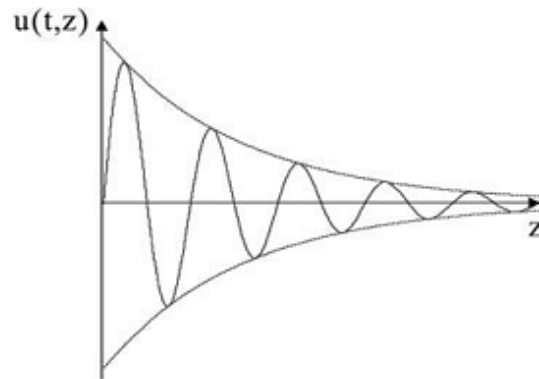


Figure 7.4 Calculated reflection coefficients (in decibels relative to a perfect reflector) for sound at normal incidence to a variety of hypothetical boundaries between biological and non-biological media. Fonte: ref.7

Atenuação



$$p(z,t) = e^{-\alpha z} \underbrace{p_0 e^{i(kz \pm \omega t)}}_{\text{onda plana}}$$

$$I(z) = I(z=0)e^{-2\alpha z} = I(0)e^{-\mu z}$$

$$[\mu] = [\alpha] = m^{-1}$$

$$\mu(D\text{B} / \text{cm}) = -\frac{1}{z} 10 \log \frac{I(z)}{I(0)} = 4.343 \mu(\text{cm}^{-1})$$

$$\alpha(D\text{B} / \text{cm}) = -\frac{1}{z} 20 \log \frac{p(z)}{p(0)} = 8.686 \alpha(\text{cm}^{-1}) = \mu(D\text{B} / \text{cm})$$

TABLE 16.1 Acoustic Properties of Tissue

Fonte: ref.6

Material	C (m/s)	$\alpha(\text{dB}/\text{MHz}^y\text{-cm})$	γ	ρ (kg/m ³)	Z (MegaRayls)
Air	343			1.21	0.0004
Bone	3360	3.54	0.9	1789	6.00
Blood	1550	0.14	1.21	1039	1.61
Fat	1450	0.6	1.0	952	1.38
Honey	2030			1420	2.89
Liver	1570	0.45	1.05	1051	1.65
Muscle	1580	0.57	1.0	1041	1.645
Water @ 20°C	1482.3	2.17×10^{-3}	2.0	1000	1.482

nte



Piezoelectricidade

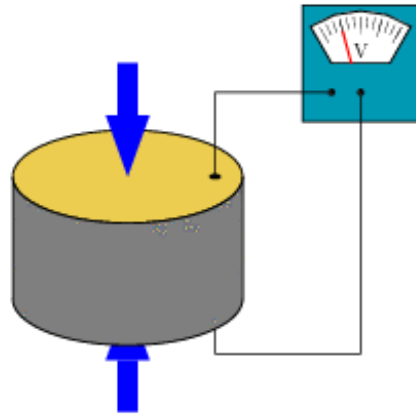


TABLE 65.1 Maerial Properties of Linear-Array Elements Made of PZT-5H Fonte: ref.8

Parameter	Symbol	Value	Units
Density	ρ	7500	kg/m ³
Speed of sound	c	3970	m/s
Acoustic impedance	Z	29.75	MRayls
Relative dielectric constant	ϵ/ϵ_0	1475	None
Electromechanical coupling coefficient	k	0.698	None
Mechanical loss tangent	$\tan \delta_m$	0.015	None
Electrical loss tangent	$\tan \delta_e$	0.02	None

PZT= cerâmica ferroeléctrica
lead-zirconate-titanate

O cristal é “piezoeléctrico”: expande-se e contrai-se ao ser aplicado um campo eléctrico, e vice-versa.

Serve como emissor e como microfone.

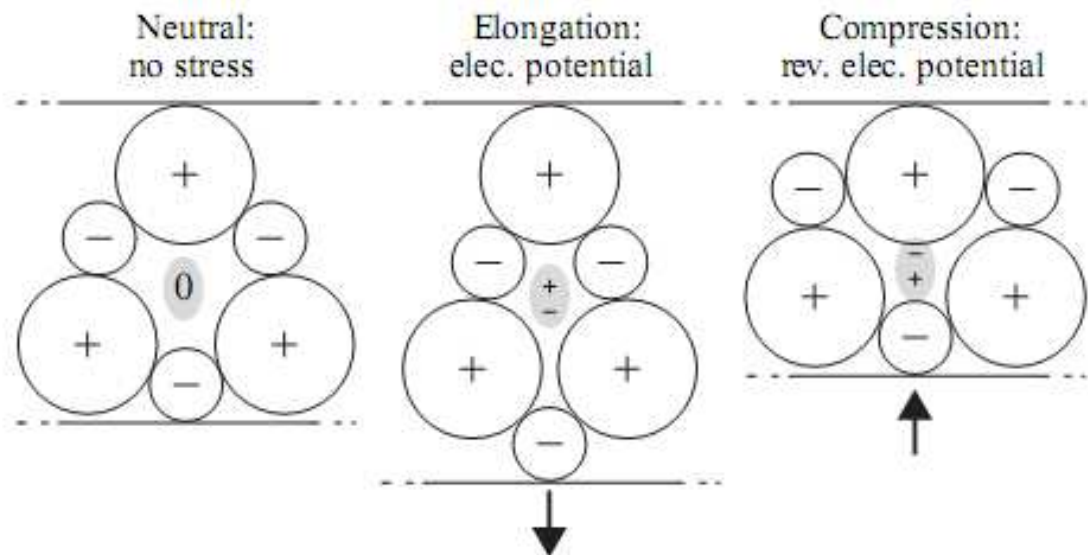
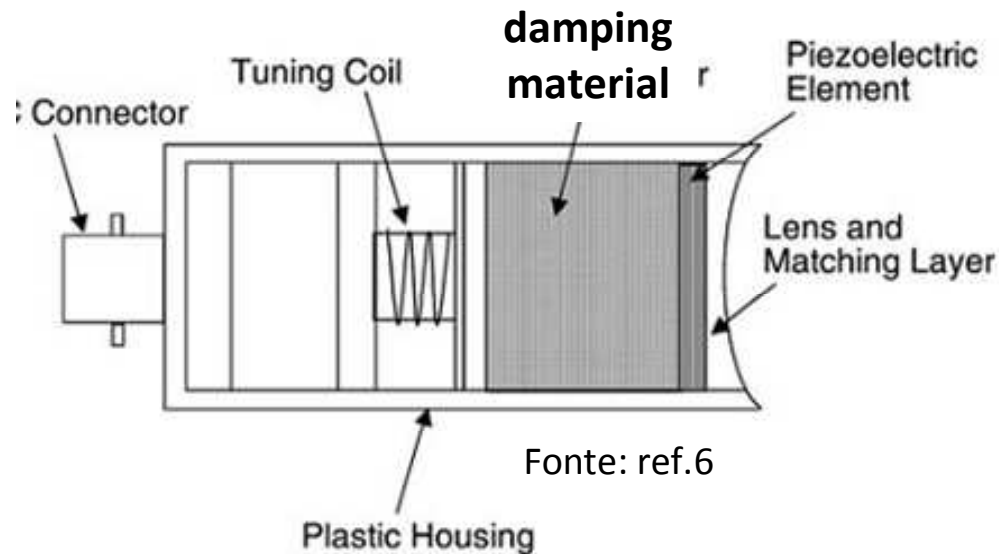


Figure 5. A piezoelectric crystal exhibits an electric charge on its surface when under mechanical stress (shown as the elementary cell response). The reverse effect is used to produce ultrasound; applying an alternating electrical potential across a piezoelectric crystal causes the crystal to vibrate along a given direction. Fonte: ref.3

Constituição de um transdutor de um único elemento



A transmissão de ondas ao paciente é maximizada se existir uma camada intermédia adaptadora (“Matching Layer”) de impedância

$$Z_{ML} = \sqrt{Z_{cristal} Z_{pele}}$$

Resposta eléctrica

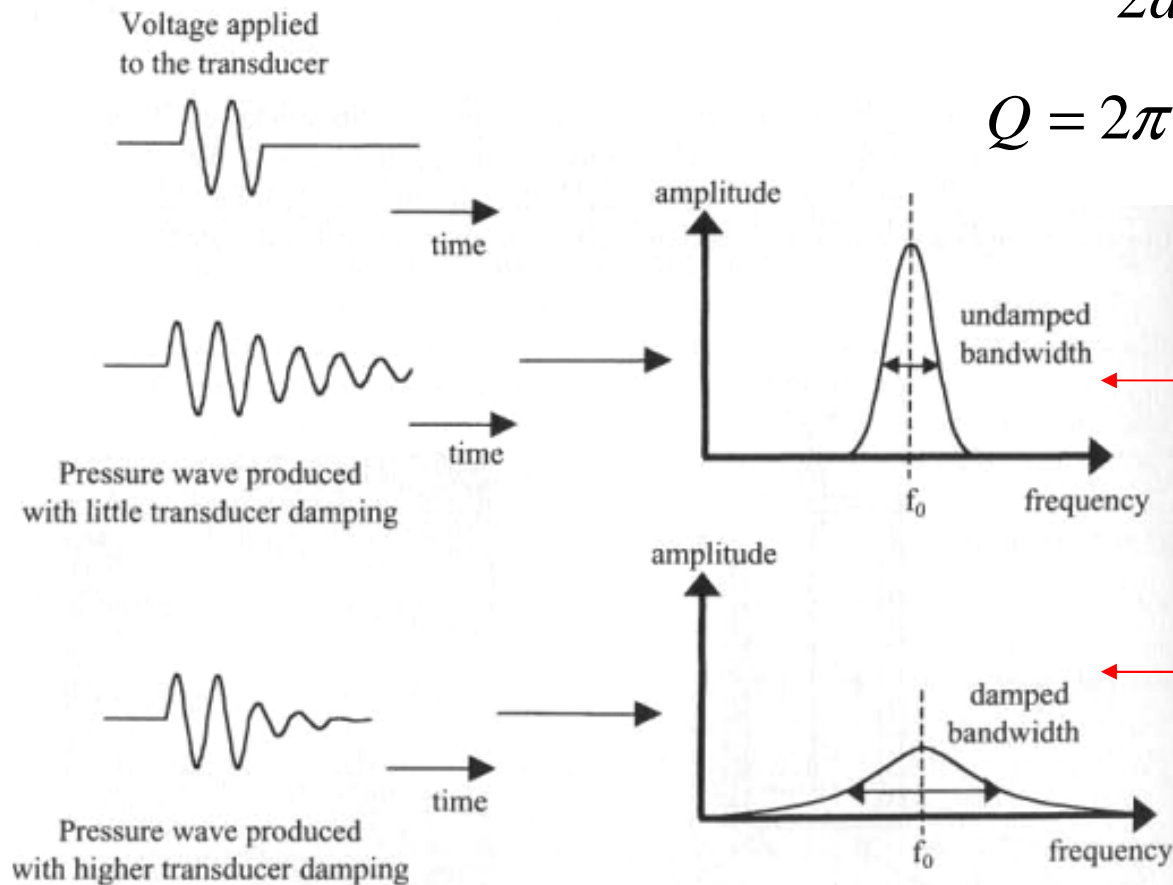


FIGURE 3.6. Time-domain and frequency-domain characteristics of damped transducers. A short pulse of alternating voltage is applied to the face of the crystal. With little mechanical damping, the crystal oscillates for a long time, producing a very sharp frequency spectrum with low bandwidth and high Q . With heavy damping, the crystal oscillations die out quickly, producing a much shorter pulse in the time domain and much broader bandwidth in the frequency domain.

