The Resolution Integral – a tool for characterising the performance of diagnostic ultrasound scanners

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Abstract

In this paper, we describe how the resolution integral can be used as a tool for characterising the grey-scale imaging of diagnostic ultrasound scanners. The definitions of resolution integral, characteristic resolution and depth of field are discussed in relation to grey-scale imaging performance, together with a method of measuring these parameters using the Edinburgh Pipe Phantom. We show how the characteristic resolution and depth of field can be used to quantify the differences between transducers designed for different applications and how they are useful in identifying and quantifying changes in the performance of individual transducers.

Keywords: Quality assurance, image quality, Edinburgh Pipe Phantom, resolution integral

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Introduction

The assessment of grey-scale imaging performance is important for several reasons: to ensure that scanners are fit-for-purpose; to assess new imaging modalities and signal-processing techniques; and to allow informed decisions to be made when purchasing new equipment. Because the resolution of ultrasound images is a strong function of depth, measurements made with conventional test objects often produce large volumes of data. These may be useful to monitor equipment during its life cycle,^{1,2} but are restricted in their ability to make absolute measurements and to compare the performance of different ultrasound systems.

Pye and Ellis³ introduced the concept of a resolution integral and published preliminary work.⁴ This was the first description of a fundamental physical approach to combining measurements of resolution made at different depths in the ultrasound image. Since then, further work has been performed to validate the resolution integral and to characterise clinical and preclinical ultrasound scanners^{5–8} and endoscopic ultrasound systems.⁹ Rowland *et al.*¹⁰ used a test phantom containing a step-change in backscatter to make independent measurements of resolution integral in the scan and elevation planes.

In this paper, we present an overview of the resolution integral (R) and its two associated parameters; characteristic resolution (D_R) and depth of field (L_R) and describe the process of making resolution integral measurements using the Edinburgh Pipe Phantom. We use example values of R , D_R and L_R for different transducers, together with typical

images obtained using the Edinburgh Pipe Phantom, to show how D_R and L_R discriminate between transducers designed for different applications. We also show measurements made on scanners manufactured over a 20-year period to demonstrate the ability of the resolution integral to identify improvements in imaging technology. Finally, two illustrative case studies are used to show the versatility of R , D_R and L_R to quantify changes in performance, whether due to the development of faults within the probe or due to the selected processing pathways.

The aim of this paper is to give a clear and concise overview of the technique, so that users may obtain a better understanding of its strengths and weaknesses. Other publications relating to the resolution integral are referenced throughout the text, and the reader may wish to use these to obtain further in-depth information about aspects of the technique.

Definition of the resolution integral, characteristic resolution and depth of field

Pye and Ellis³ proposed the term 'resolution integral' as a single, dimensionless figure of merit for assessing the technical performance of ultrasound imaging systems. The resolution integral measures performance in terms of the ratio of beam penetration to beam-width. Highly sensitive transducers, driven by modern beam formers, are associated with a large penetration depth in combination with a narrow beam-width, resulting in large values of R . Conversely,

Figure 1 The graph (left) shows a plot of L versus α for an ultrasound beam (right) with low-contrast penetration L_0 and minimum beam width D_0 . α is the reciprocal of effective beam width and the resolution integral R is equal to the area under the curve. The graph (left) also illustrates depth of field L_R and characteristic resolution D_R . The area under the dashed curve $0-L_0-1/D_0$ is equal to the resolution integral R . The rectangle $0-L_R-S_R-1/D_R$ also has an area equal to R , and line $0-S_R$ bisects both the rectangle and the area under the dashed curve, giving $R = L_R/D_R$

a less-sensitive transducer, using a single piezoelectric element, is associated with less penetration depth and greater beam-width, resulting in smaller values of R . For a collimated beam, i.e. one for which the beam-width does not vary with depth, the calculation of the resolution integral is simply the ratio of the maximum penetration depth of the beam divided by the beam-width. However, ultrasound transducers do not produce collimated beams – their beam width changes continuously with depth due to diffraction and beam focussing. The calculation of resolution integral requires a series of measurements, which are used to form a curve of depth range versus $1/\text{beam-width}$ as shown in Figure 1. The resolution integral is then equal to the area under the curve. The experimental procedure to acquire this data is described below. The intercept of the curve on the y -axis is determined by the maximum penetration of the transducer (L_0 , otherwise known as low-contrast penetration, LCP) and on the x -axis by the best lateral resolution in the beam ($1/D_0$) i.e. minimum beam-width. Two other useful parameters are also defined in Figure 1, the ‘characteristic resolution’ and ‘depth of field’. The rectangle constructed on the graph in Figure 1 has the same area as the curve relating to the diagnostic transducer. Effectively, we have constructed a collimated beam with the same resolution integral (area under the curve) as the diagnostic beam. The construction of this rectangle is such that its diagonal bisects the area under the curve. The y -axis intercept of this rectangle (L_R) is defined as the depth of field and the x -axis intercept is $1/D_R$ where D_R is the characteristic resolution. The area of the rectangle is equal to the product of L_R and $1/D_R$, and hence we can express the relationship between them as $R = L_R/D_R$. In practice, the depth of field defines a region of best imaging, analogous to the focal region, and the characteristic resolution represents the typical lateral resolution within the depth of field.

