Diagnostic Ultrasound in Small Animal practice

Paddy Mannion

BVMS, DVR, MRCVS, Diplomate ECVDI

Blackwell
Science

Diagnostic Ultrasound in Small Animal practice

Paddy Mannion

BVMS, DVR, MRCVS, Diplomate ECVDI

Blackwell
Science

Editorial offices:
Blackwell Science Ltd, 9600 Garsington Road, Oxford OX4 2DQ, UK
Tel: +44 (0) 1865 776868
Blackwell Publishing Professional, 2121 State Avenue, Ames,
Iowa 50014-8300, USA
Tel: +1 515 292 0140
Blackwell Science Asia Pty, 550 Swanston Street, Carlton,
Victoria 3053, Australia
Tel: +61 (0)3 8359 1011

First published 2006

ISBN-10: 0-632-05387-9
ISBN-13: 978-0-632-05387-2

Library of Congress Cataloging-in-Publication Data

Mannion, Paddy.
Diagnostic ultrasound in small animal practice / Paddy Mannion.
p. cm.
Includes bibliographical references and index.
ISBN-13: 978-0-632-05387-2 (pbk. : alk. paper)
ISBN-10: 0-632-05387-9 (pbk. : alk. paper)
1. Veterinary ultrasonography. I. Title.

SF772.58.M36 2005
636.089'607543–dc22 2005008357

A catalogue record for this title is available from the British Library

Set in 10/12.5pt Plantin
by Graphicraft Limited, Hong Kong
Printed and bound in the UK
by TJ International, Padstow, Cornwall

The publisher's policy is to use permanent paper from mills that operate a sustainable forestry policy, and which has been manufactured from pulp processed using acid-free and elementary chlorine-free practices. Furthermore, the publisher ensures that the text paper and cover board used have met acceptable environmental accreditation standards.

For further information on Blackwell Publishing, visit our website:
www.blackwellpublishing.com

Contents

Foreword

It gives me pleasure to write the foreword to this new version of Diagnostic Ultrasound in the Dog and Cat. Diagnostic ultrasound has advanced in leaps and bounds since the book was first published in 1990. There have been major technological advances in transducer design and image acquisition and processing, resulting in images of vastly improved resolution. Despite this, the cost of equipment has come down in relative terms, and is now a viable option for many small animal practices. Owners are increasingly well informed and are often aware of ultrasound as a safe and informative diagnostic procedure. Taking all these factors into account, more veterinary surgeons than ever before are aiming to develop or update their expertise in diagnostic ultrasound.

Several authors have been involved in the writing of this new book, reflecting perhaps the expansion of the subject and the specialist expertise which an individual is now able to develop in given areas. The authors are all very well known in the field of small animal diagnostic ultrasound, and will be bringing to the reader a wealth of practical experience as well as important theoretical knowledge; both aspects are vital in acquiring the best possible images and in reaching pertinent and accurate conclusions. This book will be valuable to those who are just beginning to develop their skills with diagnostic ultrasound, as well as to those who wish to update and extend their knowledge. I am delighted to be able to recommend it to you.

Frances Barr

Contributor List

Paddy Mannion	BVMS DVR MRCVS Dip ECVDI Cambridge Radiology Referrals RCVS Recognised Specialist in Veterinary Diagnostic Imaging European Specialist in Veterinary Diagnostic Imaging
Mairi Frame	BVMS DVR MRCVS Dip ECVDI Lecturer in Veterinary Diagnostic Imaging, Edinburgh University Vet School RCVS Diplomate in Veterinary Diagnostic Imaging European Specialist in Veterinary Diagnostic Imaging
Sharon Redrobe	BSc(Hons) BVetMed CertLAS DZooMed MRCVS RCVS Recognised Specialist in Zoo and Wildlife Medicine
Alison King	BVMS, MVM, DVR, DipECVDI, MRCVS RCVS Recognised Specialist in Veterinary Diagnostic Imaging European Specialist in Veterinary Diagnostic Imaging Lecturer in Veterinary Clinical Anatomy, Glasgow University Veterinary School
Johann Lang, Prof	Dr.med.vet., Dip ECVDI European Specialist in Veterinary Diagnostic Imaging Head, Division of Clinical Radiology Department of Clinical Veterinary Medicine University of Bern
Valerie Schmidt	DrMedVet Cert VR Dip ECVDI European Specialist in Veterinary Diagnostic Imaging

Chapter 1

Principles of Diagnostic Ultrasound

Paddy Mannion

Diagnostic ultrasound uses high frequency sound waves to produce an image of the body. Sound waves with a frequency greater than 20 KHz are classed as high frequency as these are outside the range of human hearing. For diagnostic purposes the frequency of the sound used is typically in the range of 2–10 MHz but as technology advances, higher frequency ultrasound is being used diagnostically and in some centres frequencies of 15 MHz and above are being used for high resolution work.

Sound waves

Sound energy is mechanical energy which means that it requires a medium for propagation as it produces physical movement of the molecules and particles within the material through which it travels. Sound waves are longitudinal waves in which the direction of travel of particles within the wave is the same as that of the wave itself. Each wave has cycles of compression and rarefaction (Figure 1.1). Each wave has an associated speed of travel, a wavelength and frequency. The wavelength is the distance travelled in one cycle, which is the distance between the same point in successive areas of compression or rarefaction. Frequency is the number of cycles per second and speed is the distance travelled in a particular time, usually one second. The relationship between these factors is shown in the following equation:

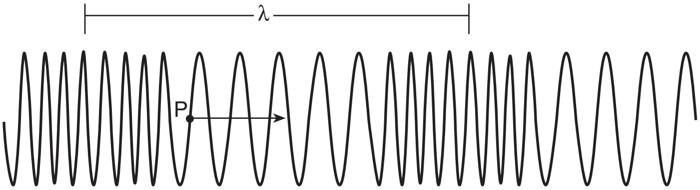


Figure 1.1 Diagram of a longitudinal wave – note the cycles of compression and rarefaction. The wavelength (λ) is shown as the distance between the centre point of successive areas of compression or rarefaction. The direction of motion of the particle (P) is the same as the direction of the wave itself.

$$v = f \times \lambda \quad > \quad f \; \alpha \; 1/\lambda$$

v = velocity   f = frequency λ = wavelength

1 cycle/second = 1 Hz
1000 cycles/second = 1 KHz
1 000 000 cycles/second = 1 MHz

In general, the speed of travel of sound through soft tissues is taken as a fairly constant value (approximately 1540 m/s) and so wavelength and frequency are inversely related. This is fundamental to the practical application of ultrasound. Essentially, the higher the frequency of the sound waves produced, the shorter the wavelength of the sound. Examples of how this relates to commonly used frequencies are shown below.

Frequency (MHz)	Wavelength (mm)
2	0.77
5	0.31
7.5	0.21

The speed of travel of sound through different tissues varies according to their individual properties, in particular their density. Therefore, while the average speed of sound in soft tissue is 1540 m/s, through bone it is 4000 m/s and through gas it is only 300 m/s.

