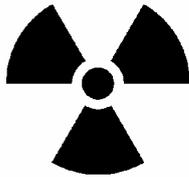


# Physical Principles and Measurement Accuracy of Bone Densitometry

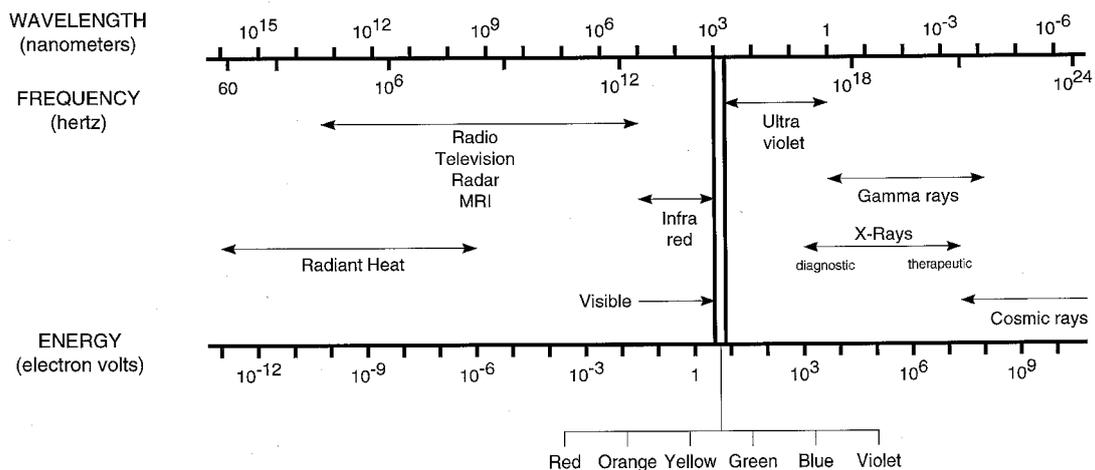
## Densitometry

### Introduction

Over the past few decades, a number of methods have been developed for the *in vivo* determination of bone density. The most important of these methods are based on measuring the attenuation (reduction in intensity) of a beam of either electromagnetic radiation (in the form of X-rays or gamma rays) or ultrasound when it passes through bone; in the latter case, the measurement of velocity has also been used for this purpose. There are significant differences between radiation and ultrasound as regards their interaction with bone. In general though, the nature of the measurement and its accuracy are influenced by the properties of both bone itself and the soft tissue that surrounds it.



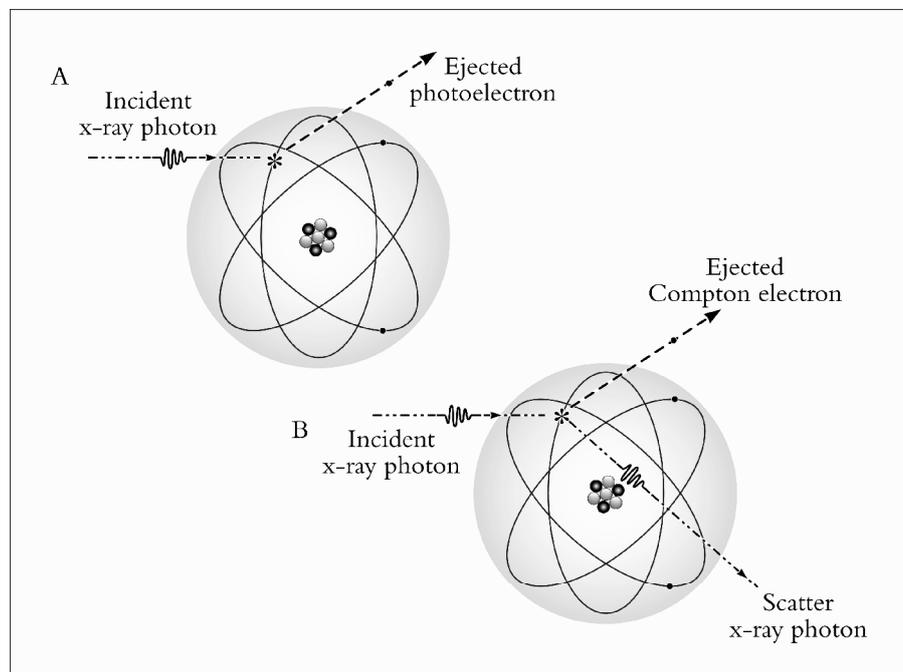
### Radiation Methods



Electromagnetic radiation is a form of energy that can travel through empty space or a material substance. It includes phenomena such as radio waves and light but of all the different types of electromagnetic radiation, it is only X-rays and gamma rays that have measurable attenuation through the range of tissue thicknesses typically found in