The Edinburgh Pipe Phantom

The Edinburgh Pipe Phantom consists of a block of agar-based tissue mimicking material (TMM) containing a series of fluid-filled cylinders that are moulded into the agar during manufacture. The cylinders, or ‘pipes’, are inclined at 40° to the vertical and have diameters ranging from <100 microns to ~ 10 mm. The largest pipes extend over a depth range of 25 cm. The agar-based TMM is described by Ramnarine *et al.*¹¹ and has acoustical properties similar to those of soft tissue, with an attenuation coefficient of $0.5 \text{ dB cm}^{-1} \text{ MHz}^{-1}$ and speed of sound of 1540 m s^{-1} . The phantom is immersed in a water/glycerol and anti-bacterial solution with speed of sound 1540 m s^{-1} . The pipes are filled with this fluid, which also provides coupling to the upper surface of the phantom. A more detailed description of the manufacturing process can be found elsewhere.⁵

Measurement of the Resolution Integral

Measurement of the resolution integral is carried out in low-ambient lighting, at a level appropriate for clinical ultrasound scanning. As with all ultrasound quality tests, the serial numbers of the transducer and scanner are noted. The transducer is coupled to the surface of the phantom using the fluid in which the pipes are submerged. In general, the largest pipe is imaged first. The transducer is placed on the surface of the phantom, and the user orients the transducer so that the pipe is in the scan plane, and the upper region of the pipe is in the centre of the screen (Figure 2(a)). The transducer is manipulated and the controls optimised so that the upper region of the pipe is visualised as superficially as possible. Optimisation of the controls can include changing the field of view, transmit frequency, TGC, focal position, fundamental/harmonic imaging, speckle reduction and compound imaging.

Figure 2 Measurement technique for the resolution integral using the Edinburgh Pipe Phantom. (a) Superficial portion of pipe imaged and optimised in scan plane. (b) Mask with slot is moved along the length of the pipe image to find the most superficial/deepest position of the mask for which the visualised pipe appears to fill the slot. (c) Most superficial detectable position of pipe is identified and measurement taken from upper edge of slot to surface of phantom. (d) Image optimised again for deepest detectable position of pipe and step (b) is repeated. (e) Deepest detectable position of pipe identified and measurement taken from lower edge of slot to surface of phantom

The image is then frozen, and the linear callipers are selected from the measurement tools. When viewing the image of the pipe, a mask made from buff-coloured paper with a slot cut in it (Figure 2(b)) is used to reduce the visual impact of adjacent structures. The slot width is typically equal to 15 times the nominal wavelength of the imaging beam. For example, at 5 MHz, the ultrasound wavelength is 0.31 mm, and 15 wavelengths are equal to 4.6 mm. So, the width of the slot in the mask is chosen to be equal to 4.6 mm measured in the image. The mask is placed on the frozen image (Figure 2(a)) and moved along the pipe to allow the user to visualise each short section of the pipe, in turn, using peripheral vision (Figure 2(b)), and to identify the most superficial position at which the pipe can be detected. The callipers are then used to measure from the uppermost edge of the slot to the surface of the phantom in the image (Figure 2(c)). The scanner is then returned to live scanning and the transducer repositioned so that the deepest part of the pipe that can be visualised is in the centre of the screen, and the image is optimised again (Figure 2(d)). In most instances, this will include adjustment of the depth and focal plane, especially for the larger pipes, which are most likely to be visualised over the greatest depth. In this instance, the distance measured is from the bottom edge of the slot to the top of the phantom (Figure 2(e)). The distance between the superficial and deep measurements is equal to the depth range L , over which the pipe can be visualised. The value of L then forms the y co-ordinate of one experimental data point on the curve on the graph as shown in Figure 1. The x co-ordinate of this data point is equal to the inverse of the effective diameter of the pipe that is being visualised. The effective pipe diameter is calculated as the geometric mean of the pipe diameter in the image and elevation planes and is equal to $d/\sqrt{(\cos 40^\circ)}$ for a pipe of diameter d . In many instances, the depth over which the largest pipes can be visualised is equal to or slightly less than the LCP depth, which is the position of the deepest speckle identifiable from noise. Measurement of the LCP is performed in real time, as this allows the identification of speckle from random electrical noise. The dead zone of the image is determined by measuring from the surface of the transducer to the most superficial region of speckle that can be observed. The intersection of the curve with the y -axis is L_0 in Figure 1 and is equal to the LCP minus the depth of the dead zone. In general, LCP is much greater than the depth of the dead zone, and so $L_0 \approx \text{LCP}$. The intersection of the curve with the x -axis is the inverse of the effective diameter of the smallest pipe that can be visualised and is $1/D_0$ in Figure 1. For most transducers, measurements are taken using five or six pipes. The more pipes that are imaged, the more data-points there will be on the curve, and the more accurate the measurements of the resolution integral, characteristic resolution and depth of field.