Fonte: ref.1

$$f_0 = \frac{c}{2d}$$

$$Q = 2\pi \frac{f_0}{BW}$$

f_0 =freq. de ressonância

Q =factor de qualidade

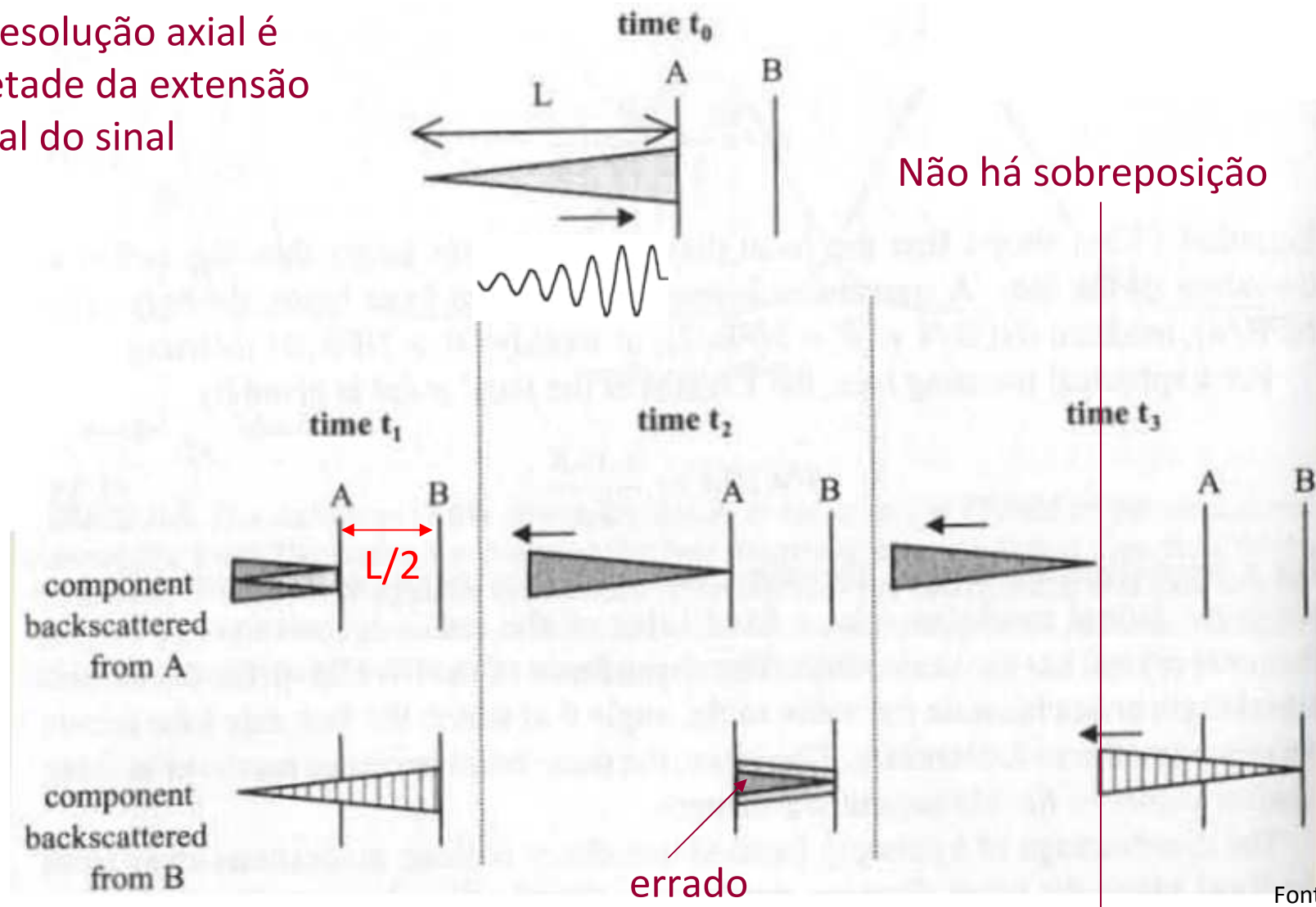
d = espessura do cristal

BW =largura de banda

Este efeito utiliza-se também nos relógios de cristal (Q pode ser extremamente elevado e f_0 muito estável)

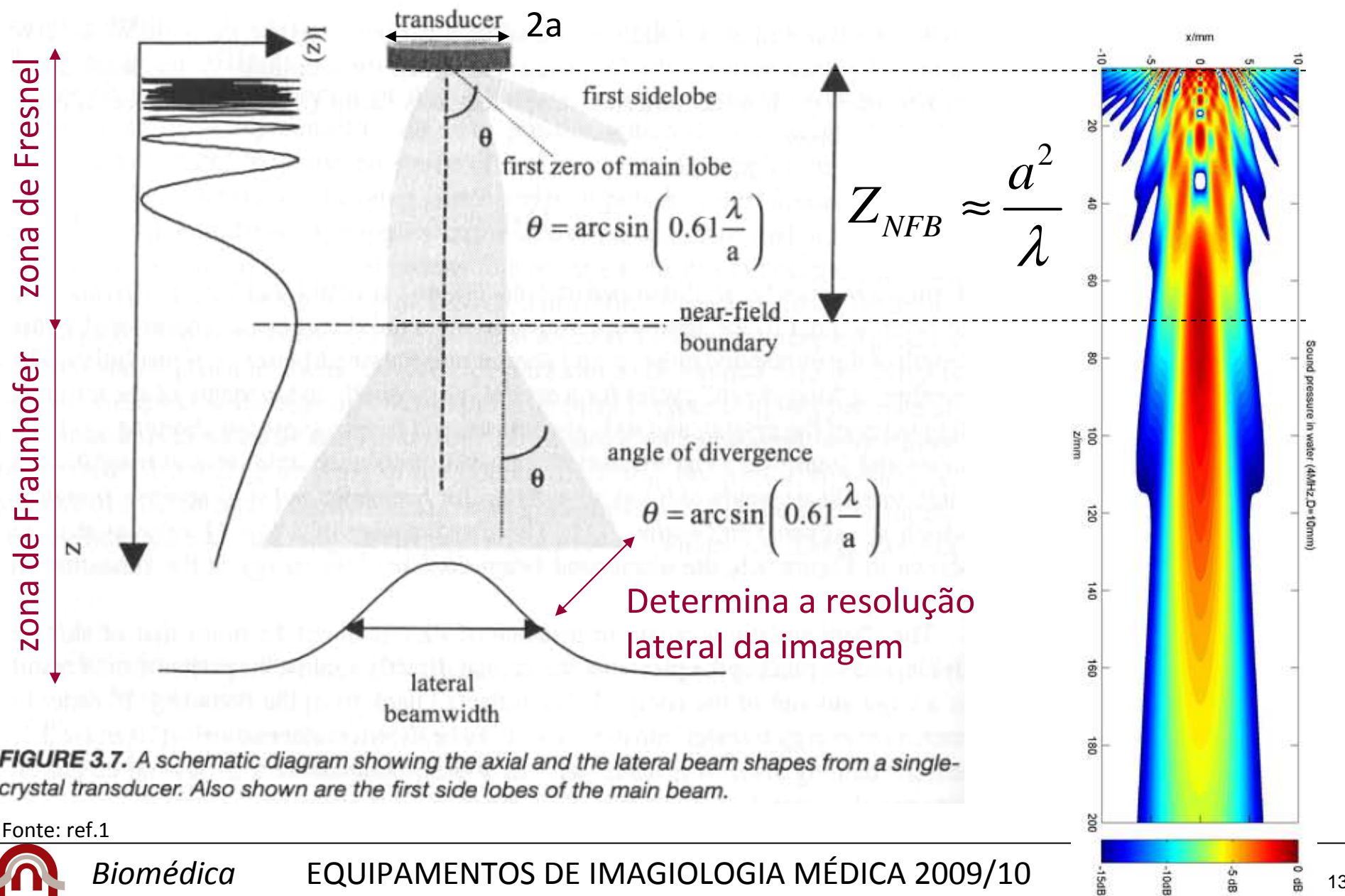
Resolução axial

A resolução axial é metade da extensão axial do sinal

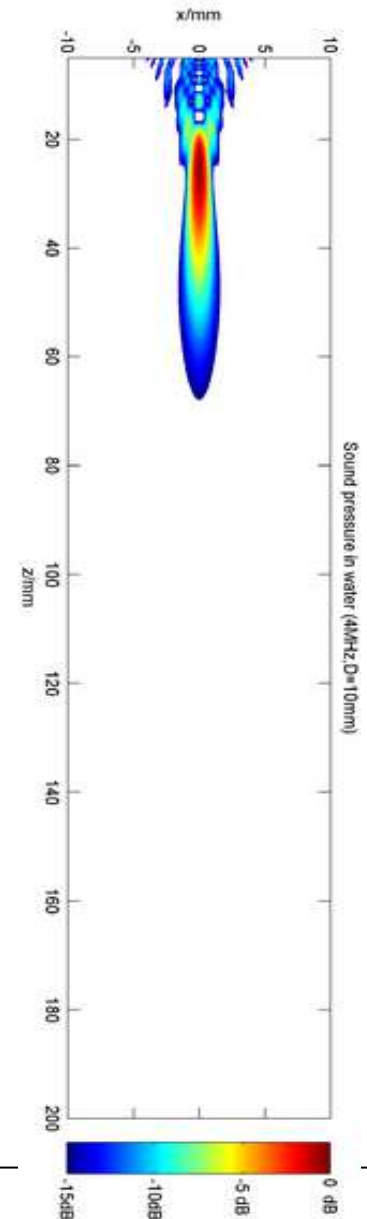
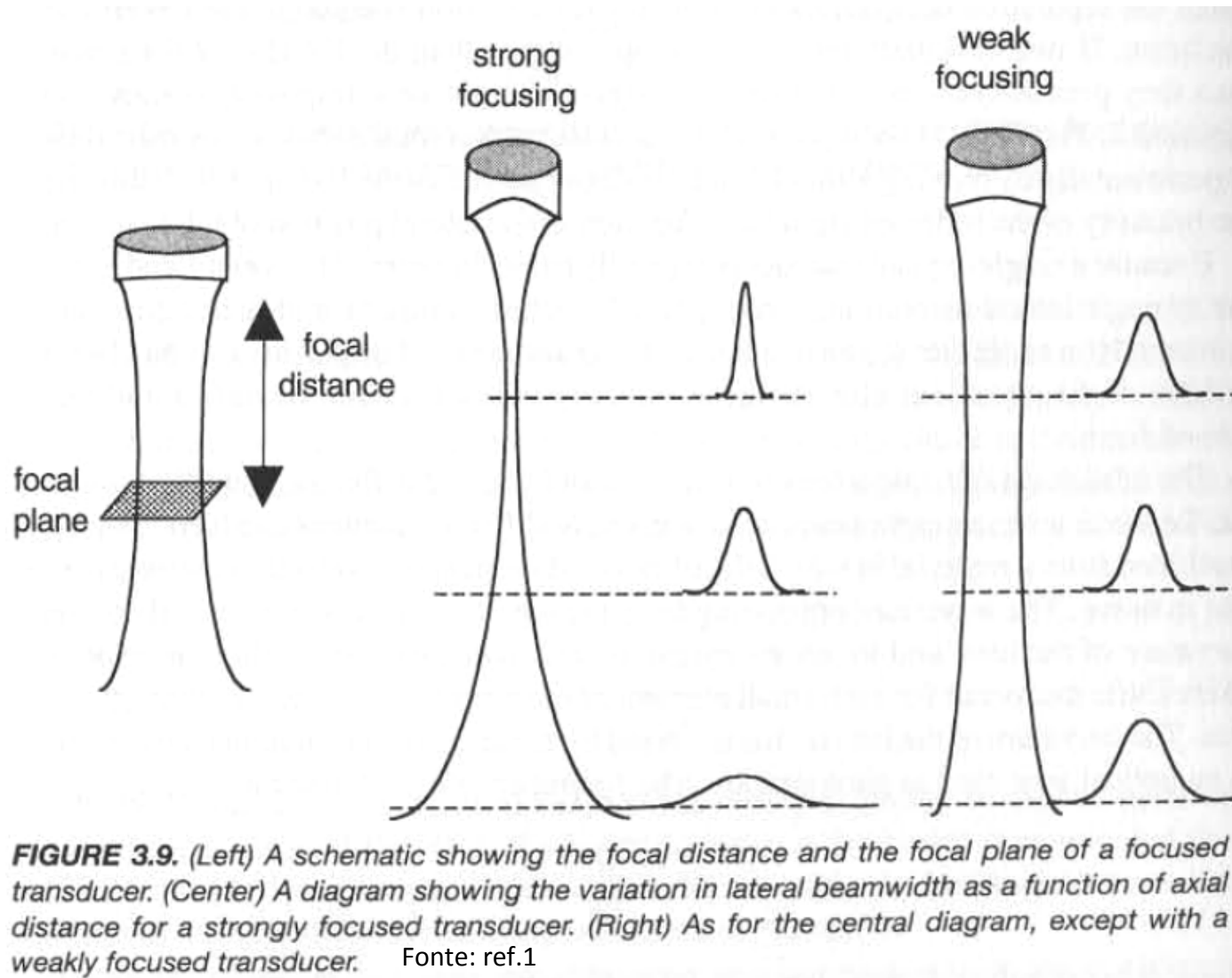