the human body. Although the velocity of the radiation is dependent on the properties of the material through which it passes, its value is so great that its measurement is impractical in a clinical setting. Furthermore, if radiation is regarded as a transverse wave phenomenon (sinusoidal variations of electric and magnetic field at right angles to the direction of travel), wavelengths in the X and gamma range ( $10^{-9}$  to  $10^{-13}$  m)<sup>1,22</sup> are much smaller than the dimensions associated with even individual trabeculae; this means that attenuation is dependent only on the amount of bone and not on its structure.

### Photon absorptiometry



In describing the interaction of electromagnetic radiation with tissue, it is helpful to use an alternative to the wave representation. In this alternative quantum description, the radiation is regarded as small packets of energy called photons. At the energies used in bone densitometry, these photons interact with atomic electrons mainly by two processes: photoelectric absorption and Compton scattering. In the former case, the photon yields all its energy to an electron and therefore disappears; in Compton scattering on the other hand, only part of the photon energy is given to an electron and the interaction produces a scattered photon whose direction has changed and whose energy is reduced. In both cases, energy is absorbed by tissue through the electron as it loses kinetic energy in ionising atoms of the tissue. If all the photons initially have

the same energy (i.e. they are monoenergetic or monochromatic), the attenuation of a radiation beam by a thickness  $x$  of a single tissue is described by an exponential law:

$$I_r = I_i \exp(-\nu x) \quad \text{Equation (1)}$$

where  $I_i$ ,  $I_r$  are the incident and transmitted intensities and  $\nu$  is a property of the tissue known as the linear attenuation coefficient; if  $x$  is expressed in cm, the unit of  $\nu$  is  $\text{cm}^{-1}$ . Here intensity represents the total energy per second in unit area of the beam and the attenuation is given by  $\ln_e(I_r/I_i)$ . The linear attenuation coefficient is directly proportional to the physical density  $\rho$  ( $\text{g cm}^{-3}$ ) of the tissue but also depends on the atomic number of the elements within it (i.e. on the number of electrons surrounding each atom) and on the photon energy<sup>1,22</sup>.

When equation (1) is applied to bone densitometry, the simplest model of the human body that can be used is one that comprises two distinct types of tissue: bone mineral and soft tissue. Here, soft tissue can include muscle, fat, blood, skin, viscera etc. since their linear attenuation coefficients are similar. This follows because they have broadly similar physical densities and are composed of the same elements (mainly carbon, hydrogen and oxygen). Bone mineral, on the other hand, has a significantly greater attenuation coefficient because its density is relatively high and it includes higher atomic number elements (phosphorus and, more importantly, calcium). Furthermore, bone mineral (calcium hydroxyapatite) is distinct from whole bone tissue; indeed, the organic framework of bone (consisting mainly of collagen) and bone marrow are regarded as soft tissue components.

If a monoenergetic radiation beam is passed through part of the body that contains bone surrounded by soft tissue, the transmitted intensity is given by:

$$I_r = I_i \exp(-\nu_b x_b - \nu_s x_s) \quad \text{Equation (2)}$$

where  $b$ ,  $s$  represent bone mineral and soft tissue respectively. In equation (2),  $x_b$  is the equivalent thickness of bone mineral in the path of the radiation beam as if it had been isolated from whole bone tissue as a homogenous layer. On the other hand,  $x_s$  is the equivalent thickness of some average soft tissue that gives the same attenuation as the combination of actual soft tissue components in the beam path.

Equation (2) can be the basis for bone densitometry by experimental measurement of the incident and transmitted intensities (with a suitable radiation detector such as a scintillation crystal) and the use of assumed values for the attenuation coefficients. In this context, it is usual to replace the tissue thickness  $x$  by its area density (in  $\text{g cm}^{-2}$ ) since these quantities are directly proportional:

$$x = \frac{1}{\rho} \cdot \frac{m}{A} \quad \text{Equation (3)}$$

where  $m$  is the mass of tissue in the beam and  $A$  is the beam area. Thus in equations (1) and (2) above, the product  $\nu x$  may be replaced by  $\mu M$ ; here  $\mu$  is the mass attenuation coefficient (given by  $\nu/\rho$  in  $\text{cm}^2 \text{g}^{-1}$ ) and  $M$  is the area density (given by  $m/A$ )<sup>2</sup>.