Discriminating between transducers for different applications

The values of L_R and D_R measured for a specific transducer can be plotted on a graph of L_R and D_R such as the one shown in Figure 3. Figure 3 also gives a graphical summary

of the results obtained from measurements of over 150 clinical and preclinical transducers manufactured over a 25-year period.^{6,8} Depth of field is plotted as a function of characteristic resolution for four broadly defined applications – preclinical imaging, small parts imaging, endo-cavity imaging and general radiology (abdominal/obstetric) imaging. Note that from the relationship $R = L_R/D_R$, the resolution integral of a transducer is equal to the gradient of the line joining the (L_R , D_R) coordinates of the transducer to the origin. Lines of R equal to 25, 50 and 75 are drawn on Figure 3. Low values of resolution integral are indicative of older technology, whilst high values are indicative of higher performance and newer technology.

Example scanners and probes detailed in Table 1 have been included in Figure 3. As can be seen from Figure 3, preclinical transducers demonstrate excellent characteristic resolution (very small values of D_R) but a limited depth of field (very low values of L_R). This is consistent with the performance required of preclinical probes; high ultrasound resolution with minimal depth penetration, and the example in Figure 3 is an MS400 array transducer (Fujifilm, VisualSonics Inc, Ontario, Canada). Small parts transducers are some of the highest frequency clinical transducers, and the example in Figure 3 is a Siemens VF10-5 linear array (Siemens Healthcare, Erlangen, Germany), exhibiting low values of L_R and D_R . There is a substantial overlap in the graph between these probes and endo-cavity probes, the example shown being an Olympus GF-UE260 radial EUS probe (Olympus Corporation, Tokyo, Japan). Often, endo-cavity probes operate at comparable frequencies to linear probes; however, they must function within constrained environments; as such, they can have tight curvatures that can limit the values of L_R and D_R . Probes for general radiology imaging exhibit the largest range in characteristic resolution and depth of field. Such probes are required to image to depths of up to 20 cm within the abdomen and so tend to operate at lower frequencies, exhibiting large depth of fields and relatively large characteristic resolutions. The probe illustrated is the Siemens CH6-2 curvilinear array (Siemens Healthcare, Erlangen, Germany). Figure 3(b) contains example images of the 1.5 mm diameter pipe obtained using each of the four transducers.

Notwithstanding the overlap of some of the groupings, Figure 3 illustrates the effectiveness of the characteristic resolution and depth of field in discriminating between transducers designed for different imaging applications.

Determining improvements in transducer technology

Figure 4 illustrates the measurements of resolution integral for scanner/transducer combinations manufactured by one company (Siemens Healthcare) between 1986 and 2008. Older scanners are on the left of the bar chart and newer ones on the right. Each bar represents a single general radiology/abdominal transducer. It is evident from Figure 4 that as transducer technology develops over 20 years from the single element transducer of the late-1980s (Sonoline SX and Sonoline AC) to the modern broadband array (Antares Premium and S2000), the magnitude of R increases by

Figure 3 (a) Graph showing depth of field versus characteristic resolution. The regions associated with transducers for different applications are outlined and labelled: preclinical imaging, small parts imaging, endo-cavity imaging and general radiology imaging. Example results are given for one transducer designed for each application: (I) VisualSonics MS400, (II) Siemens VF10-5, (III) Olympus GF-UE260 and (IV) Siemens CH6-2. (b) Representative images of the 1.5 mm diameter pipe acquired using (I) VisualSonics MS400, (II) Siemens VF10-5, (III) Olympus GF-UE260 and (IV) Siemens CH6-2. The arrow on each image highlights the 1.5 mm pipe imaged within the Edinburgh Pipe Phantom. For comparison, a 1 cm scale bar is shown on each image

roughly a factor of two from 40–50 to >80. Recent measurements on array probes have demonstrated R values up to 100. In practice, this means that the lateral resolution of many modern transducers is better than half that of a single element transducer with the same penetration depth.