Fonte: ref.1

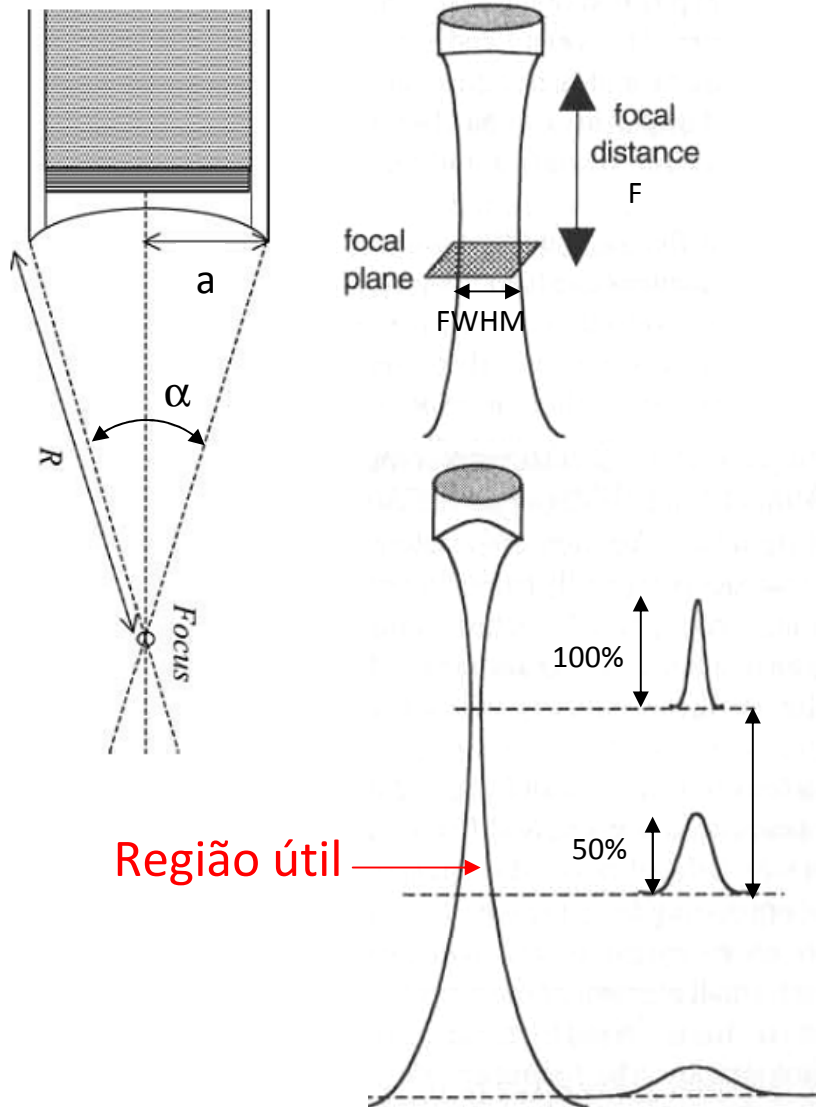
Campo acústico de um transdutor plano



Campo acústico de um transdutor focalizado por lente



Campo acústico de um transdutor focalizado por lente



Abertura numérica

$$f \# = \frac{R}{2a} \approx \tan(\alpha)$$

Distância focal

$$F = \frac{R}{1 - 1/f \#} > R$$

Resolução lateral

$$FWHM \approx \frac{1.1\lambda R}{2a}$$

Profundidade de campo (DOF)

$$DOF = 15(1 - 0.01 \arcsin(a/R)) FWHM$$

Campo acústico de um transdutor focalizado por lente

TABLE 3.2. Focusing Properties of Various Transducers at 1.5 and 5 MHz

Fonte: ref.1

Frequency (MHz)	λ In Tissue (mm)	R (cm)	a (cm)	f -Number	FWHM (mm)	DOF (mm)
1.5	1.0	2	1	1	1.1	11.9
1.5	1.0	3	1	1.5	1.7	20.5
1.5	1.0	5	1	2.5	2.8	37.5
1.5	1.0	10	1	5	5.6	79.8
5	0.31	2	2	1	0.3	3.6
5	0.31	3	2	1.5	0.5	6.1
5	0.31	5	2	2.5	0.8	11.2
5	0.31	10	2	5	1.7	24.0

Só é sensível numa linha de comprimento igual à DOF, que se forma a uma profundidade fixa:

- muito limitado (útil, por exemplo, para investigar batimentos cardíacos)
- varrimento apenas manual

Princípio do transdutor linear segmentado

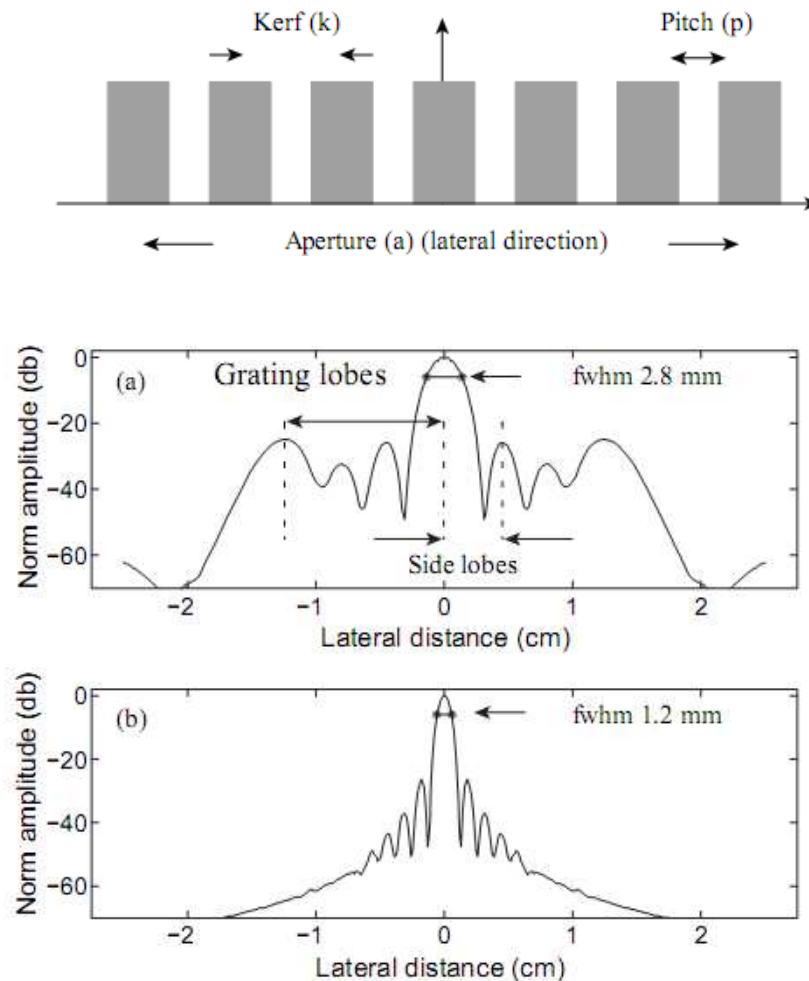


Figure 7. (a) Four elements ($200 \mu\text{m}$ kerf, $1000 \mu\text{m}$ pitch) using 4 mm aperture, yield a main sound lobe width of 2.8 mm fwhm. (b) Forty elements ($50 \mu\text{m}$ kerf, $250 \mu\text{m}$ pitch) using 1 cm aperture, cause the main lobe to narrow to 1.2 mm fwhm and the sidelobes and grating lobes to diminish. Half maximum corresponds to -6 dB. Fonte: ref.3

Ajustando os parâmetros mecânicos (passo, espaçamento) e eléctricos (fase) é possível gerar feixes estreitos a partir de um pequeno nº de cristais minúsculos.