### Single photon absorptiometry

However, a problem remains: there are two unknown quantities (the area densities of bone mineral and soft tissue) but only one equation to describe their combined attenuation of the radiation beam. In the technique of single photon absorptiometry (SPA), this problem was overcome by surrounding the body part by a soft tissue equivalent material (usually water) such that the total thickness of bone and soft tissue was constant. In practice, this restricted the measurement to sites on the forearm or other parts of the appendicular skeleton. Water was chosen because its attenuation coefficient closely matches that of the soft tissue components. By measuring the transmitted intensity at a point through bone surrounded by water and soft tissue and through a point adjacent to bone comprising only soft tissue and water ( $I_b$  and  $I_s$  respectively), it is possible to calculate the area density of bone mineral as:

$$M_b = \frac{\ln(I_s / I_b)}{\mu_b - (\rho_s / \rho_b) \mu_s} \quad \text{Equation (4)}$$

Again, values must be assumed for the physical density and mass attenuation coefficients of bone mineral and soft tissue<sup>3,22</sup>.

$M_b$  is the bone mineral density (BMD) at a measurement point that corresponds to a particular path of the radiation beam through the body. If it is measured at regular intervals as the beam is scanned across the bone (such as radius or ulna) and the

values integrated (summed), it is possible to calculate the bone mineral content (BMC) at that measurement site. Under these circumstances, BMC is expressed in units of  $\text{g cm}^{-1}$ , i.e. it is a linear density (the mass of bone mineral per unit length of bone). Alternatively, several such projections across the bone may be made by scanning the radiation beam in a raster pattern over a region of the forearm, for example. If the area density measurements are now integrated in both directions (across the bone and along the bone), this yields a BMC which is the total mass of bone mineral (in g) in that region. It is also possible to use the data to calculate the projected area of bone in the region; dividing BMC by area gives average BMD (i.e. the average area density of bone mineral) and it is this quantity that is usually quoted in clinical bone densitometry. Its utility lies in the fact that it is an important determinant of bone strength and hence fracture risk.

### Dual photon absorptiometry

The restrictions on measurement site imposed by the need to surround the body part by water was the stimulus that led to the development of dual photon absorptiometry (DPA). In this approach, a pair of simultaneous equations of the same type as equation (2) are generated by measuring transmitted intensities at two different photon energies, thus allowing their solution for the area densities of bone mineral and soft tissue. In particular:

$$M_b = \frac{\ln \left( I_i^L / I_r^L \right) - R_s \cdot \ln \left( I_i^H / I_r^H \right)}{\mu_b^L - R_s \cdot \mu_b^H} \quad \text{Equation (5)}$$