Performance of individual transducers

Fundamental versus harmonic imaging

It is also possible to study transducers being driven in different imaging modes. In Figure 5, a Sonosite 180 Plus

Table 1 Examples of transducers for each group in Figure 3

No.	Scanner	Probe	Type	Frequency (MHz)	Imaging Mode	Preset
I	VisualSonics Vevo 2100	MS400	L	30	Fundamental	Preclinical cardiology
II	Siemens G40	VF10-5	L	8	Harmonic	MSK
III	Aloka α 10	GF-UE260	EC	7.5	Fundamental	EUS
IV	Siemens Antares	CH6-2	CL	3.64	Harmonic	Abdominal

CL: curvilinear array; EC: endo-cavity; L: linear array.

Figure 4 Resolution integral values for general radiology transducers on a range of scanners manufactured by Siemens Healthcare. Older scanners are on the left of the chart, and the newer ones are on the right (date of manufacture in brackets)

C60/5-2 transducer is shown with the resolution integral calculated when the transducer is used first in fundamental imaging mode and second in harmonic mode. In harmonic mode, the transducer is driven at its fundamental frequency, but it is the signal which is at twice this fundamental frequency that is isolated on return to the transducer and used to form the image. From Figure 5, it is evident that by switching from fundamental to harmonic mode, the depth of field is slightly reduced (smaller value of L_R), and the characteristic resolution improves (smaller value of D_R). This is consistent with the concept that at higher frequencies, such as at the second harmonic, resolution is improved but the depth of penetration is reduced. Note that in this instance, there is little change in resolution integral in changing from fundamental to harmonic imaging i.e. the lines joining both points to the origin have a similar gradient.

Working versus faulty

The resolution integral can also be used to measure a deterioration in imaging performance e.g. a faulty transducer. Faults can exhibit as an increase in characteristic resolution (larger D_R) and/or a decrease in depth of field (smaller L_R), usually accompanied by a decrease in resolution integral.

Figure 5 shows measurements of a Siemens linear array probe, both when it was functioning correctly, and after it had been damaged. In this example, the transducer was initially imaging well but had a damaged strain relief.

The strain relief was repaired by a third-party company, but when the transducer was returned to clinical use, the operators felt that its imaging had deteriorated, although there were no visible signs of failed elements in the image. Another series of measurements were made using the Edinburgh Pipe Phantom, and D_R was found to have increased i.e. the transducer could no longer image small pipes to the same extent. The probe was withdrawn from service and reimbursement obtained from the repair company.

Discussion

The resolution integral combines a number of ultrasound image parameters (resolution at different depths, low-contrast penetration and dead zone) in a physically meaningful way. Figure 2 illustrates the technique of measuring the resolution integral, while the results shown in Figure 4 demonstrate that R is able to characterise improvements in ultrasound imaging technology, where the values of R for a series of scanners produced by a manufacturer increase between older models and modern high-end equipment. Figure 3 shows where transducers designed for preclinical, small parts, endo-cavity and general radiology imaging appear on a graph of L_R versus D_R . Transducers designed for each clinical application group together in different parts of the graph, and although there is some overlap, it is evident that transducers for specific applications have

Figure 5 Depth of field versus characteristic resolution for two transducers: a Sonosite 180 Plus C60/5-2 driven in fundamental imaging mode (●) and then in harmonic imaging mode (○) and a Siemens Elegra Advance VF13-5 when working normally (◇) and after developing a fault (◆)

similar characteristic resolution and depth of field. We have also illustrated how the use of the resolution integral can enable the objective assessment of diagnostic imaging performance and that faulty probes and different signal-processing techniques can be identified and quantified. R , L_R and D_R characterise technical performance in terms of sensitivity and resolution but are not direct indicators of diagnostic accuracy.

Conclusions

The resolution integral, measured using the Edinburgh Pipe Phantom, is able to discriminate between ultrasound-imaging transducers with differing levels of performance. The depth of field L_R and characteristic resolution D_R provides a quantitative summary of the imaging characteristics of individual transducers that can be concisely displayed on a plot of L_R versus D_R .

DECLARATIONS

Competing interests: The authors have no conflicts of interest to declare.

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Guarantor: CM.

Contributorship: SP developed the concept of the Resolution Integral and developed the Edinburgh Pipe Phantom. SP, SI and CMM performed the resolution integral measurements. CMM wrote the first draft of the manuscript; SI drew the figures. All authors reviewed and approved the final version of the manuscript.

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