Baseado inteiramente no fenómeno de interferência de ondas

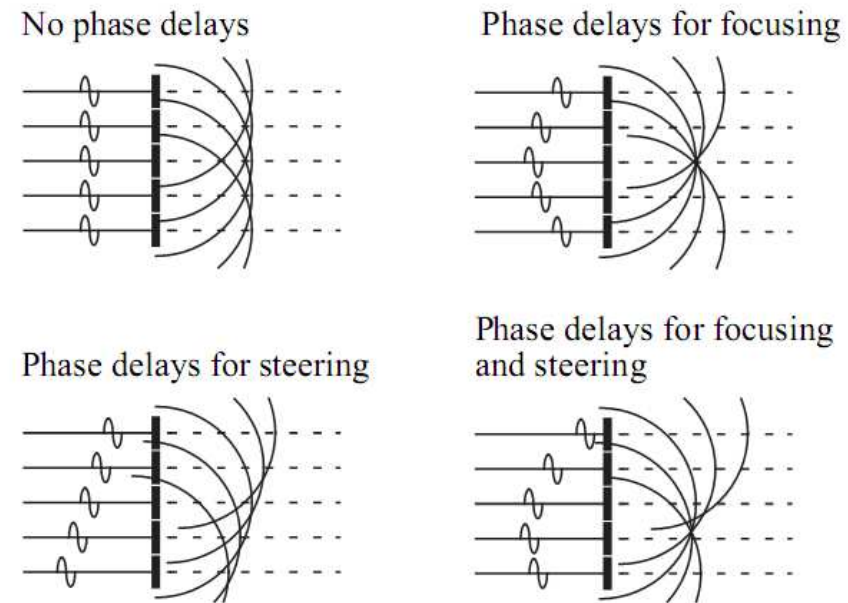
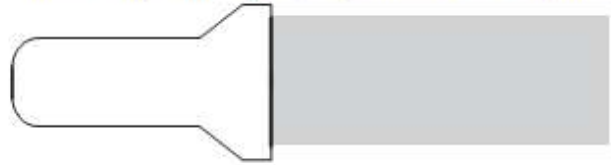


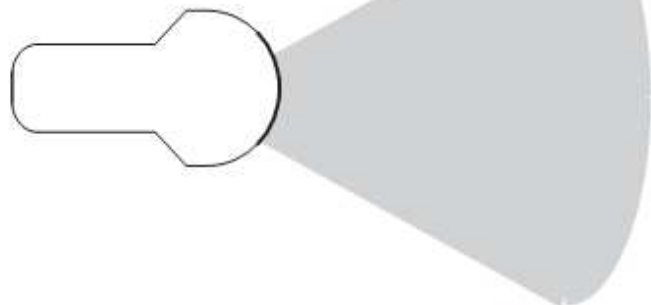
Figure 10. Delaying or advancing the phase of waves emitted from neighboring elements relative to the center element can achieve focusing and steering of the acoustic wave front. Fonte: ref.3

Configurações principais

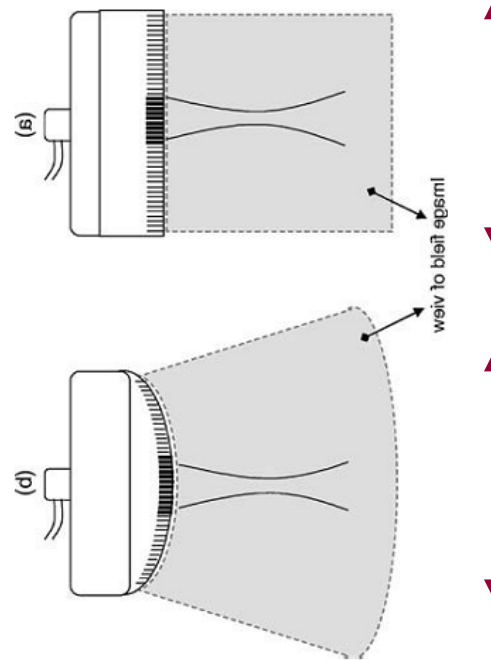
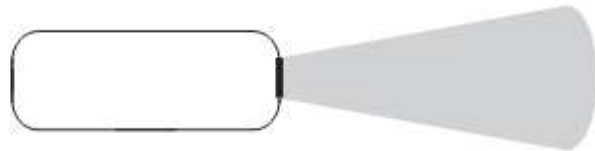
Linear array: small parts, superficial vascular, obstetrics



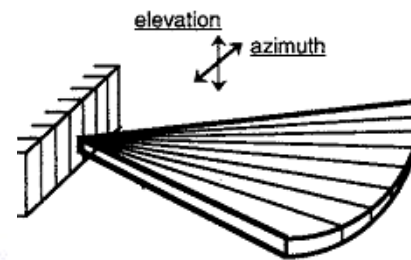
Curved array: abdominal, obstetric, transabdominal, or for transvaginal or transrectal, or pediatric imaging



Phased array: heart, liver, spleen, fontanelle, temple



Varrimento sequencial
“linear array”
(pequeno número de
cristais de cada vez com
fases fixas)



Varrimento faseado
“phased array”
(todos os cristais, com
fases variáveis)

Figure 9. Transducer geometries include (curved) linear arrays and phased arrays. Linear arrays transmit and receive acoustic beams perpendicular to their surface area, whereas phased arrays steer the acoustic beam by a phasing scheme (Fig. 10). Fonte: ref.3

Menor qualidade de imagem, mas mais
pequeno
(cabe entre as costelas ou na fontanela)

Configurações principais

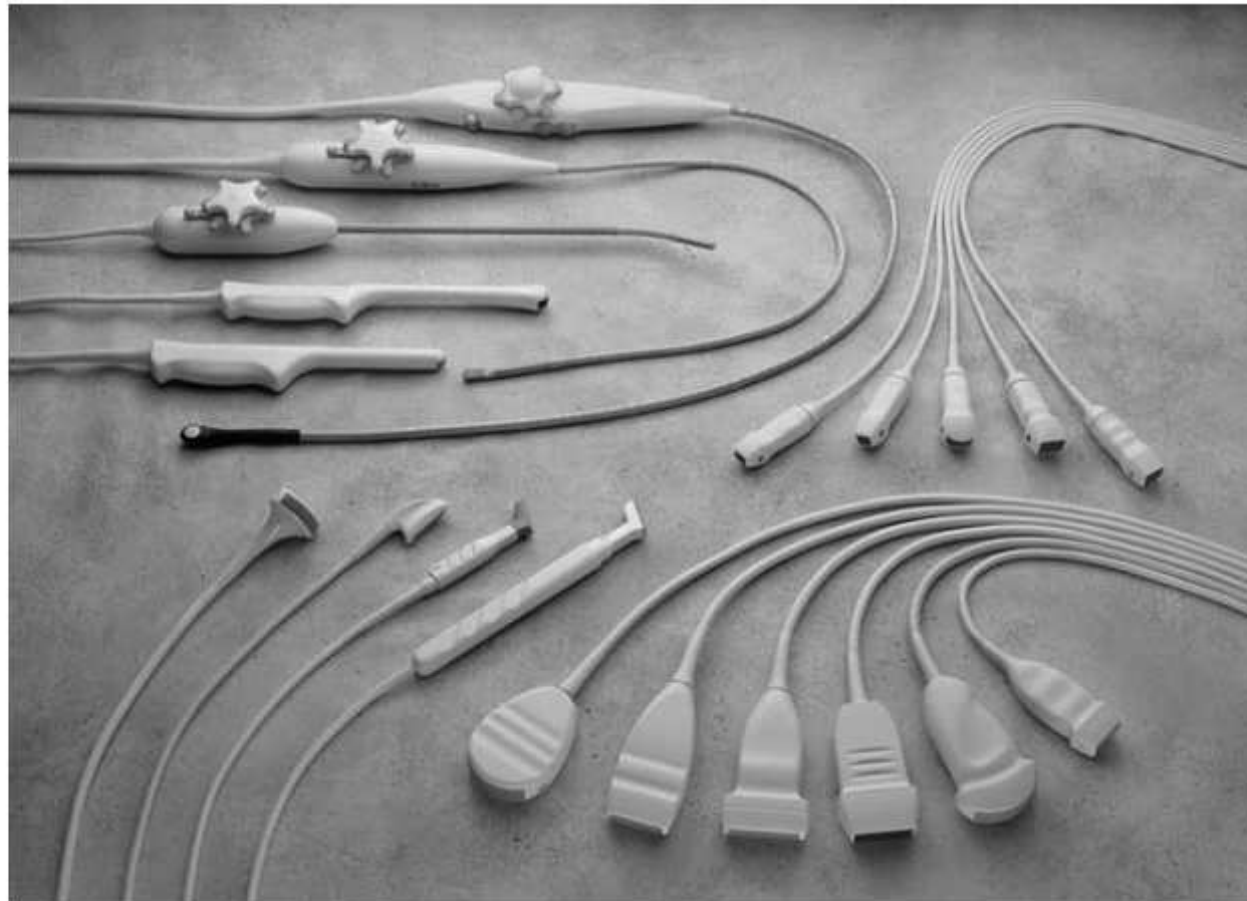


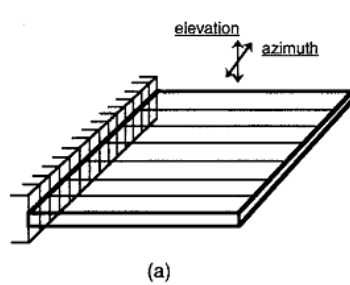
Figure 16.18 Transducers available on a modern imaging system are designed for a wide variety of clinical applications. Transducer groups: bottom right: linear and curved linear arrays; top right: phased arrays; left side: specialty probes including intraoperative, transesophageal, and transvaginal arrays (courtesy of Philips Medical Systems).

Fonte: ref.2

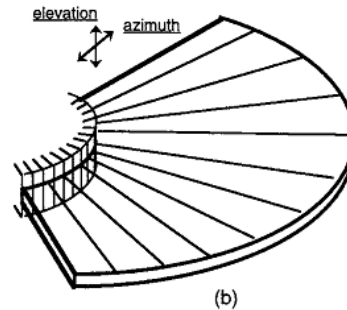
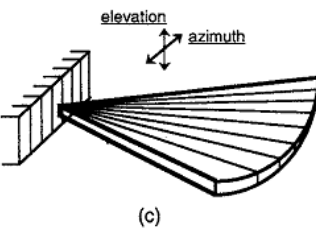
Configurações ...