Material	Speed of Travel
Bone	4080 m/s
Blood	1570 m/s
Liver	1560 m/s
Fat	1440 m/s
Air	330 m/s

These differences are important as will be seen later. It is clear therefore that the transmission of sound relies on the structure of the medium and the denser the medium the faster the transmission.

Acoustic impedance

Each tissue has inherent acoustic impedance, which is essentially the resistance to the transmission of the sound wave within the material.

Acoustic impedance = Density × Speed

$$\mathbf{Z} = v \times \rho$$

Z = acoustic impedance v = velocity of sound
ρ = density of material

Acoustic impedance is important in its own right, but the difference in acoustic impedance between adjacent tissues is especially important. Where there is a great difference in acoustic impedance of adjacent tissues there is greater reflection of the sound waves from the interface of the tissues. At the interface between soft tissue and bone there is reflection of almost 50% of the ultrasound beam, whereas at the interface between soft tissue and gas this increases to almost 99%. This is very important for successful application of ultrasound as an imaging tool and interpretation of the resulting image as shown in Figures 1.2 and 1.3.

Ultrasound and tissue

The fundamental principle of diagnostic ultrasound is that sound waves pass through the tissues and are either reflected, refracted or absorbed. The sound waves which return to the transducer are responsible for producing the image. The greater the amount of sound which travels back to the transducer the brighter the image which is displayed on the screen (when using B-mode ultrasound). It is important to understand what governs the interaction between ultrasound and tissue in order to be able to interpret the image correctly. These three processes, reflection, refraction and absorption are quite different but are related.

Reflection is responsible for producing the image, as it is the reflected ultrasound waves which are transformed into the image when they reach the transducer. Reflection depends on the size of the reflecting structure and also the frequency of the sound waves in question. Higher frequency sound waves are reflected from smaller structures and are attenuated more quickly so that higher frequency sound waves are used when imaging more superficial structures. When there is a difference in acoustic impedance as the waves travel from one tissue to another there is a greater amount of reflection, as discussed above, and fewer waves remain to pass through into the deeper tissues. This explains the need for good coupling between the transducer and the skin surface.

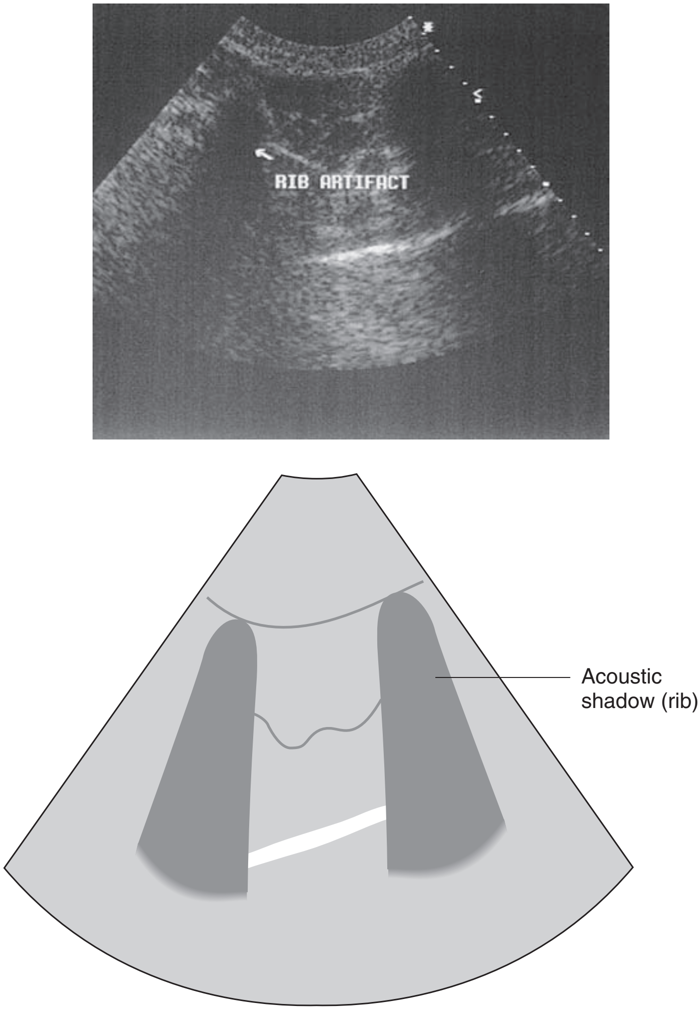
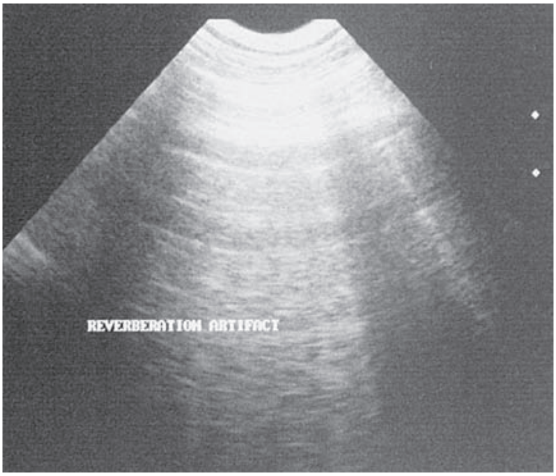


Figure 1.2 Thoracic ultrasound in a dog showing rib artefact. This image illustrates the result of 50% reflection of the ultrasound beam at the soft tissue–bone interface and absorption of the remainder of the beam within the dense bone. As a result no information is gained beyond the surface as no ultrasound passes deep to the bone. An acoustic shadow results.

Reflection is quite straightforward when applied to the ultrasound beam, which is perpendicular to the skin surface. Where the incident ultrasound beam is not perpendicular, the reflected ultrasound beam has an angle equal to the angle of incidence, provided the speed of travel within the tissues is equal (Figure 1.4). If the speed of travel in the two tissues is different refraction occurs. Reflection from a large smooth interface with dimensions much greater than the ultrasonic wavelength, is known as specular reflection. Very often though, reflection occurs from surfaces which



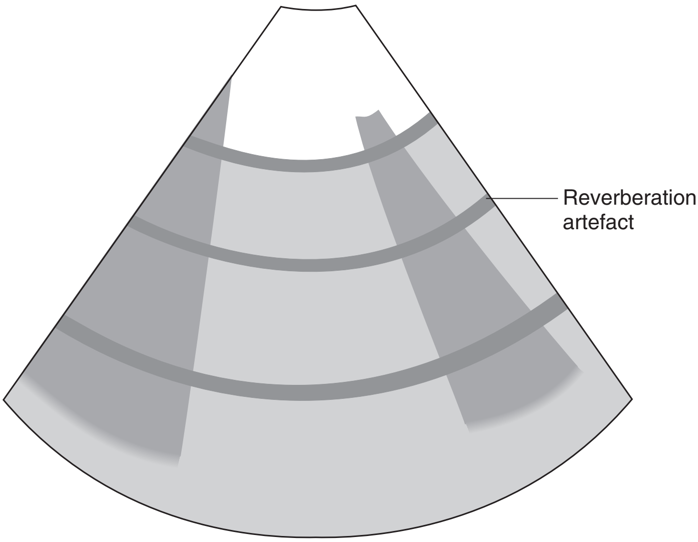


Figure 1.3 Thoracic ultrasound showing reverberation artefact in a dog. This image displays the effect of 99% reflection of the ultrasound beam at the soft tissue–gas interface. This poses a problem wherever there is gas.