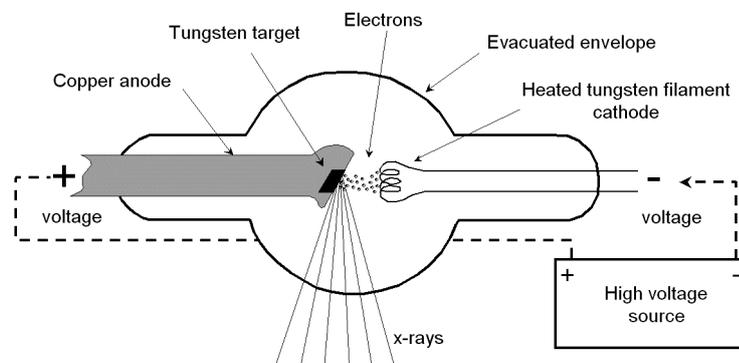
where

$$R_s = \mu_s^L / \mu_s^H \quad \text{Equation (6)}$$

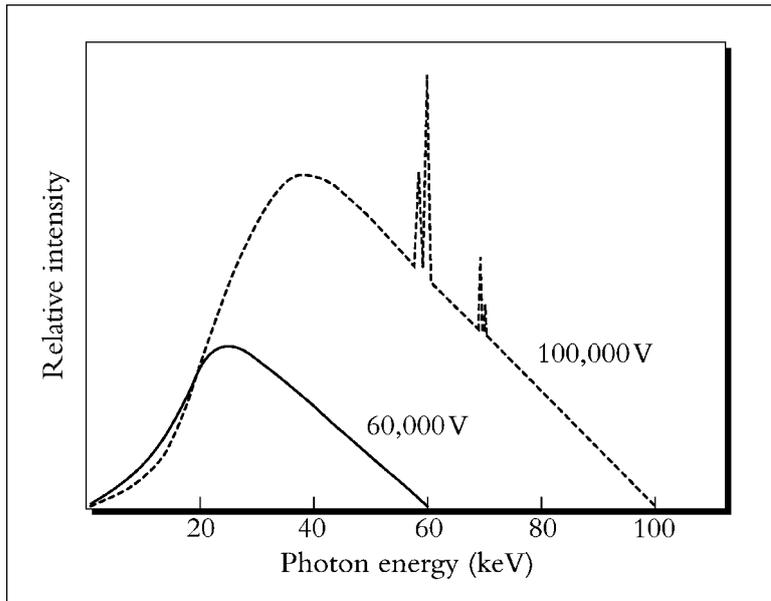
and H, L represent the high and low photon energies respectively<sup>2,4,22</sup>. The ratio  $R_s$  may be determined for an individual patient from measurements of transmitted intensity at points which do not contain bone (where  $M_b$  is zero)<sup>5</sup>. In this way, it became possible to measure BMD and BMC in clinically important sites of osteoporotic fracture such as the spine and hip. In principle, the method could be applied to the measurement of these quantities in the whole body or any part of it.

In both implementations of photon absorptiometry, radioactive nuclides that emitted photons at one or two discrete energies were used as the source of electromagnetic radiation. Most single photon absorptimeters used the 27 kiloelectronvolt (keV) characteristic X-rays produced as a result of the decay of  $^{125}\text{I}$  although others were based on the 60 keV gamma radiation from  $^{241}\text{Am}$ . Almost universally, dual photon equipment had sources of  $^{153}\text{Gd}$ ; its gamma photon energies of 44 and 103 keV were ideally suited for bone densitometry<sup>5</sup>. Here keV is a unit of energy commonly used in radiological physics; it is equal to  $1.6 \times 10^{-19}$  joule.

### X-ray absorptiometry



The number of radioactive nuclei in the radiation source (proportional to its activity measured in becquerels) was limited by cost and radiation safety considerations. This, in turn, limited the emission rate of photons and beam intensity that prolonged scanning time and had an adverse effect on measurement precision. These practical disadvantages led directly to the introduction of an X-ray tube as the radiation source because of its higher photon output and, in effect, SPA and DPA have been replaced by single energy X-ray absorptiometry (SXA) and dual energy X-ray absorptiometry (DXA) respectively.



Unlike radioactive sources, X-ray tubes produce radiation with a continuous spectrum of photon energies from low values up to a well-defined maximum governed by the value of the high voltage used to drive the tube<sup>1,22</sup>. In SXA, this polyenergetic (or polychromatic) spectrum may be used in essentially unmodified form and equation (4) applies if the attenuation coefficients are calculated at the effective photon energy of the beam. Like SPA, SXA suffers from the disadvantage that the body part must be immersed in water<sup>6</sup>, thus restricting its application to peripheral sites. DXA, on the other hand can access the same range of measurement sites as DPA, although specialised equipment has been developed specifically for appendicular skeletal regions such as the forearm (peripheral- or p-DXA)<sup>7</sup>. In both types of DXA equipment, the spectrum must be modified to produce two effective photon energies and equipment manufacturers have achieved this using one of two possible approaches; in both cases though, point BMD is calculated using equation (5).

In the first of these approaches (as used by GE and Norland), the X-ray generating voltage is stabilised to a high degree and a thin sheet of metal (a filter) is placed in the radiation beam near the X-ray tube. The metal is carefully chosen such that its electronic K-shell absorption edge<sup>1</sup> lies near the mid-range photon energy. This has the effect of producing a spectrum that has two intensity peaks mimicking the dichromatic emissions from <sup>153</sup>Gd. In one type of DXA scanner, a cerium filter is used with a tube voltage of 80 kV to give effective photon energies of about 40 and 70

keV<sup>8</sup>. In another, a samarium filter and a generating voltage of 100 kV give effective energies of 47 and 80 keV<sup>9</sup>.

