The Edinburgh Pipe Phantom: characterising ultrasound scanners beyond 50 MHz

C M Moran¹, W Ellis², A Janeczko², D Bell³ and S D Pye²

¹ Medical Physics, University of Edinburgh, Edinburgh, EH16 4TJ, UK
² Medical Physics Department, NHS Lothian University Hospitals Division, Royal Infirmary, Edinburgh EH16 4SA, UK
³ Precision Acoustics Ltd, Hampton Farm Business Park, Dorset, DT2 8QH, UK

E-mail: carmel.moran@ed.ac.uk

Abstract. The ability to measure the imaging performance of pre-clinical and clinical ultrasound scanners is important but difficult to achieve objectively. The Edinburgh Pipe Phantom was originally developed to assess the technical performance of clinical scanners up to 15MHz. It comprises a series of anechoic cylinders with diameters 0.4 – 8mm embedded in agar-based tissue mimic. This design enables measurement of the characteristics (Resolution Integral $R$, Depth of Field $L_R$, Characteristic Resolution $D_R$) of grey-scale images with transducer centre frequencies from about 2.5 to 15MHz. We describe further development of the Edinburgh Pipe Phantom as a tool for characterising ultrasound scanners with centre frequencies up to at least 50MHz. This was achieved by moulding a series of anechoic pipe structures (diameters 0.045 – 1.5mm) into a block of agar-based tissue mimic. We report measurements of $R$, $L_R$ and $D_R$ for a series of 10 transducers (5 single element and 5 array transducers) designed for pre-clinical scanning, with centre frequencies in the range 15-55 MHz. Values of $R$ ranged from 18-72 for single element transducers and 49-58 for linear array transducers. In conclusion, the pre-clinical pipe phantom was able to successfully determine the imaging characteristics of ultrasound probes up to 55MHz.

1. Introduction

The ability to measure the imaging performance of both pre-clinical and clinical scanners is important but difficult to achieve objectively. We have previously shown the versatility of using the Resolution Integral as a figure-of-merit for characterising clinical scanners [1,2]. More recently we have extended this work to include the quantitative assessment of scanners and transducers for pre-clinical applications [3]. In this paper we employ the Resolution Integral to compare 10 transducers designed for pre-clinical applications: 5 single element transducers ranging in centre frequency from 25-55MHz and 5 linear array transducers ranging in centre frequency from 15-40MHz.

The Resolution Integral, $R$, has been defined elsewhere [4] and will only briefly be described here. Essentially $R$ is the ratio of the penetration of an ultrasound beam in soft tissue to the ultrasound beam width. Transducers that perform well combine good penetration with narrow beam width, and thus $R$ will be large. Two other parameters are also defined in [4], Depth of Field $L_R$, and Characteristic Resolution $D_R$. The Depth of Field defines the axial extent of a region of optimum imaging and the Characteristic Resolution is representative of the resolution within the Depth of Field, taking into...
consideration both the lateral and elevation planes. The three quantities are related by the expression \( R = \frac{L_R}{D_R} \).

2. Methods

Measurements of \( R, L_R \) and \( D_R \) were made on 10 transducers from two different pre-clinical ultrasound scanners. Five linear array transducers on a Vevo 2100 scanner were tested, and also 5 single element transducers on a Vevo 770 scanner (Visualsonics Inc, Toronto, Canada). The manufacturer’s data is summarised in Table 1. Further details of the measurement method described here are contained in [2] and [4].

Measurements were made using two Edinburgh Pipe Phantoms [1] both composed of agar-based tissue mimicking material (TMM) with attenuation coefficient 0.5dBcm\(^{-1}\) MHz\(^{-1}\) and speed of sound of 1540ms\(^{-1}\) determined at frequencies up to 20MHz by [5]. A series of anechoic wall-less pipes were moulded within the TMM at an angle of 40\(^0\) to the vertical. The first (“clinical”) pipe phantom was constructed from a block of TMM (250 x 250 x 100mm) contained in a perspex casing and with pipe diameters from 0.35mm to 7.9mm. The second (“pre-clinical”) pipe phantom comprised a series of cylindrically-shaped blocks of TMM with diameter 60mm and height 40mm. Embedded within these cylinders were pipes of diameter 1470, 550, 330, 193, 139, 92, 68 and 45 microns. The TMM of both phantoms was immersed in a water and glycerol mixture (speed of sound 1540ms\(^{-1}\)) which filled the pipes and provided coupling to the upper surface of the phantoms.