are not completely smooth and are in the order of size of the ultrasound wavelength. These are known as diffuse reflectors as they travel in many directions and have low amplitude. This is advantageous in a way since although diffuse reflectors are weaker, they are less dependent on incident angle than specular reflectors and are used heavily to provide information on texture of the organ. Changes in scatter from one area to another are responsible for changes in brightness which are referred to as hyperechoic and hypoechoic. A hyperechoic appearance results from an increase

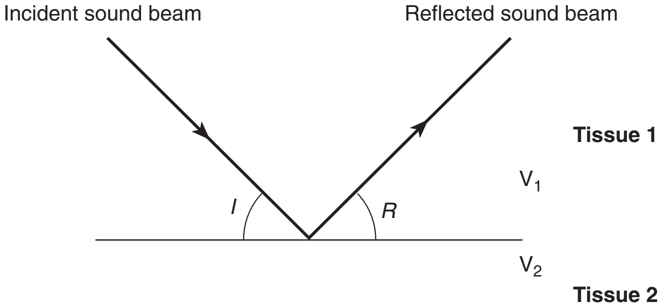


Figure 1.4 Where the incident ultrasound beam is not perpendicular to the interface between two structures the angle of the reflected beam (R) is equal to that of the incident beam (I), provided the speed of travel within the two tissues is the same. (V_1 = speed of sound through tissue 1, V_2 = speed of sound through tissue 2.) This is known as Snell's law.

in scattering when compared with the surrounding tissue, whereas a hypoechoic appearance results where there is a reduction in the scattering compared with the surrounding material.

Refraction is a change in direction of the sound waves as they pass from one medium into another, where the speed of travel is slightly different and occurs if the incident sound waves are oblique. Typically, this occurs where there is a fluid-filled viscus within a more solid structure, such as at the edges of the gall bladder (Figure 1.5). This is more pronounced in materials with greater acoustic impedance. As the refracted beam is travelling in a slightly different direction, the angle of reflection is also different and so the position of the imaged structure may differ from the real structure. This can produce confusing artefacts.

Where the scattering structures are much smaller than the wavelength of the incident beam and they are numerous these are known as Rayleigh scatterers. An example of this would be red blood cells. Scattering from such structures is proportional to frequency raised to the power of four so that doubling the frequency increases scattering by a factor of 16.

Attenuation is reduction in the intensity of the ultrasound beam as it passes through the tissue and occurs due to two processes; Rayleigh scatter and absorption. When sound is absorbed the energy is converted into heat by frictional forces within the tissue. This increases with the density of the material involved and explains why less sound is absorbed in fluid than soft tissue and indirectly explains the phenomenon of distant acoustic enhancement.

Attenuation is directly proportional to frequency and is also greater in tissue such as fat. This explains why higher frequency ultrasound is used at shallower depths and why

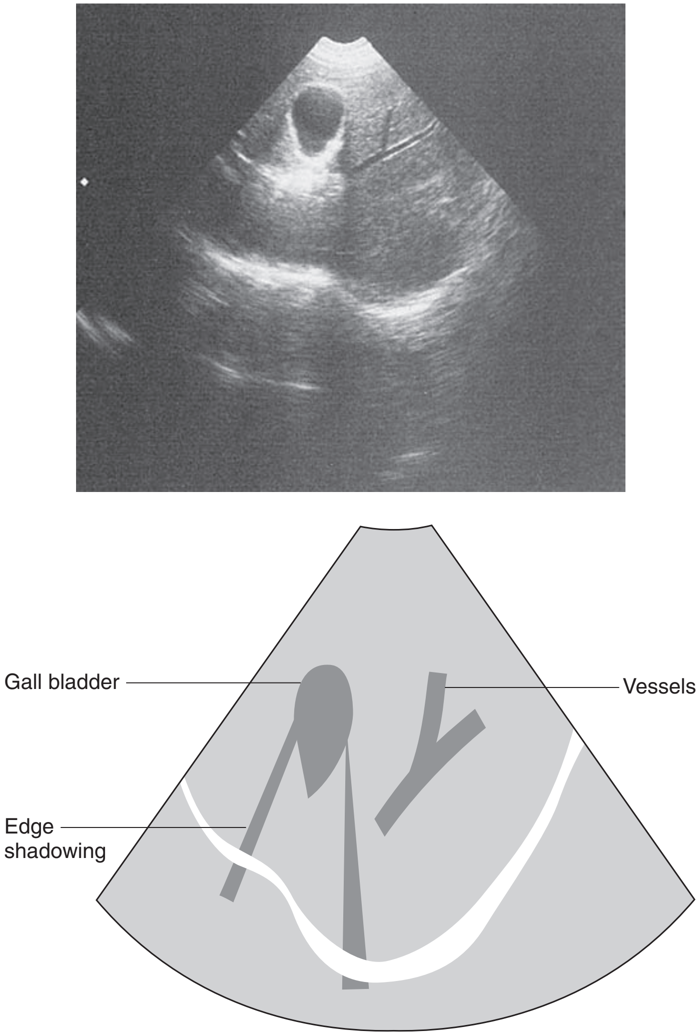


Figure 1.5 This ultrasound image shows the effect of edge shadowing at the edge of the gall bladder This is caused by refraction of the ultrasound beam as the speed of travel in the fluid viscus differs from that in the liver parenchyma.

in obese animals lower frequency ultrasound may be required to obtain a diagnostic image. The following is a general guide to the depth of penetration of a particular frequency transducer:

Frequency (MHz)	Depth (cm)
5.0	12–15
7.5	6–8
10	4

Production of ultrasound

A piezoelectric crystal within the transducer produces ultrasound waves. The purpose of the transducer is twofold; to convert electrical energy into sound energy and sound energy into electrical energy in return. Typically these crystals are ceramics or composite ceramics which have been heated to very high temperatures so that they develop piezoelectric or pressure electric properties. When a voltage or potential difference is applied across the crystal it is deformed and, in response, sound waves are produced. This is known as the piezoelectric effect. The composite ceramics can now produce sound of variable frequency as chosen. The voltage is applied intermittently and the crystal will only produce sound for approximately 1% of the time. Each small package of sound is only 2–3 wavelengths. This is known as the pulse length. The transducer receives the returning echoes for the remaining 99% of the time. When the echoes are returned the crystal is deformed again and this time an electrical signal is produced which is then displayed on the screen. This is known as the inverse piezoelectric effect. The greater the potential difference applied across the crystal the greater the intensity of the ultrasound beam produced. By increasing the power the intensity of the sound beam can be increased, but it is important to remember that power and intensity are not synonymous terms. Increasing the power will increase the intensity of the ultrasound beam but increasing the power actually increases the potential difference applied to the crystal and as a result of this the reverberations are greater and the intensity of the sound produced is greater. The frequency of production of the sound waves is known as the pulse repetition frequency and depends on the length of time taken for the sound waves to return from the tissues to the transducer. Only when the echoes have been received can another pulse be submitted. When imaging superficial structures a higher pulse repetition frequency is possible and at greater depths this must be lower.