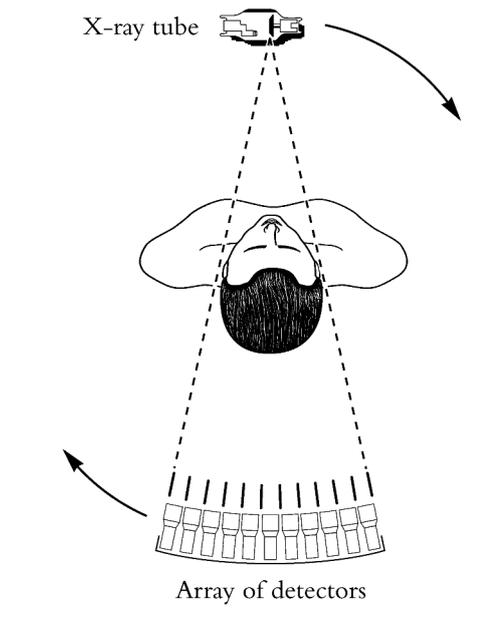
Hologic's approach to DXA is rather different: the tube voltage is continuously switched between a low (70 kV) and high (140 kV) value in synchronism with the frequency of the electrical mains<sup>10</sup>. Thus the effective energy of the X-ray beam alternates between 45 and 100 keV<sup>2</sup>. With this method, beam hardening (the preferential attenuation of lower energy photons as radiation passes through the body) is a significant problem. This has been overcome by passing the beam through a rapidly spinning calibration wheel or cylinder that permits measurement of the energy-dependent attenuation coefficients used in equation (5) on a point by point basis<sup>2, 11</sup>.

In early X-ray absorptiometers, the radiation was collimated (shaped) to form a thin cylindrical (pencil) beam, but in more modern machines, the beam is in the shape of a thin fan. This allows the simultaneous measurement of intensities transmitted through many paths through the body that, in turn, significantly reduces scanning time. Both pencil and fan beam systems display individual BMD values as a digital image, each pixel (picture element) of which corresponds to a measurement point through the patient. Because of the uncertainties involved in its calculation,  $M_b$  may have non-zero values even when the radiation beam does not pass through bone. An edge detection algorithm (implemented as a computer programme) finds the bone edges, usually by taking some cut-off value of  $M_b$  (e.g.  $0.1 \text{ g cm}^{-2}$ ). Once the edges have been found, the projected area of bone may be calculated and this allows the computation of BMC and average BMD as in the earlier photon absorptiometry.

## Quantitative Computed Tomography

Photon and X-ray absorptiometry are projection techniques. As a consequence, BMD can be measured only as an area density and it includes mineral from both cortical and trabecular bone in the beam path (so called integral bone). The best that can be achieved with these methods as regards separate measurement of the two types of bone tissue is to choose skeletal sites and projections such that the radiation beam passes through bone that contains predominantly one type or the other. For example, an antero-posterior (AP) projection of the mid-shaft radius and ulna yields the BMD

of mainly cortical bone whereas for a lateral projection of a lumbar vertebral body, a relatively large amount of trabecular bone is included in the measurement. In the latter case, lateral and AP measurements can be combined to give an estimate of volume density in the vertebral body, but the calculation relies on many assumptions (about bone shape etc.) and is not commonly performed.



These problems may be removed by using a tomographic (slice) technique, and one such method, X-ray computed tomography (CT), has been used for general imaging applications for some time. In CT, the X-rays are again collimated to a thin fan-shaped beam but the intensity of radiation transmitted through the body is measured not just in one projection, but in many projections as the X-ray tube is rotated around the body. From these multiple projections, a computer algorithm reconstructs an image of a transverse body slice<sup>1,22</sup>. In this digital image, each pixel now represents a small volume (voxel) of tissue within the body (rather than a thin cylinder of tissue through the patient as in radiation absorptiometry) and the attenuation data for a particular tissue  $t$  is stored as Hounsfield numbers  $H_t$  where:

$$H_t = 1000 \frac{(v_t - v_w)}{v_w} \quad \text{Equation (7)}$$

In equation (7),  $\nu_t$  and  $\nu_w$  are the linear attenuation coefficients for the tissue and water respectively; the values of the former are measured during each scan whereas the value of the latter is obtained from a prior calibration scan with a water phantom.

Quantitative computed tomography (QCT) is an extension of ordinary CT in which the Hounsfield numbers are used not only to produce an image but also to express tissue density. If  $t$  represents whole bone tissue that can be assumed to consist of bone mineral and soft tissue components as before, then

$$C_b = \frac{H_t - H_s}{H_b - H_s} \cdot \rho_b \quad \text{Equation (8)}$$

where  $C_b$  is the concentration of bone mineral in whole bone (i.e. its volume density)<sup>12</sup>. In practice,  $H_t$  is averaged over a particular scan area e.g. the interior of the vertebral body that comprises pure trabecular bone. Equation (8) applies to single energy QCT in which the X-ray tube is driven at one voltage only.  $H_t$  is measured during the scan and it is possible to calculate  $C_b$  using assumed values for the other parameters at the effective photon energy of the radiation beam. More often though,  $H_b$  and  $H_s$  are also measured by simultaneously scanning a calibration phantom containing bone mineral and soft tissue equivalent materials; in this case,  $C_b$  is expressed as the concentration of the mineral equivalent. As in X-ray absorptiometry, the method may be extended to dual energy by scanning at two different tube voltages<sup>12</sup>.