Sequencial

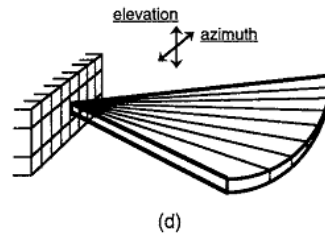
Linear



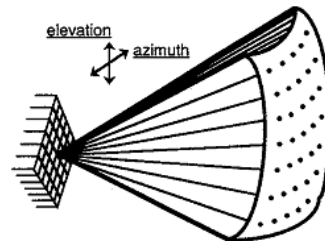
Sector

Faseado

1D



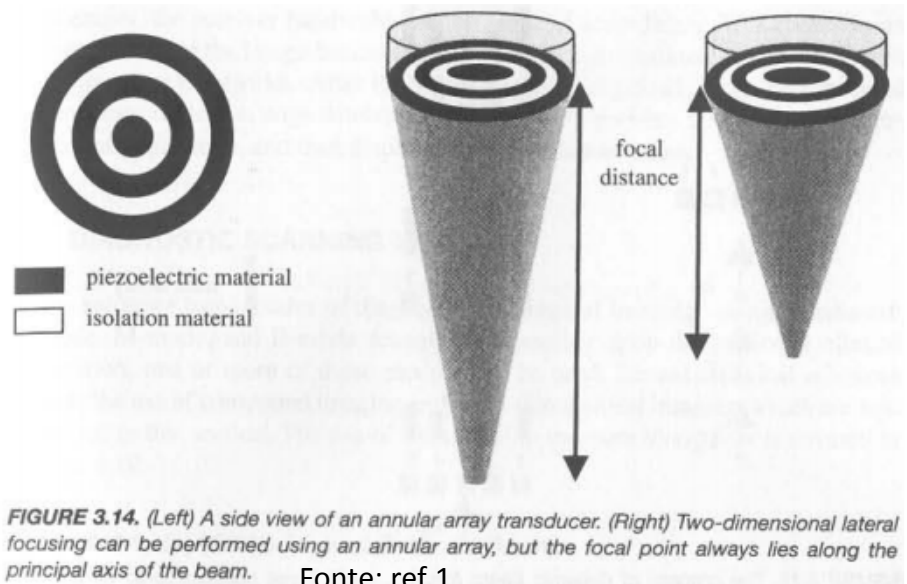
1.5D



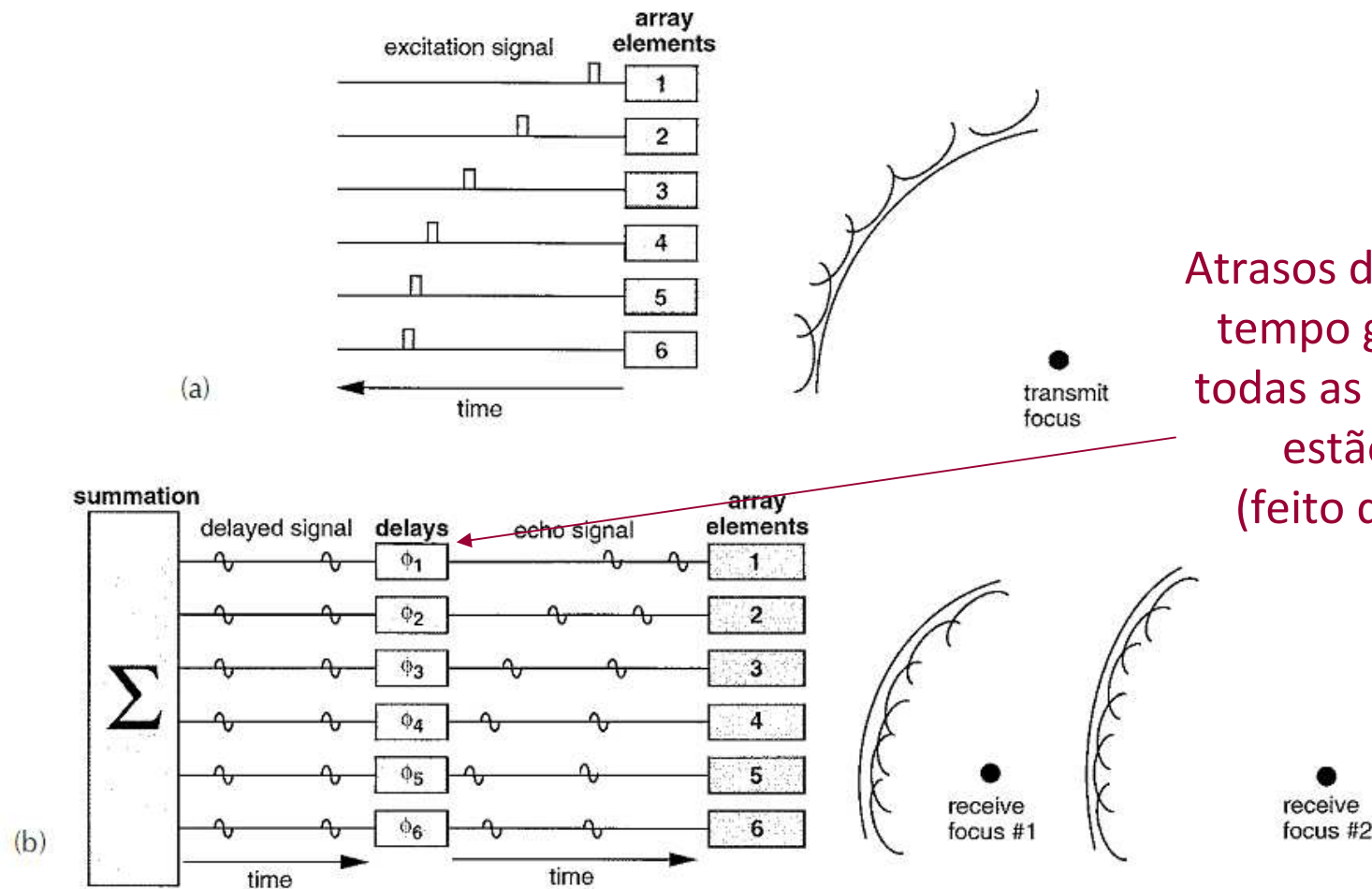
2D

FIGURE 65.2 Array-element configurations and the region scanned by the acoustic beam. (a) A sequential linear array scans a rectangular region; (b) a curvilinear array scans a sector-shaped region; (c) a linear phased array scans a sector-shaped region; (d) a 1.5D array scans a sector-shaped region; (e) a 2D array scans a pyramidal-shaped region.

Fonte: ref.8

Anular

Focalização dinâmica e "beamforming"



Atrasos dependentes do tempo garantem que todas as profundidades estão em foco. (feito digitalmente)

FIGURE 65.1 Focusing and steering an acoustic beam using a phased array. A 6-element linear array is shown (a) in the transmit mode and (b) in the receive mode. Dynamic focusing in receive allows the scanner focus to track the range of returning echoes. Fonte: ref.8

Compensação tempo-ganho

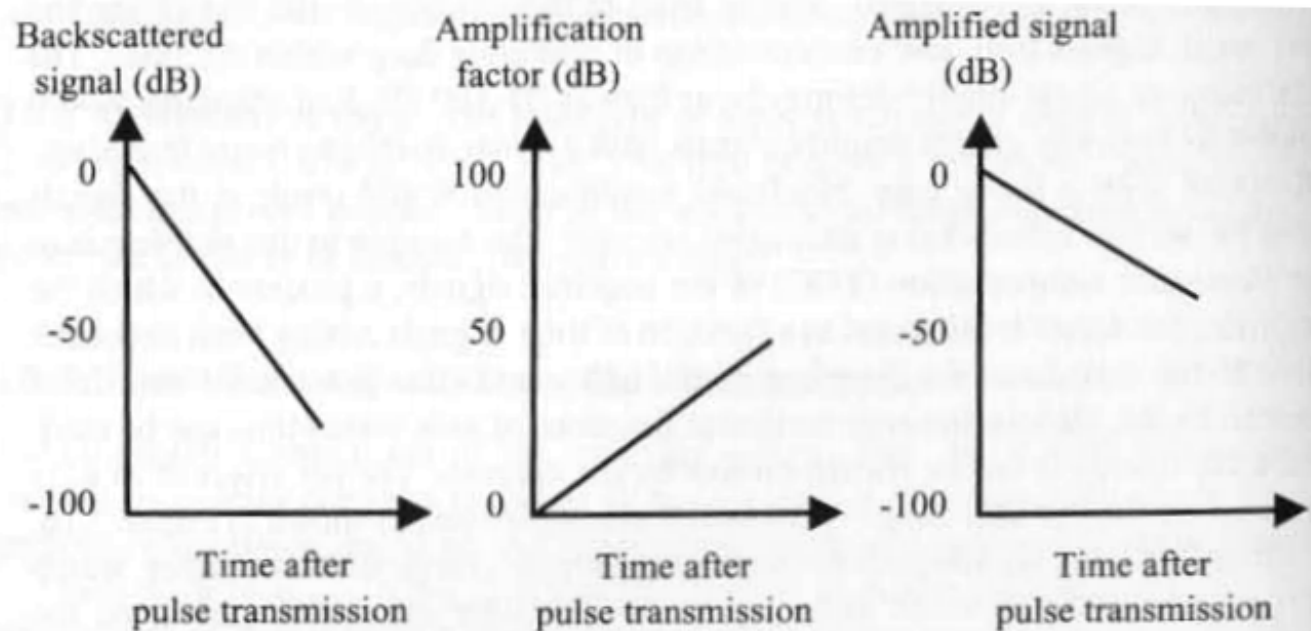


FIGURE 3.16. The principle of time–gain compensation. (Left) The dynamic range of backscattered echoes is larger than can be amplified linearly. (Center) A time-dependent amplification factor is applied to the backscattered echoes. (Right) The dynamic range of the amplified signals has been reduced to a level appropriate for further processing.

O ganho dos amplificadores tem de ser aumentado em função do tempo decorrido após a emissão do impulso, para compensar a atenuação do sinal a maiores profundidades

Modos A, B e M

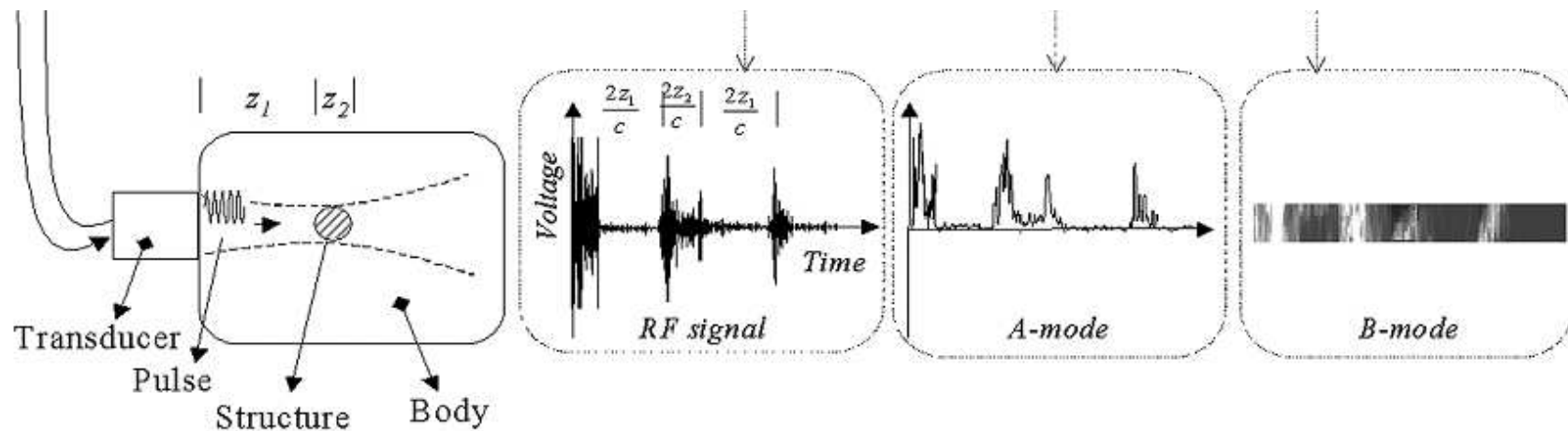


FIGURE 9.17 Block diagram of a pulsed-wave system and the resulting signal or image at three different steps. Fonte: ref.8