When sound is produced it is produced in all directions but only that moving in a forward motion is useful for production of an image. The transducer therefore also houses a backing block which absorbs those waves which travel backwards. It is important that the material covering the transducer does not block or reflect any of the sound waves as they pass into the tissue and therefore a special material is used.

Figure 1.6a This diagram shows the shape of the unfocussed ultrasound beam as it emerges from the transducer with a narrow near field (NF) and a diverging far field (FF).

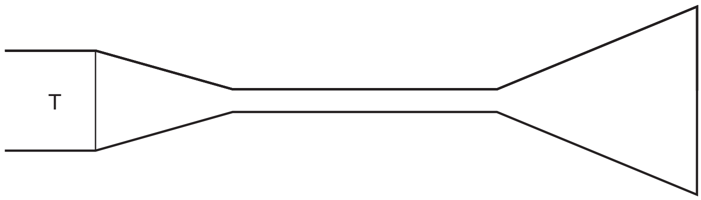


Figure 1.6b This diagram shows the effect of focussing the ultrasound beam so that there is a narrowed region of the beam where there will be improved resolution.

When the ultrasound beam is produced it diverges as it passes from the transducer into the tissues. The normal shape of the sound beam is shown simplistically in Figure 1.6, where a focussed ultrasound beam is also seen. As the ultrasound beam diverges the resolution of the ultrasound image is decreased. Resolution is an extremely important part of the ultrasound process. Spatial resolution can be divided into axial resolution and lateral resolution.

Axial resolution

Axial resolution is the ability to differentiate two points along the length of the ultrasound beam. With better axial resolution there is better image quality or detail of the image. The frequency of the transducer being used is of fundamental importance since with shorter pulse length the resolution is improved. Higher frequency ultrasound produces ultrasound with a shorter pulse length. Axial resolution cannot be any better than half the pulse length.

$$\text{Axial resolution} = \tfrac{1}{2} \times \text{Pulse length}$$

Therefore, two structures must be one pulse length apart to be recognised as separate structures (Figure 1.7).

Lateral resolution

Lateral resolution is the ability to differentiate two points lying side by side perpendicular to the ultrasound beam.

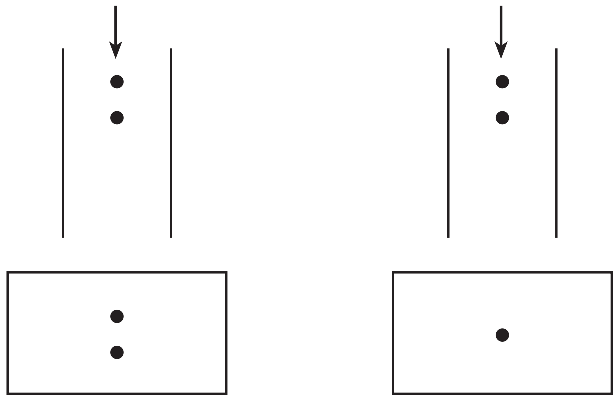


Figure 1.7 These diagrams show the principles of axial resolution. Where the two points are separated by one pulse length or more, they are displayed as separate structures on the screen. Where they are less than a pulse length apart they are seen as one point.

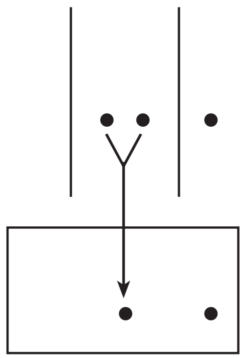


Figure 1.8 This diagram shows the principle of lateral resolution where two points perpendicular to the direction of the beam can be distinguished. Where these are within the beam they are seen as one but where they are separated by a distance greater than the beam width, they are distinguishable as separate objects.

Two points which are imaged within the beam are not displayed as separate structures but two points, where one is within the beam width and the other is not, will be displayed separately (Figure 1.8). In other words if the objects are separated by a beam width they are seen as separate structures and if not they are the same structure. Higher frequency transducers have a longer near field where the beam is narrower and so have higher lateral resolution. As lateral resolution depends on beam width it is better to use a narrower or higher frequency beam or to scan in the area within the focal zone of the transducer. Most modern transducers have a focussed beam and many have a variable or even more than one focal zone where this can be adjusted to suit the image.

Ultrasound modes

A mode (amplitude mode)

Amplitude mode, or A mode, is used less often today, following improvements in real-time B-mode ultrasound. In A mode, as its name suggests, the intensity of the returning echoes is displayed on the ultrasound screen as amplitude spikes. It was used predominantly for ocular ultrasound, but specialised equipment is required which uses a single, fixed ultrasound beam and it is beyond the scope of this text to go into more detail. In some centres A-mode ultrasound is still used.

B mode (brightness mode)

B mode, or brightness mode, uses the principle that each returning echo is displayed on the screen as a dot; the brighter the dot the higher the intensity of the returning echoes. Many ultrasound beams are used and a cross-sectional image is obtained and displayed on the screen. Real-time ultrasound, where the image displayed on the screen is the image which has just been acquired and which is constantly updated, uses B-mode ultrasound (Figure 1.9). The length of time each image stays on the ultrasound screen is governed by the persistence. This may be altered on most machines. B mode, real-time ultrasound, is currently the most commonly used in diagnostic imaging.

M mode (motion mode)

M mode, or motion mode, ultrasound uses a single ultrasound beam which is in a fixed position and records how the dimensions of the section being interrogated changes with time. The image is displayed on the screen with the dimensions on the vertical or Y-axis and time on the horizontal or X-axis (Figure 1.10). This is used predominantly in echocardiography to assess dimensions of cardiac chambers and also to allow the thickness of the walls of the heart to be assessed in relation to the cardiac cycle.

The ultrasound machine

Essentially the ultrasound machine comprises the control panel with some form of monitor, either a television or