In principle, any whole body CT scanner may be used for QCT but not all manufacturers have developed specific software and calibration phantoms. Typically, measurements are made with a tube voltage of 120-140 kV that gives an X-ray beam with effective energy 60-70 keV<sup>13</sup>. Specialised small-bore CT scanners have been produced for bone mineral measurement in the appendicular skeleton (peripheral- or p-QCT); the early models used a radioactive source of <sup>125</sup>I but more modern devices are based on an X-ray tube<sup>7</sup>.

## Accuracy of radiation methods

The accuracy of a measurement technique describes its ability to determine the true value of the quantity it purports to measure. In clinical bone densitometry, good accuracy is important when a patient's BMD is compared with a reference range although for the estimation of BMD change with time, it is precision (or reproducibility of measurement) which is the more relevant.

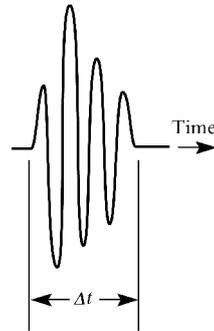
Several factors limit the accuracy of bone mineral measurement by radiation absorptiometry, the most important of which is the inhomogeneity of soft tissue. As has been mentioned, soft tissue is really a mixture of several different tissues, each with its own attenuation coefficient. In particular, fat is relatively less attenuating than the other components (because of its greater hydrogen content) and, in general, it is unevenly distributed within the region of BMD measurement. Another factor that affects accuracy is uncertainty in the values of density and attenuation coefficients for both the bone mineral and soft tissue components. To an extent this may be overcome by direct measurement, as described for DPA and DXA above.

As regards in vivo bone densitometry, the actual measurement of accuracy is not a practical possibility. However, it is possible to simulate the human body using models (or phantoms) of various types. One approach is to measure the BMC of excised bones submerged in water. These are subsequently heated to a high temperature in a furnace to remove the organic (soft tissue) components and the mass of residual bone ash (mineral) is compared with the measured BMC. Another approach involves the use of known amounts of calcium hydroxyapatite itself or a compound that has very similar attenuation properties surrounded by a soft tissue equivalent material.

The accuracy error of photon absorptiometry has been estimated as 4-8% with a similar value for SXA and a somewhat better range of 4-6% for DXA<sup>2</sup>. However, it can be as poor as 11%<sup>14</sup> and is worse for lateral projections compared with the more usual AP projections<sup>15</sup>.

Inhomogeneities and uncertainties in the composition of soft tissue also influence the accuracy of QCT. In particular, the fat content of yellow marrow can have a serious

effect on the measurement of vertebral trabecular bone<sup>16</sup>. Accuracy is further degraded by beam hardening and other machine-related factors<sup>13</sup> so that overall, its value for single-energy methods lies in the range 5-15%<sup>2</sup>. With two effective beam energies, this improves to 3-10% but at the cost of poorer precision<sup>13</sup>.



## Ultrasound methods

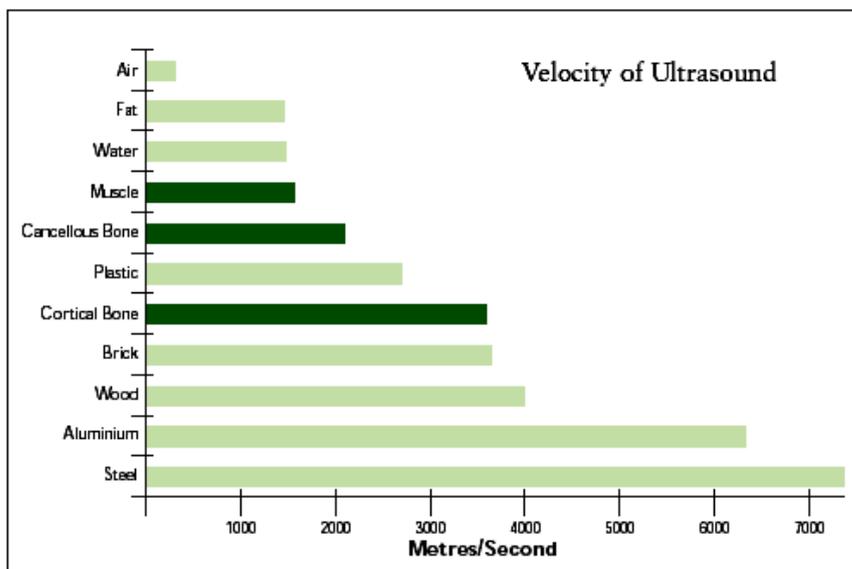
Ultrasound is a term that is applied to sound waves whose frequency lies above the threshold of human hearing i.e. greater than 20 kilohertz (kHz) where 1Hz is one wave per second. In contrast to electromagnetic radiation, it is a mechanical phenomenon that propagates through tissue as a sinusoidal vibration<sup>1,2,17,22,23</sup>.