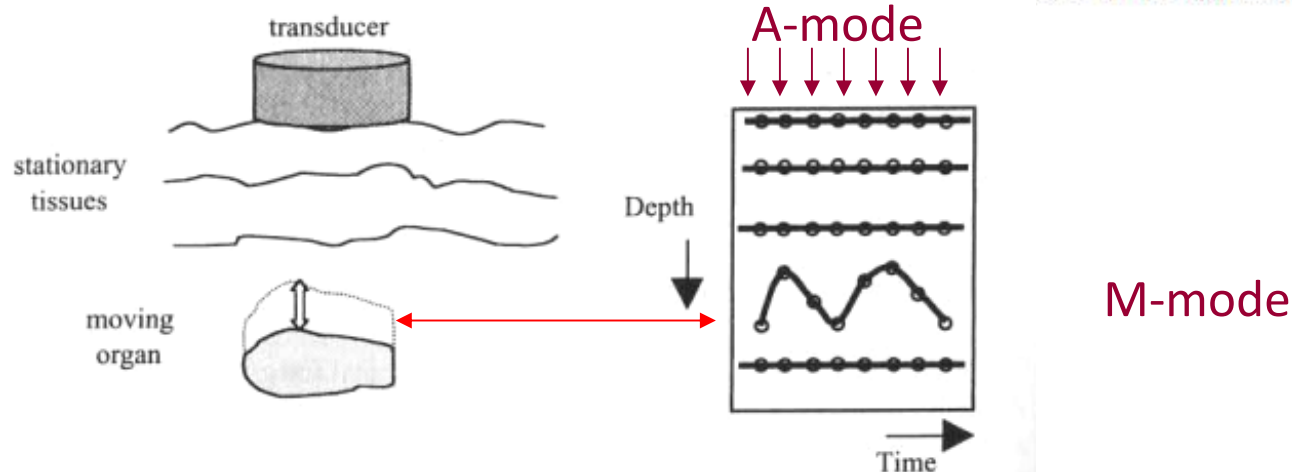
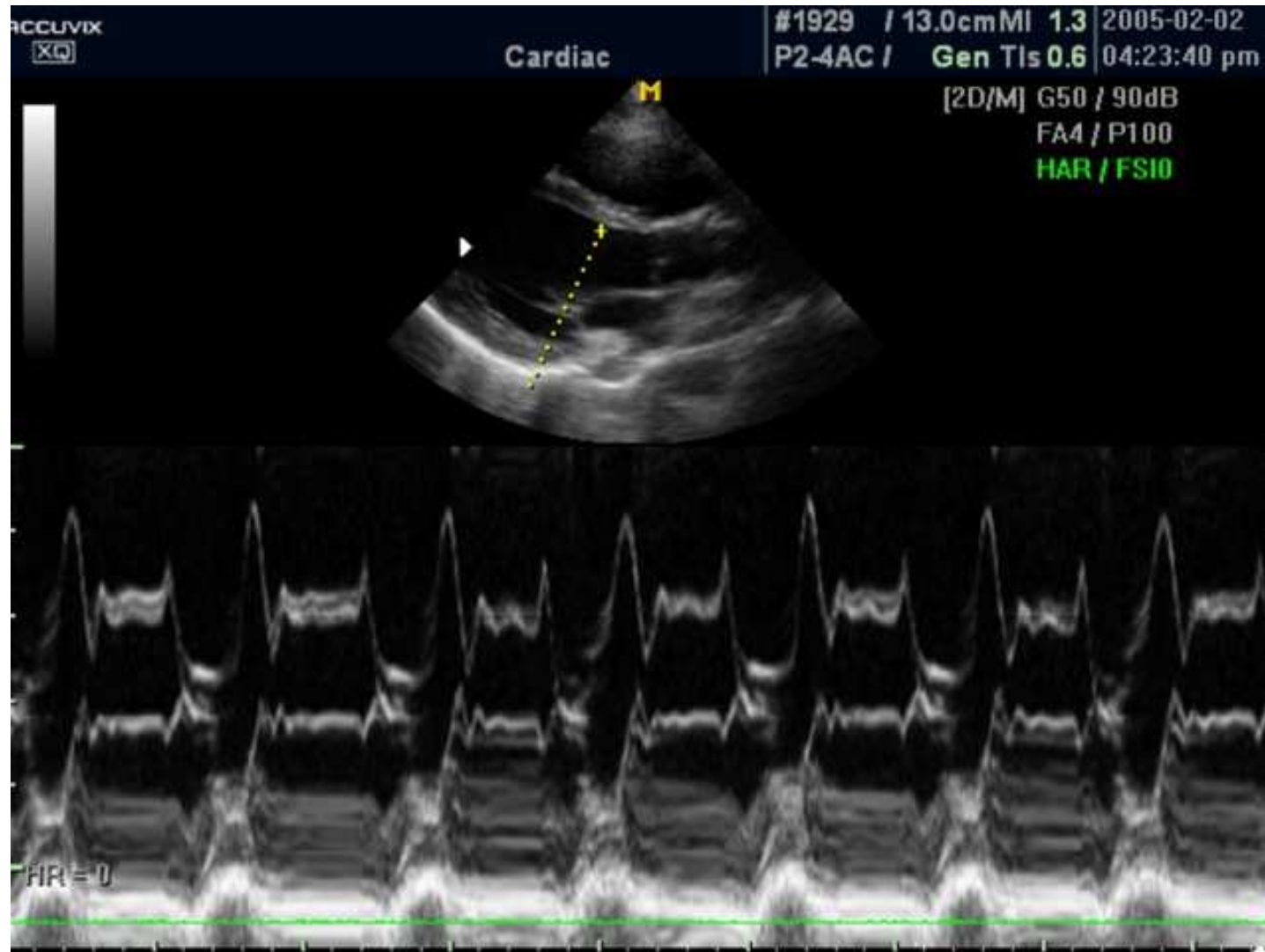


FIGURE 3.17. A schematic of M-mode ultrasound scanning. (Left) A transducer is placed over the area of interest, which consists of several stationary tissues and one that is moving periodically. (Right) The M-mode scan comprises a time series of A-mode scans, allowing the degree of movement of the individual tissues to be seen. Fonte: ref.1

B e M duplex

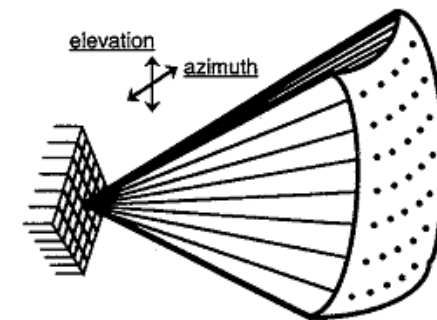
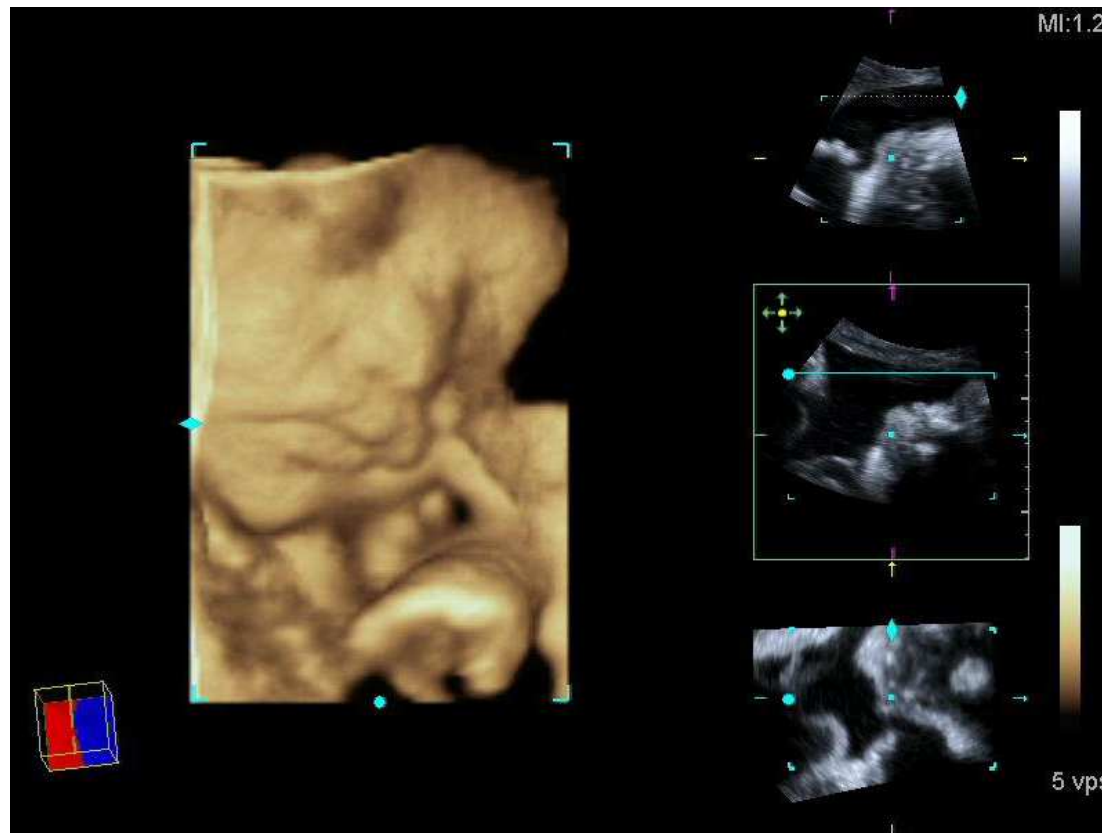


Modo B

Modo M

Modos de apresentação

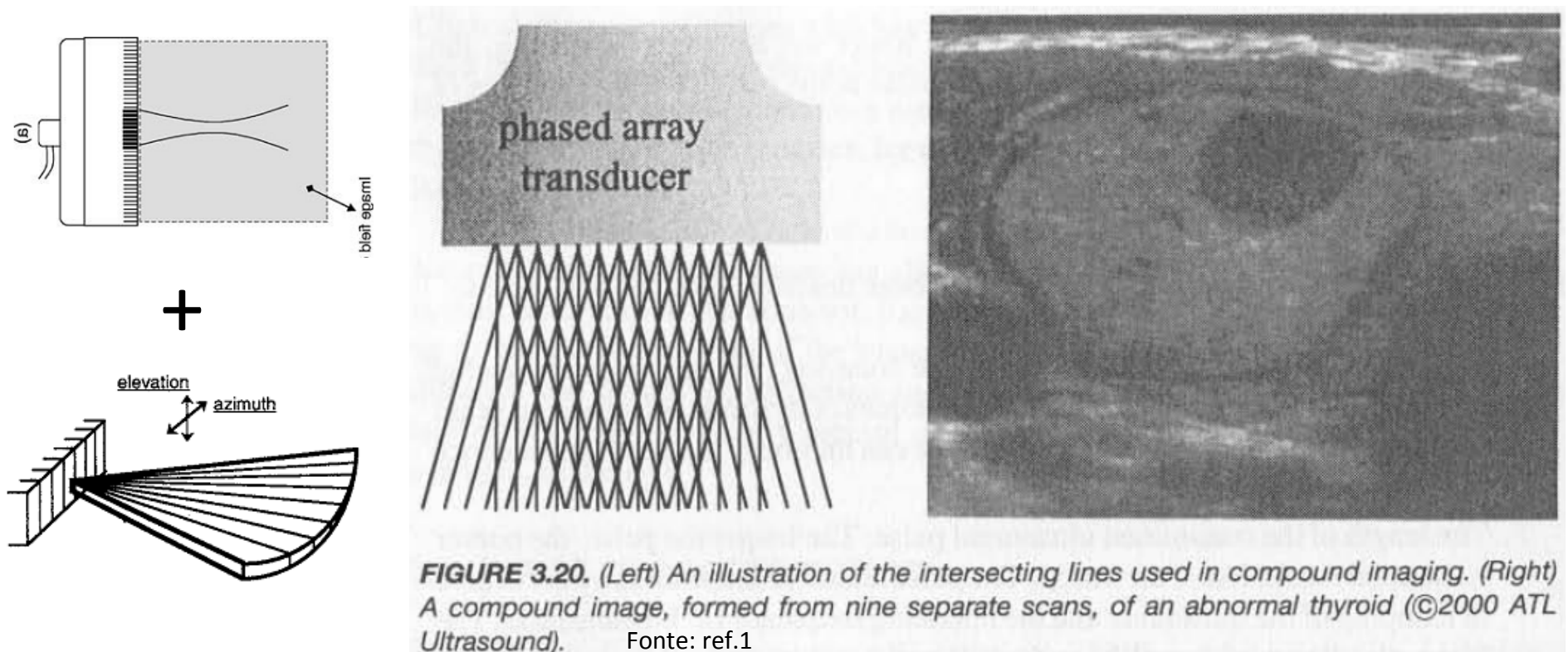
3D!