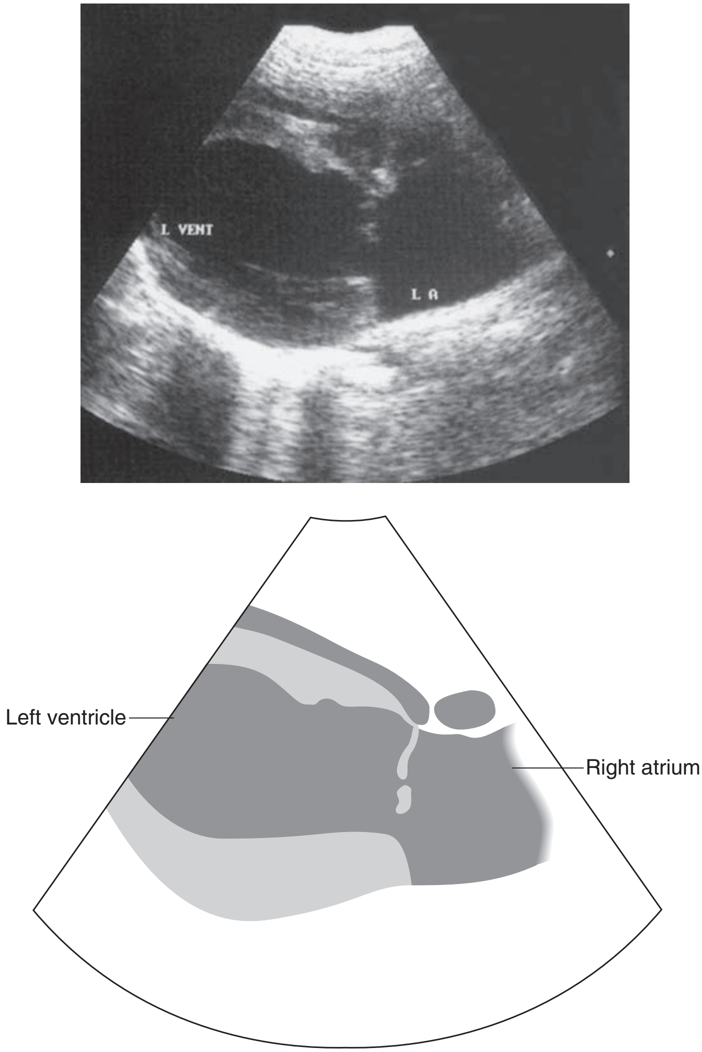


Figure 1.9 A typical B-mode image of the heart. Note the left atrial dilation in this Siamese cat with cardiomyopathy.

computer screen, and the transducers. The type of ultrasound machine used governs the latter.

Control panel

The control panel of each machine will differ in layout but essentially all have similar basic controls. All have a power control, gain/reject control, TGC control, and the ability to alter both the sector angle and depth control. All allow

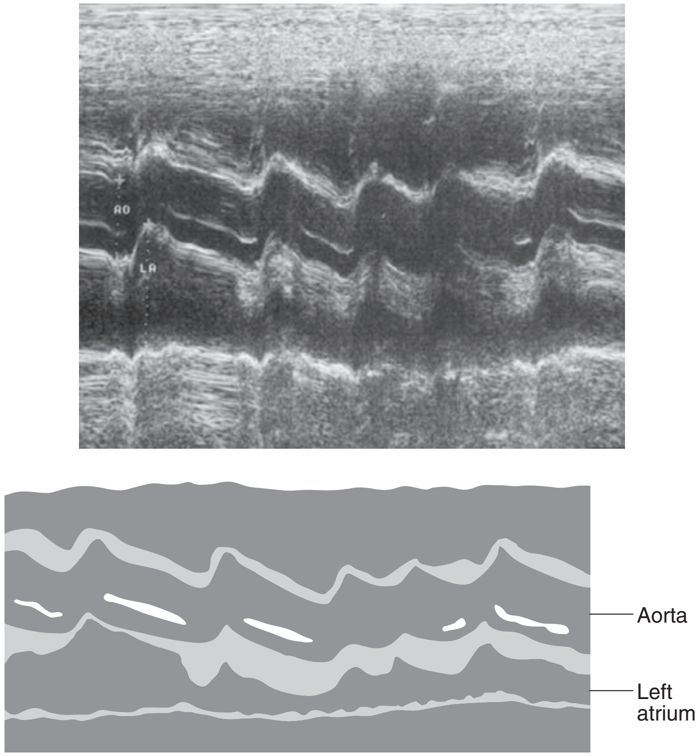


Figure 1.10 A typical M-mode image showing the left atrium and the aorta.

patient identification, date and time of the examination as well as annotation of the image. Most will allow at least distance measurements.

If Doppler ultrasound is available then there are controls for choosing PW/CW (pulsed or continuous wave) or even perhaps CF (colour flow) and there may be the option to choose duplex Doppler, which is the display of a standard B-mode image and Doppler tracing concurrently. It is also possible to choose different colour maps, to zoom into areas of interest and to use a split screen on many machines.

Power

By increasing power a larger voltage or amplitude of signal is applied across the piezoelectric crystal. The effect of this is that a greater intensity of sound is produced resulting in a brighter image. An analogy which has often been used to illustrate this is that of a gong being struck; when a greater force is used to strike the gong a louder noise is produced. When a greater potential difference is applied across the crystal a greater intensity of sound is produced.

Gain/reject

Gain is the degree of amplification which must be applied to the returning echoes since these are in most cases too weak to be detected readily. The amount or degree of gain applied is the ratio of the output signal to the input signal. This is usually displayed as overall gain and time gain compensation. Overall gain is increased amplification at all levels. This should not be confused with power.

The function of the reject is to allow rejection of the weak echoes, as they perhaps do not contribute to production of a clear or good image. This should be used sparingly as some fine detail may be lost.

Time gain compensation (TGC)

This control may be in the form of twist knobs or a sliding scale. Its function is to dampen down the higher intensity echoes which return from the more superficial structures and to amplify the echoes that return from the deeper regions, resulting in a more uniform image. This control is very important for producing a high quality diagnostic image.

Persistence

This affects the length of time the image remains on the screen before it is updated. Where possible the persistence should be as low as possible to get an improved and smoother image.

Sector width

The angle of the sector image can be changed on most machines and in general for overview work the angle is quite wide, but for high resolution work it is better to use a smaller angle which will use a higher frame rate.

Frame rate

This is the rate at which the images are updated on the display. This is affected by the type of examination, such as whether cardiac or abdominal; the depth of scanning, (higher frame rates are possible at shallower depths) and the angle of interrogation (with a narrower sector width there is a higher frame rate).

Split frame

In many machines it is possible to display two or more images on the screen so that images can be compared side by side. This is not to be confused with duplex scanning where B mode and Doppler or M mode can be displayed simultaneously.

ECG

This is used in the cardiology packages so that an ECG can be displayed concurrently with the B-mode, M-mode and Doppler images. It is usually possible to switch this function on and off.

Pre- and post-processing

These controls allow manipulation of the signals both before they are stored in the scan convertor of the machine (pre-processing) and after they have been stored, but before they have been displayed (post-processing). These controls must be used with caution as it is possible to manipulate the image adversely and therefore lose valuable information. Usually they have been set by the ultrasound manufacturers but may need to be reset for each type of application such as cardiac, abdominal or vascular.