Whether the application is diagnostic imaging or bone densitometry, ultrasound is produced and received by a transducer; this takes the form of a piezo-electric crystal whose dimensions change when a high voltage is applied to it. If the voltage is applied as a pulse (i.e. for only a short period of time), the crystal vibrates at a characteristic frequency that depends on its thickness. Under these circumstances, contact between the crystal and a material (such as water or tissue) allows a pulse of ultrasound to travel through that material. However, an ultrasound pulse (as opposed to a continuous wave) is represented by a range of frequencies centred on the characteristic frequency of the piezo-electric crystal: hence the term 'broadband' ultrasound<sup>18</sup>.

Ultrasound methods used in bone densitometry are mainly transmission techniques: the sound pulse enters bone at one point and is detected after it has travelled a distance through the tissue. This contrasts with conventional diagnostic imaging with ultrasound that is based on its reflection at interfaces between tissues such that the

pulse travels back towards its point of origin (the pulse-echo method). Furthermore, the frequencies used for bone densitometry (100 kHz to 1.5 MHz) are lower than those appropriate for soft tissue imaging (2.5 to 15 MHz)<sup>1, 7</sup>. The pulse travels through bone with measurable velocity (given by the product of wavelength and frequency) and is attenuated. Good acoustic coupling between the transducers and the skin surface is very important; this may be achieved either by using an ultrasound gel between them or by immersing both the transducers and the body part in water.

Ultrasound interacts with whole bone tissue, either in cortical or trabecular form. As a consequence, measured parameters (such as velocity and attenuation) reflect properties of the whole tissue and not just its mineral component as with electromagnetic radiation methods. Of course, bone mineral contributes to these bulk tissue properties but so do the organic framework and bone marrow. Quantitative ultrasound (QUS) has become accepted as a generic term for ultrasound measurements of velocity and attenuation in bone.



## Velocity

When ultrasound travels through a tissue (such as bone), mechanical motion is transmitted from one particle to the next through attractive forces between them. Thus there is no bulk movement of the material but its particles vibrate about their mean positions. Most of the ultrasound energy moves as a longitudinal wave (i.e. one

in which the vibrations occur in the same direction as the wave travels) although in bone, the propagation of transverse (shear) waves is also possible. The velocity of ultrasound in bone ( $V_t$  in units of  $\text{ms}^{-1}$ ) may be calculated simply by dividing the length or width of bone by the measured time it takes for the pulse to traverse it<sup>19</sup>. For cortical bone velocity in the tibial shaft, for example, a fixed bone length is determined by the distance between the transmit and receive transducers. On the other hand, for bone velocity determination in the calcaneus (a site of mainly trabecular bone), bone width may be determined from heel width by the use of a pulse-echo method to measure the thickness of overlying soft tissue<sup>7,20</sup>.

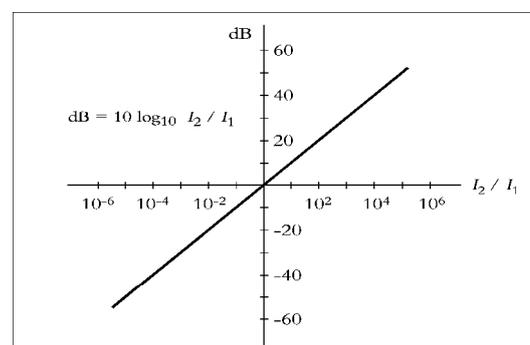
Even though bone is anisotropic and inhomogenous, to a good approximation the velocity of longitudinal waves is given by the expression

$$V_t = \sqrt{E / \rho_t} \quad \text{Equation (9)}$$

where  $t$  again stands for whole bone tissue and  $\rho$  for physical density<sup>22,23</sup>. Here  $E$  is a property of bone known as its modulus of elasticity or Young's modulus.  $E$  is a measure of the tissue's resistance to deformation and it increases as the strength of the inter-particle force increases<sup>7,17</sup>. However,  $E$  is itself a function of  $\rho_t$  as is the ultimate tensile strength of bone and so  $V_t$  may be regarded as a surrogate for bone strength.