Fonte: www.siemensultrasound.com

SonoCT ("compound imaging")

O objecto é explorado de múltiplos ângulos a partir dum transdutor linear em modo de varrimento faseado



Muito melhor relação sinal-ruído

Factores que afectam a relação sinal-ruído

Fontes de ruído

- Ruído electrónico dos amplificadores
- Ruído especular (“speckle”): interferência entre os diversos ecos nos tecidos
- Ruído secundário (“clutter”): ecos dos lobos de emissão laterais, reflexões secundárias, movimento dos tecidos, etc

Factores que aumentam o sinal

- Intensidade do sinal produzido pelo transdutor
- Menor frequência do sinal (menor atenuação nos tecidos, menor resolução)
- Grau de focagem (pode melhorar um ponto em detrimento dos restantes)
- Menor amortecimento do transdutor (menor resolução axial)

Factores que afectam a resolução

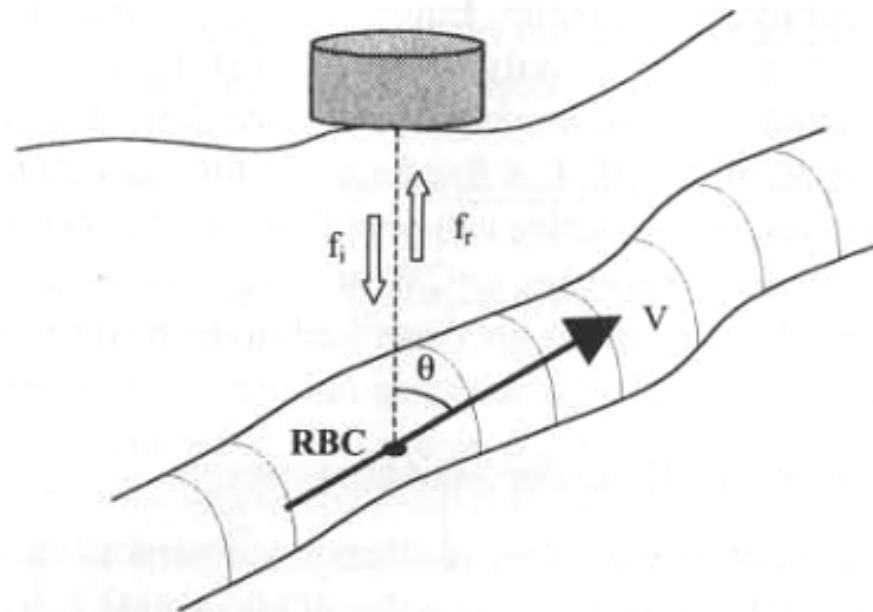
Grau de focagem afecta a resolução lateral

- Com focagem dinâmica e beamforming é possível minimizar a dependência da resolução com a profundidade

Duração do impulso afecta a resolução axial

- Impulsos mais curtos dependem de utilizar frequências mais altas e de um amortecimento adequado do transdutor. Ambos os casos diminuem o sinal disponível.

Princípio do Doppler de onda contínua



Fonte: ref.1

FIGURE 3.21. An illustration of the Doppler shift in flowing blood. The RBCs scattering the ultrasound beam are traveling at velocity v . Here f_i is the ultrasound frequency transmitted by the transducer and f_r is the frequency of the backscattered echo. The difference in these frequencies is the measured Doppler shift.

$$\frac{\Delta f}{f_i} = \frac{f_r - f_i}{f_i} = \frac{2v \cos \theta}{c}$$

Exemplo:
 $v=50 \text{ cm/s}$, $\theta=45^\circ$
 $\Delta f/f=5 \times 10^{-4}$

Doppler de onda contínua

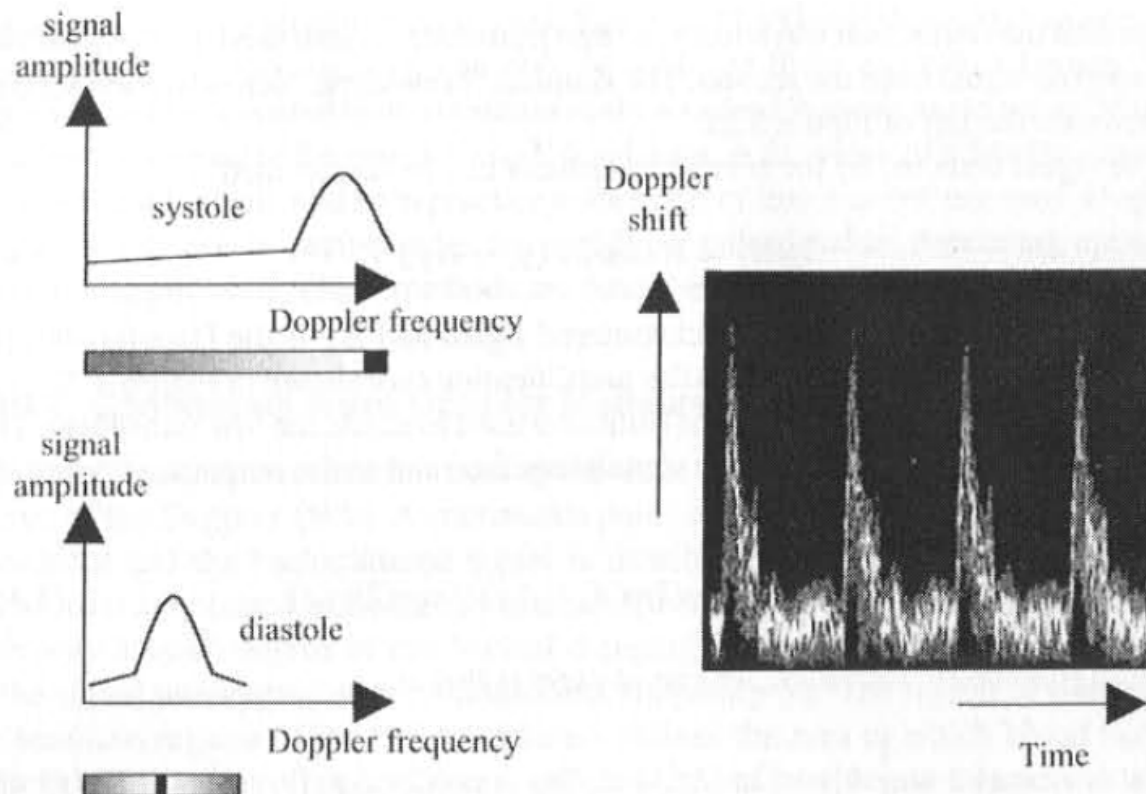


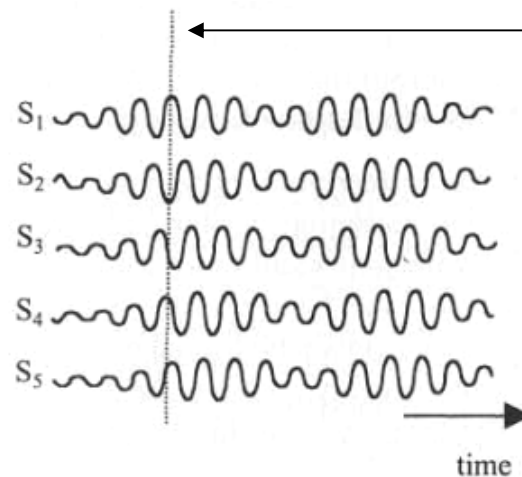
FIGURE 3.23. Spectral Doppler shifts from an area encompassing the carotid artery. Large frequency shifts, corresponding to high blood flow rates, are measured during the systolic part of the heart cycle (top left), with lower frequency shifts measured during diastole (bottom left). At each time point, the two-dimensional amplitude versus Doppler frequency graphs are reduced to a single "spectral dimension," with the length of the "bar" representing the total range of frequencies present and the color representing the amplitude at each frequency (gray to white representing low to high amplitudes, respectively). (Right) A two-dimensional display of spectral Doppler plots (vertical axis) as a function of time over several cardiac cycles. Fonte: ref.1

Principal problema:
as variações de frequência
ocorrem em qualquer
ponto ao longo de uma
linha.

Não há sensibilidade à
profundidade.

Doppler pulsado

Para haver sensibilidade à profundidade é necessário usar impulsos. Mas os impulsos não têm uma frequência bem definida, o que impede a medida das pequeníssimas variações de frequência.



Tempo correspondente à profundidade que se quer medir.

Utiliza-se o facto de que o efeito doppler altera também a fase do sinal. Esta é medida num trem de impulsos.

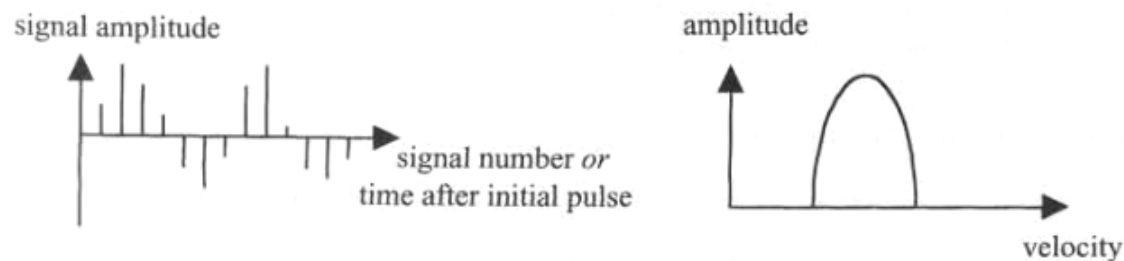


FIGURE 3.24. The basic processing steps in pulsed-mode Doppler ultrasound. (Top) After each pulse in a pulse train a backscattered signal (S_1, S_2, \dots, S_5) is recorded. (Bottom left) The signal amplitude at a particular depth, corresponding to the dotted line in the top figure, is plotted as a function of the time after the initial pulse. (Bottom right) Fourier transformation of this plot results in the Doppler frequency, and hence blood velocity, distribution at the chosen location. Fonte: ref.1

Métodos de correlação

Utiliza-se o facto de que um objecto que se move a certa velocidade, muda de posição no espaço.

Comparando a forma dos ecos em diferentes medidas separadas por algum tempo é possível determinar a diferença de posições do objecto e logo a sua velocidade.