Transducer types

The ultrasound transducer or probe is a key part of the ultrasound system. Transducers are classified according to whether they are mechanical or electronic and according to the shape of the field of view they produce. In most cases the latter can be seen from the shape of the probe itself. Most transducers at this time use an array of crystals rather

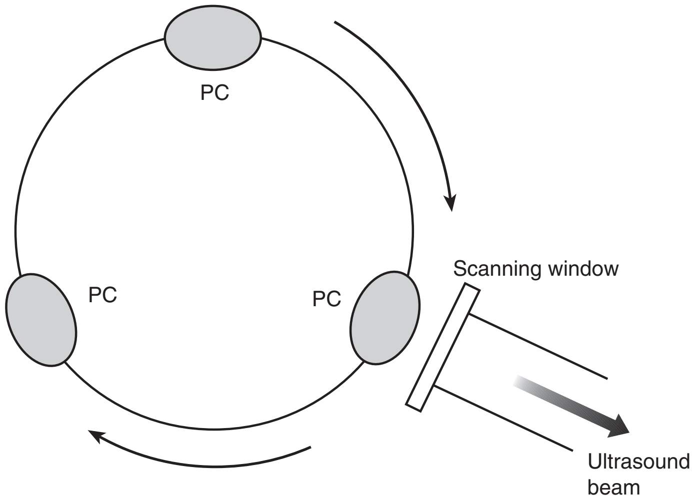


Figure 1.11 This diagram shows the basic construction of a mechanical sector transducer with the crystal elements arranged around the rotating wheel. The elements are swept past the scanning window. PC = piezo-electric crystal.

than a single crystal element. There are four main types of arrays in existence and use: linear array, curvilinear array, phased array and annular array. Only the latter is mechanical, the other three allow electronic steering. All four allow control of the beam width and focal distance.

Mechanical probes contain two to four crystals, on the rim of a rotating wheel. These are mechanically swept across the field of view to produce a fan-shaped or sector image (Figure 1.11). As mechanical probes have moving parts they produce a less sharp image and have a greater likelihood of breakdown. In some cases the crystals may be arrayed in concentric rings; these mechanical annular array transducers also produce a high quality image and may in some cases be slightly better than electronic linear arrays for detecting small focal masses. However they do not produce such good and reliable colour flow and Doppler signals as the electronic probes. Sector angle can be changed on most machines with a wider sector used to obtain an overview, while for better resolution a narrower sector is used with a faster frame rate.

Electronic probes have an array of crystals which are electronically fired to produce the image. The sequence in which this happens determines the shape of the field of view. Linear array transducers have a large number of rectangular crystal elements arranged in a line; sequential groups of these are fired intermittently to produce a rectangular image. There may be up to 250 elements with groups of up to 20 fired off each time. When the signal has returned to the transducer the next beam is fired off and this is from the parallel and adjacent group of crystals. This process is continued to the end of the array and then started over again. The whole scan takes in the order of $^1/_{30}$ of a second. This is repeated continuously and the image replaced each time on the screen. The advantage of this type of transducer is that there is a wide field of view with good definition of structures in the near field. The disadvantage is that there is a wide footprint, which restricts its application.

Curvilinear and phased array transducers produce a sector-shaped image, which is the shape of a section of a pie chart. Curvilinear arrays are really an adaptation of linear arrays and so are very similar in construction except that the crystal elements are laid out on a convex surface and the beam lines are not parallel but emerge like spokes on a wheel. These have a much bigger footprint than the phased arrays but they do have the advantage that the beam is perpendicular to the surface of the probe whereas the phased array transducers steer the beam off to the side so that it is non-perpendicular to the probe. This is to try to pick up echoes from the edges of the field but very often this is insensitive and there may be poor resolution at the edges of the image.

The phased array transducers have fewer elements, approximately 128. These transducers are smaller than the linear and curvilinear and so the elements are also narrower. All of the elements are fired off each time and beam steering is possible using time delay methods. One advantage of phased array over linear and curvilinear arrays is that there is a smaller footprint allowing access between the ribs so that these are suitable for echocardiography. The angle of the sector can be changed on most machines with the wide sector allowing a broad overview but the narrower beam providing more detail. For echocardiography a small footprint is necessary to allow access between the ribs and therefore curvilinear probes are not suitable. Phased array or microconvex probes may be suitable in these cases. The size of the footprint is the determining factor in choice of probe in veterinary practice where the small size of the patient body surface makes large probes inappropriate.

Image recording

Once the returning sound has been converted into energy at the transducer, the information is then passed to a scan convertor which stores the information and allows it to be displayed in a recognisable form such as on a TV or computer monitor. In most cases it is necessary to have some record of some or all of the examination and so it is important to be able to keep some hard copy. This may be in the form of photographic film such as in a multi-format camera, thermal printing, videotape recording or in many of the modern systems by digital archiving. For each particular system a range of options will be available and this should be discussed with the manufacturers or their representatives.

Biological safety

Biological safety of ultrasound is a complicated and as yet unresolved issue and the purpose of mentioning it in this book is to alert users to the potential complications that do exist. It is beyond the scope of this book to explore this issue fully and so the reader is directed to other texts for this (see Suggested Reading section at the end of the chapter).

It appears to be clear that ultrasound remains the safest of the diagnostic imaging modes given the potential effects of X-rays, gamma rays and even MRI. However, ultrasound can have potentially damaging effects such as tissue heating, cavitation and bruising if used irresponsibly.

The thermal effects of ultrasound are perhaps seen most at the body surface and at the surface of bone where increases in temperature of 2–3°C are quite possible. As cells may be damaged by extremes of temperature, it is important that the ultrasound mode be correctly set for mechanical and thermal indices.

Acoustic cavitation is commonly cited as a potential side-effect of ultrasound but this has not been proven where there are not pre-existing gas bodies.

Bruising is a potential problem but not usually at current diagnostic levels and any bruising would be expected to repair in a normal way.

As use of ultrasound increases it is clear that ideas on the safety of ultrasound are developing but this is still being researched. This data has been adapted from the literature in the field of human ultrasound.

Suggested reading

Boon, J. A. (1998) The Physics of Ultrasound. In: *Manual of Veterinary Echocardiography*, 1st edn, pp. 1–34. Lippincott Williams & Wilkins, Hagerstown, MD.

Curry III, T. S., Dowdey, J. E. & Murry Jr, R. C. (1990) Ultrasound. In: *Christensen's Physics of Diagnostic Radiology*, 4th edn. Leo & Febiger, Philadelphia.

Duck, F. A. (2 Jan 2003) Working Towards the Boundaries of Safety. In: *EFSUMB Newsletter*, **16**, 8–11.

Nyland, T. G. & Mattoon, J. S. (2002) Physical Principles, Instrumentation and Safety of Diagnostic Ultrasound. In: *Small Animal Diagnostic Ultrasound*, 2nd edn, pp. 1–18. W. B. Saunders Co., Philadelphia.

Zagzebski, J. A. (1996) *Essentials of Ultrasound Physics*. Mosby, St Louis.

Chapter 2

Ultrasound Artefacts

Johann Lang

Unlike radiographic artefacts, many ultrasound artefacts may be useful and a clear understanding of what they signify may be helpful with interpretation of an image. Improper use of the equipment, in particular when setting the controls, poor technique or inadequate patient preparation, may affect the quality and interpretation of an ultrasound image.

The artefacts discussed in this chapter are caused by physical interaction between the ultrasound beam and matter and are not due to improper scan technique.