## Attenuation

In general, the attenuation of ultrasound is mainly caused by absorption and scatter as is the case for electromagnetic radiation. However, the mechanisms involved are very different. Ultrasound absorption is primarily due to internal friction between the vibrating particles of the tissue and in this way the energy is eventually dissipated as heat. It is the dominant attenuation mechanism in cortical bone. Scattering occurs when particles absorb ultrasound energy and re-radiate it in different directions. It is particularly important when the ultrasound wavelength is greater than the dimensions



of the scattering particle<sup>7</sup>.

In a uniform tissue which has no boundaries (at which reflection and refraction can occur), the attenuation of a plane ultrasound wave is described by equation (1) where  $\nu$  is simply known as the attenuation coefficient. It is common to express the attenuation in decibels (dB) i.e. as  $10\log_{10}(I_r/I_i)$  (where r, i again refer to the transmitted and incident wave), in which case the units of  $\nu$  are  $\text{dB cm}^{-1}$ . For ultrasound, the attenuation coefficient is approximately proportional to frequency f:

$$\nu \approx kf \quad \text{Equation (10)}$$

where k is a constant (with usual units of  $\text{dB cm}^{-1} \text{MHz}^{-1}$ )<sup>17,22,23</sup>. In contrast to velocity, no theoretical relationship between ultrasound attenuation and the mechanical properties of bone has been established<sup>20</sup>.

Bone attenuation measurements are made by comparing the magnitude or amplitude A of an ultrasound pulse transmitted through bone (t) with that transmitted through an equal length of water (w) as a reference material, over a chosen frequency range (spectrum)<sup>18</sup>. This substitution method compensates for any variation in pulse amplitude with frequency for a particular pair of transducers<sup>7</sup>. The attenuation is expressed as the ratio of pulse amplitude through water to that through bone; in dB it is given by  $20\log_{10}(A_w/A_t)$  since I is proportional to  $A^2$ . Provided that an appropriate frequency range is chosen (e.g. 0.2 to 0.6 MHz), the variation of attenuation with frequency is linear (from equation (10)) and the slope is known as the broadband ultrasound attenuation (BUA in  $\text{dB MHz}^{-1}$ ). Because of the relatively low attenuation of cortical bone, measurements are generally performed at sites of mainly trabecular bone, in particular the calcaneus. BUA may be normalised for bone size by dividing it by bone width to yield a measurement in units of  $\text{dB MHz}^{-1} \text{cm}^{-1}$ . BUA itself is related to the average BMD (i.e. area density of bone mineral) at the site of measurement, whereas BUA/Bone Width is related to the volume density of mineral<sup>18</sup>.

## Accuracy of ultrasound methods

As with electromagnetic radiation, the accuracy of QUS is affected by variability in the thickness and composition of soft tissues within and surrounding bone. For example, when velocity is measured in the calcaneus by a contact method, inclusion of the overlying soft tissue thickness in the total width yields a limb velocity that differs from the bone velocity  $V_t$ . Measuring the ultrasound transit time between a pair of fixed transducers in a water bath with and without the heel in position gives a time of flight velocity which differs from both the limb and bone velocities.

Similarly, corrections must be made for the effects of overlying soft tissue in order to record a true bone velocity in the tibial shaft. In trabecular bone, fatty marrow in the inter-trabecular spaces influences both the measured BUA<sup>21</sup> and velocity. Patient positioning with respect to the transducers and factors such as the reflection and diffraction of sound represent other sources of error<sup>7</sup>.

Since ultrasound transmission is affected not only by the mineral content of bone but also by other material and structural properties, it has been argued that accuracy should be expressed with respect to these properties. Although bone strength would be the most appropriate, it is difficult to measure and so accuracy estimation may be confined to surrogates such as BMD and elasticity. Equipment manufacturers have attempted to combine velocity and attenuation measurements into a single parameter that may better characterise bone strength<sup>7</sup>.

## Conclusion

The physical basis of the interaction of high-energy electromagnetic radiation with matter is well established and consistent mathematical descriptions are possible. As a consequence, there is clear understanding of the bone density parameters that are measured with radiation methods. Such is not the case with ultrasound, and a complete description of its interaction with bone still remains an important if elusive goal for those who advocate its expanding role in bone densitometry.

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