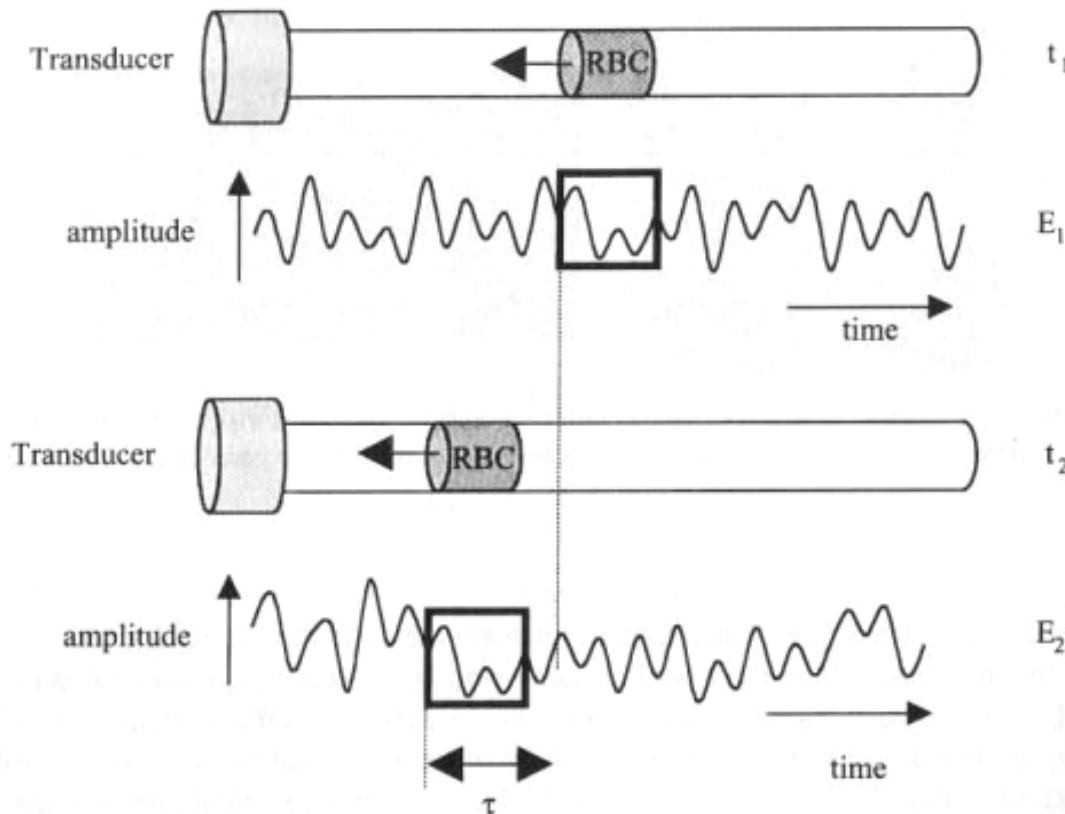


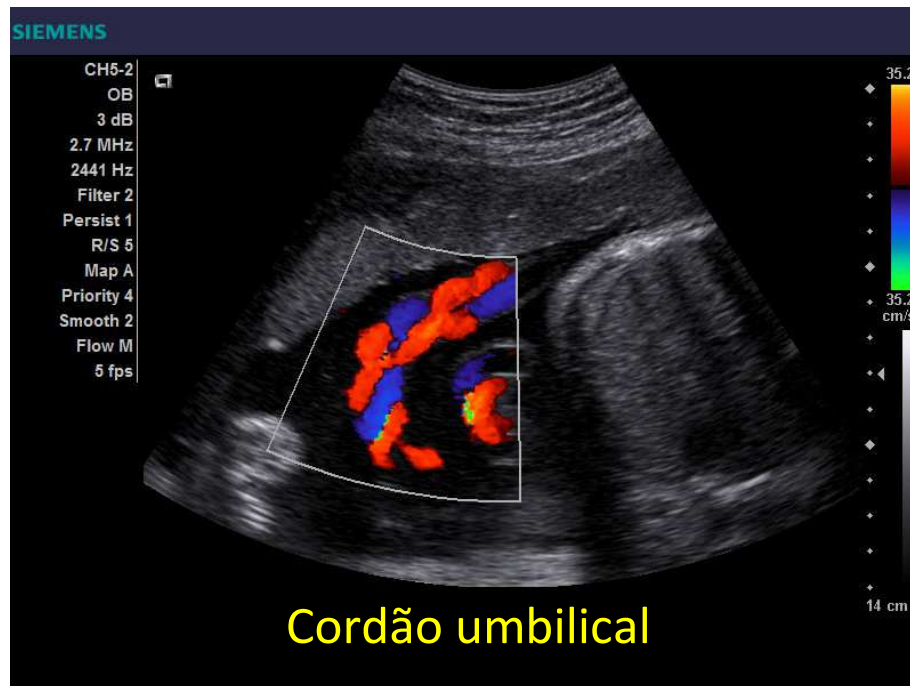
FIGURE 3.26. A diagram showing the basis of time-domain correlation methods for measuring blood velocity. A pulse of ultrasound is transmitted at time t_1 and the backscattered echo E_1 recorded. A second pulse is transmitted at time t_2 and the signature signal from the particular group of RBCs is time-shifted by an amount τ in the corresponding echo E_2 . Correlation methods, as described in the text, are used to estimate the value of τ and hence the blood velocity.

Fonte: ref.1

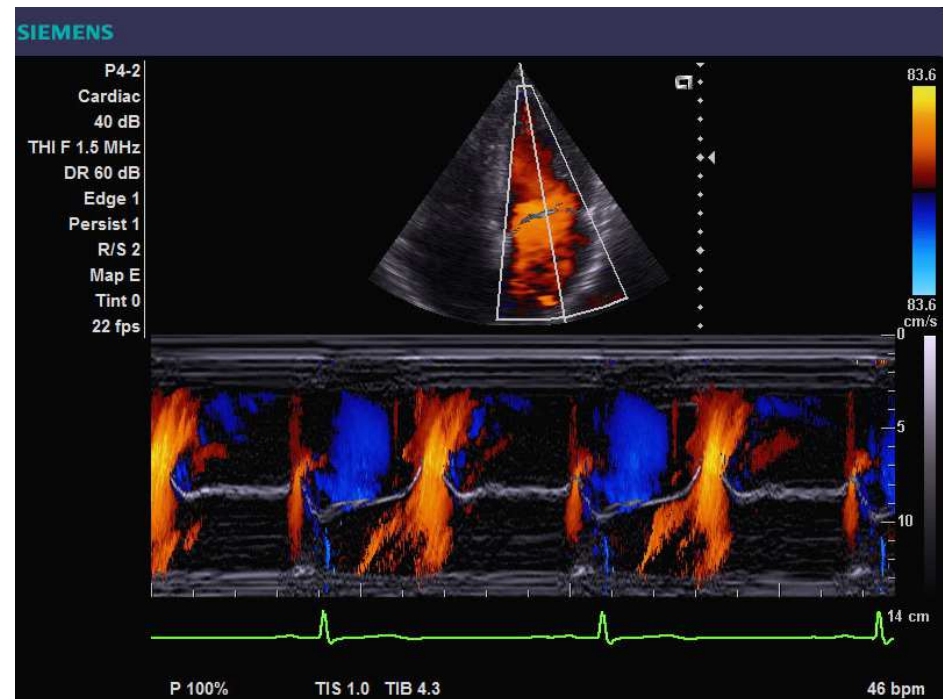
$$v = \frac{1}{2} \frac{c\tau}{t_2 - t_1}$$

Modos de apresentação

Color-flow Doppler



M-Doppler



Fonte: www.siemensultrasound.com

Agentes contrastantes para o sangue

Table 3. Modern Ultrasound Contrast Agents^a Fonte: ref.3

Manufacturer	Agent Name	Interior Gas	Shell Material
Acusphere	AI-700	Perfluorocarbon	Copolymers
Alliance Pharma. / Photogen	Imavist	Perfluorohexane, air	Surfactant
Bracco	SonoVue	Sulfur hexafluoride	Phospholipid
Bristol-Myers Squibb Medical Imaging, Inc.	Definity	Perfluoropropane	Lipid bilayer
	MRX-815-stroke	Perfluoropropane	Lipid bilayer
Molecular Biosys.	Albunex	Air	Albumin
	Oralex	Air	Dextrose
GE/ Amersham	Optison	Perflutren	Albumin
Nycomed Imaging AS	Sonazoid	Perfluorobutane	Lipid
Schering AG	Echovist	Air	Galactose
	Levovist	Air	Lipid layer
Sonus Pharma.	EchoGen	Dodecafluoropentane	Albumin
	SonoGen		Charged surfactant

^aSee Refs. 3,7, and 9.

Microesferas de 1 a 10 μm cheias de gás têm uma frequência natural de ressonância semelhante às frequências usadas para ecografia e difundem eficientemente as ondas sonoras.

Imagiologia de harmônicos com agentes contrastantes

Os agentes contrastantes têm um comportamento não linear e alteram o timbre do som (criam frequências que não existiam no sinal injectado). Para estas frequências a relação sinal-ruído é muito melhor.

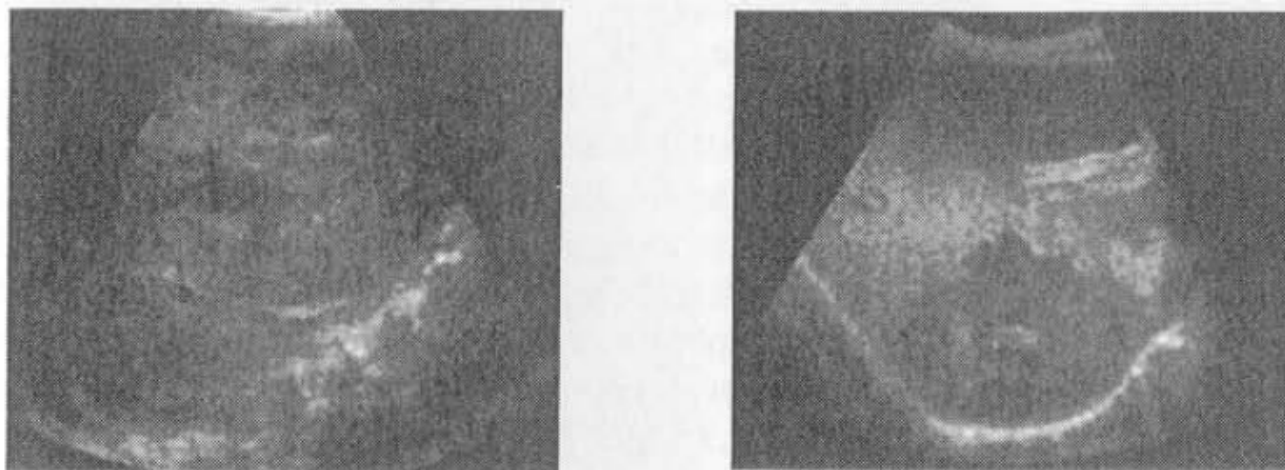


FIGURE 3.27. (Left) A fundamental mode image of a liver mass using a contrast agent. (Right) The corresponding second harmonic image, showing much clearer delineation of the tumor mass (©2000 ATL Ultrasound).

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