Acoustic shadowing

Acoustic shadowing is produced by structures such as gas or bone, which reflect and/or absorb nearly 100% of the ultrasound beam (Figure 2.1). The result is that no echoes pass beyond the surface into the deeper tissues and this is displayed on the resultant image as a bright, echogenic line at the surface while the distant area is anechoic or black. This is known as acoustic shadowing. Both 'clean' and 'dirty' acoustic shadows have been described. Urinary

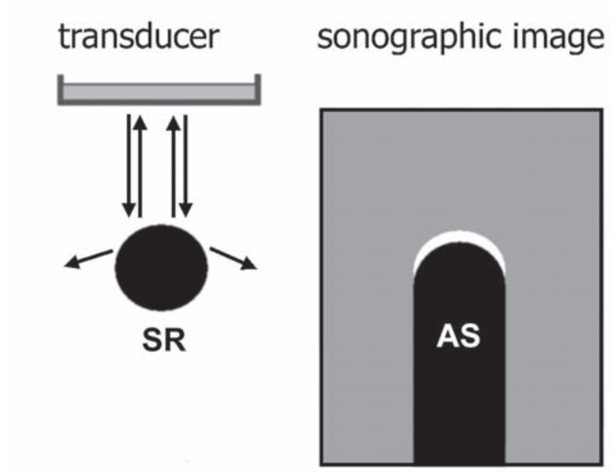


Figure 2.1 Acoustic shadowing. Structures with high attenuation (strong reflectors: SR) lead to complete reflection and/or absorption of the sound energy. Therefore, the reflective border of these structures is highly echogenic (white), while the area distant to such structures appears anechoic (acoustic shadowing: AS).

calculi (Figure 7.13, p. 135), gall stones, some foreign bodies (Figure 4.6, p. 47 distant to the suture – ‘FADEN’ = suture), or barium within the intestine behave in a manner similar to bone, reflecting and absorbing almost the entire sound energy and usually producing a completely black area distant to the object. This is described as a ‘clean’ shadow. Gas may produce a clean shadow, but can also lead to multiple reflections and reverberations thus creating an inhomogeneous or ‘dirty’ shadow. The type of shadow created therefore not only depends on the type of the object, but also on the size, composition and surface of the structure, as well as its position relative to the focal zone of the transducer.

A special type of acoustic shadowing, called edge shadowing, is produced at the lateral margins of cystic and other rounded fluid-filled structures such as gall bladder and urinary bladder, and may even be seen at the renal margins (Figures 2.2, 7.3, p. 117). Edge shadowing is caused by refraction of the sound beam at a fluid–tissue interface and is mainly due to different speed of sound through tissue and fluid with the rounded borders of a cyst acting as ‘lens’.

Acoustic enhancement

The energy of the ultrasound beam is attenuated as it passes through tissue. While travelling through a structure with low attenuation, the sound beam loses less energy than in the surrounding tissue. The result is an increase in the strength

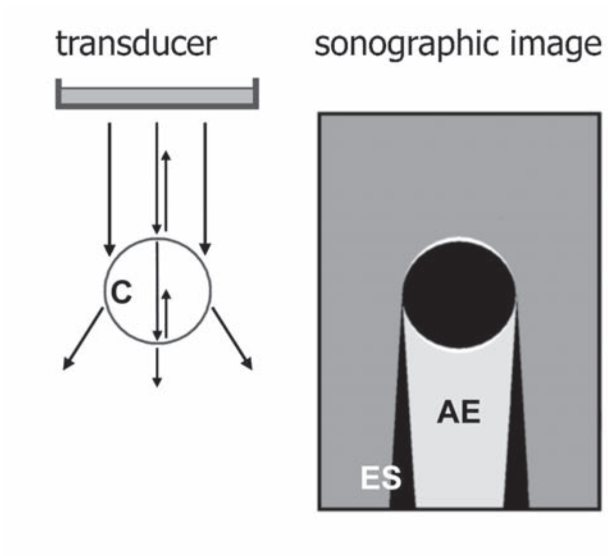


Figure 2.2 Edge shadowing. The lateral margins of a cyst (C) act as a lens deviating the sound beam either laterally or medially, thus creating an area with no echoes (Edge shadowing: ES) lateral to the area with acoustic enhancement (AE).

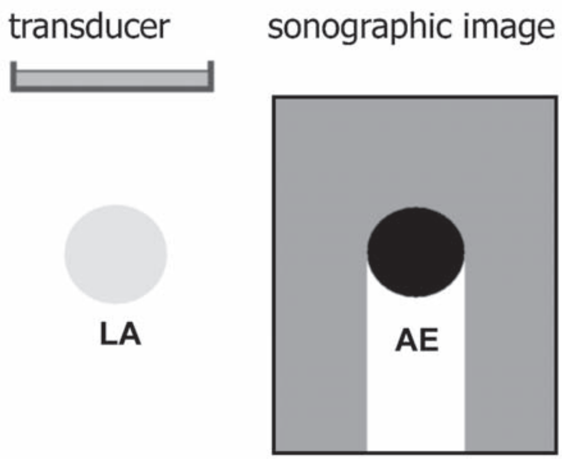


Figure 2.3 Acoustic enhancement.
The area distant to structures with low attenuation (LA) presents with increased echogenicity (AE).

of echoes returning from distant to this structure and this is displayed as an area of increased brightness on the screen (Figures 2.3, 7.3, p. 117). This mainly occurs distant to fluid-filled structures such as the gall bladder, urinary bladder or any cystic structure and is helpful in differentiating hypoechoic from fluid-filled structures. However, some solid hypoechoic structures may also show some distant acoustic enhancement.

Reverberation

Reverberation artefact involves reflection of the ultrasound beam backwards and forwards between the transducer and a highly reflective surface (Figure 2.4). This occurs commonly at the interface of the transducer and the body wall (external reverberation) but can occur at the interface of any highly reflective surface in the path of the ultrasound beam such as the small intestine or between the body wall and lung (internal reverberation).

Using thoracic ultrasound as an example, the ultrasound beam passes from the transducer, through the chest wall into the tissues and is reflected back to the transducer from the surface of the lung by the air. The transducer records this returning echo and an echogenic line is shown on the image. The echo bounces back to the lung surface and is again reflected back to the transducer. This is recorded but as this echo has travelled twice the distance and has taken twice as long to come back the ultrasound machine records it as having originated deep to the first. This is repeated

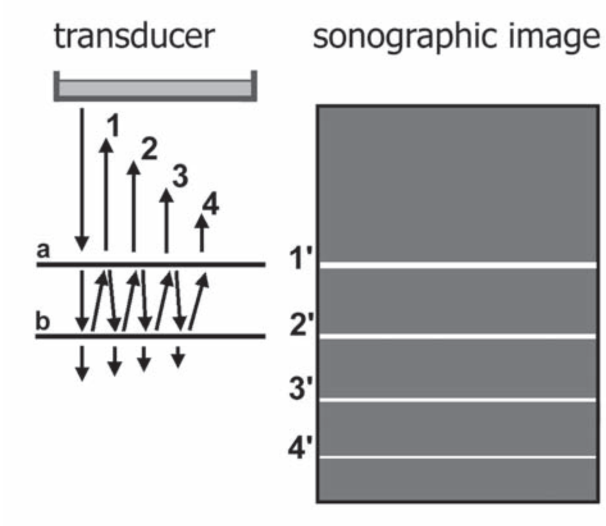


Figure 2.4
Reverberation. Here the sound beam is bouncing back and forth between the two strong reflectors (a, b), creating multiple echoes from one pulse (1–4). Because they arrive at a later time at the transducer, the echoes are displayed as lines deeper in the image or distant to the reflectors (1′–4′).

numerous times and concentric lines are displayed on the image. No information is gained beyond the surface and this explains why ultrasound has limited use for investigation of pulmonary disease and also why acoustic windows are so important for echocardiography. Comet-tail artefact is a special form of reverberation artefact and is characterised by regular bright continuous echoes. This tends to be produced by small superficially located foreign bodies or gas bubbles.

Mirror-image artefact

An ultrasound image is generated by transforming the time taken by the ultrasound beam to be reflected back from the tissues to the transducer into a location or depth, assuming that the ultrasound beam travels in a straight line to and from the reflector (Figure 2.5). Strongly reflective concave and convex interfaces such as the diaphragm–lung interface will reflect the sound beam back into the adjacent organ such as liver, where the echo is reflected back to the diaphragm–lung interface, from where it eventually is reflected back to the transducer. Because this echo has taken longer to return to the machine than if it had travelled in a straight line, and the computer assumes that the path of the beam has been straight it also assumes its origin has been in front of the diaphragm. Mirror-image artefact must not be confused with a diaphragmatic rupture or hernia and it is important to note that it is not seen in the presence of pleural effusion.

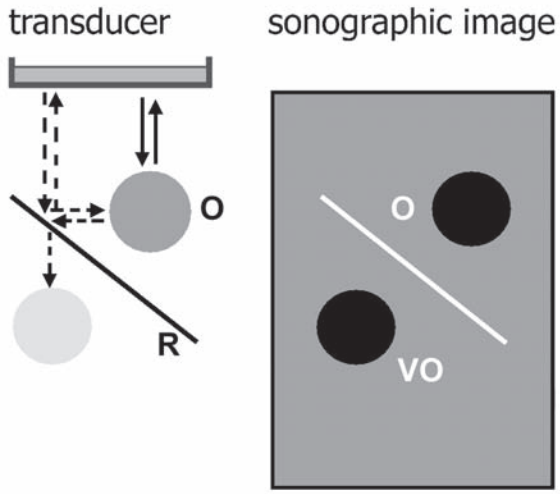


Figure 2.5 Mirror-image artefact. A strong, obliquely oriented surface with high acoustic impedance (R) may reflect the sound beam into an organ. Objects (O) reflect the sound beam back to this surface and from there to the transducer. Because of the longer return time of the sound waves, the object will be misplaced distant to the reflector (VO = virtual object).

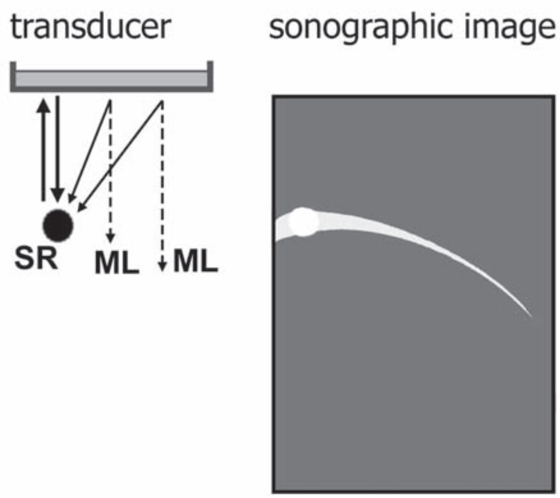


Figure 2.6 Side-lobe artefact. Strong reflectors (SR) in minor beams off the main sound beam can create echoes which are misplaced in the path of the respective main lobes (ML).

Side-lobe artefact

The ultrasound beam is composed of a main lobe and weaker secondary lobes, or side lobes. Normally, the image results from reflective objects in the path of the primary beam. However, highly reflective interfaces in the path of a side lobe can also result in an echo returning to the transducer. The returning echo will be 'misplaced' into the path of the main lobe. This artefact is generated if curved surfaces and strong reflectors such as air are present (Figure 2.6).

A variant of the side-lobe artefact is slice-thickness artefact, created in structures like the gall bladder and the urinary bladder. It mimics the presence of sediment within these structures and is called 'pseudo sludge' (Figure 7.10b, p. 130). Normally, if the entire width of the sound beam is placed within the bladder, pseudo sludge will disappear. True sediment can be differentiated from pseudo sludge by changing the position of the patient as it remains on the dependent side. The surface of the pseudo sludge will always be perpendicular to the sound beam.

Suggested reading

Barthez, P. Y., Leveille, R. & Scrivani, P. V. (1997) Side Lobes and Gating Lobe Artefacts in Ultrasound Imaging. In: *Veterinary Radiology and Ultrasound*, **38**, 387–393.

Curry III, T. S., Dowdey, J. E. & Murry Jr, R. C. (1990) Ultrasound. In: *Christensen's Physics of Diagnostic Radiology*, 4th edn, pp. 323–371. Leo & Febiger, Philadelphia.

Herring, D. S. & Bjornton, G. (1985) Physics, Facts and Artefacts of Diagnostic Ultrasound. *Veterinary Clinics of North America – Small Animal Practice*, **15**, 1107–1122.

Kirberger, R. M. (1995) Imaging Artefacts in Diagnostic Ultrasound: A Review. In: *Veterinary Radiology and Ultrasound*, **36**, 297–306.

Nyland, T. G. & Mattoon, J. S. (2002) Artefacts. In: *Small Animal Diagnostic Ultrasound*, 2nd edn, pp. 19–29. W.B. Saunders Co., Philadelphia.

Chapter 3

Indications and Technique

Paddy Mannion

Indications

Rarely should ultrasound be considered a substitute for radiography and usually it is only one of several investigative procedures to be carried out. The indications are varied and include obvious abnormalities such as a palpable mass or an audible cardiac murmur where the area of examination is clear. On some occasions the examination may be prompted by abnormal haematological or biochemical parameters and on other occasions may be as a result of clinical findings such as haematuria, jaundice or weight loss, where the underlying cause is not clear.

In most cases survey radiography must precede ultrasound and two projections of the area of interest, taken at right angles to each other, are considered as standard. Exceptions to this include respiratory distress where restraint for radiography may prove too stressful for the patient, or for pregnancy diagnosis where ionising radiations should be avoided. Where neoplasia is suspected, thoracic radiographs taken to check for pulmonary metastatic spread may also be appropriate. Usually right and left lateral projections taken on inspiration are preferable. The radiographs must be examined for abnormality and for any factors which might affect the quality and reliability of the ultrasound examination.

Patient preparation

The patient preparation required prior to the ultrasound examination is fairly straightforward. Preferably, the animal should have been starved for at least 12 hours prior to the examination. Food and gas in the stomach makes assessment of the lumen impossible and may obscure surrounding structures. In addition, faecal material in the colon may also obscure surrounding structures and precludes full