The lowest frequency transducers (15 and 21MHz) were evaluated using both the clinical and pre-clinical pipe phantoms. The other transducers were all evaluated using the pre-clinical pipe phantom. To carry out measurements, each transducer was placed on the surface of the phantom so that the axis of the largest pipe lay in the scan plane. The image was then optimised to ensure that the upper section of the pipe was visualised as superficially as possible. For the array transducers, a minimum of two focal positions were chosen around the location of the upper section of the pipe. The image was frozen and the minimum depth at which the pipe could be visualised was measured using on-screen callipers. The transducer was then re-positioned to enable the lower section of the pipe to be visualised as deep as possible within the phantom. Again, a minimum of two focal positions were chosen and located around this depth. The image was frozen and the maximum depth at which the pipe could be visualised was measured using on-screen callipers. To determine the locations of the most superficial and deepest imaged extent of the pipe, a mask was manufactured containing a slot of approximate width 15 times the wavelength of the centre frequency of the transducer. This was placed over the frozen image and the slot moved along the extent of the pipe to establish the positions where the pipe could be confidently visualised with respect to surrounding speckle at the same depth. The difference between the most superficial and the deepest imaged depths is \( L \) (figure 1). Three sets of measurements were taken from three pipes of the same diameter, and the mean of these was calculated. The procedure was repeated for each pipe and each mean value of \( L \) was plotted against the appropriate value of \( \alpha \). \( \alpha \) is the inverse of the effective pipe diameter, calculated as the geometric mean of the pipe diameter in the image and elevation planes, and given by \( d/\sqrt{(\cos 40^0)} \) where \( d \) is pipe diameter. The low contrast penetration depth in the image was identified as the depth at which speckle could just be differentiated from system noise. The dead zone was identified as the most superficial depth at which speckle could be imaged. The axial distance between the dead zone and the low contrast penetration is given by the intercept of the \( L-\alpha \) curve with the ordinate in figure 1. The area under the curve is equal to the Resolution Integral [4].

The same procedure was followed for the single element transducers except that the deepest and most superficial positions where each pipe could be visualised were both measured using the same optimised image. In addition, an alternative method of measuring the Resolution Integral was employed for the RMV704 single element transducer, whereby the distance between the transducer and surface of the phantom was varied to enable the extent of the pipe to be tracked through the depth of the phantom. This technique of tracking the pipes throughout the phantom is consistent with the recommended use of the single element transducers for pre-clinical scanning, where coupling gel is
often used as a stand-off to ensure that superficial regions of interest can be located within the focal region.

| Transducer | Single element (SE) or linear array (LA) | Centre Frequency (MHz) | Focal-Length (mm) | Field of View (mm) |
|------------|------------------------------------------|------------------------|-------------------|-------------------|
| RMV710     | SE                                       | 25                     | 15                | 20                |
| RMV707B    | SE                                       | 30                     | 12.7              | 20                |
| RMV704     | SE                                       | 40                     | 6                 | 14.6              |
| RMV708     | SE                                       | 55                     | 4.5               | 10.7              |
| RMV711     | SE                                       | 55                     | 6                 | 8.5               |
| MS200      | LA                                       | 15                     | 18                | 36                |
| MS250      | LA                                       | 21                     | 15                | 30                |
| MS400      | LA                                       | 30                     | 9                 | 20                |
| MS550S     | LA                                       | 40                     | 6                 | 15                |
| MS550D     | LA                                       | 40                     | 7                 | 15                |

Table 1. Manufacturer’s data for the 10 pre-clinical transducers.

Figure 1. Plot of $L$ (depth range over which pipe can be visualised) against $\alpha$ (1/effective pipe diameter) for a Vevo 2100 MS550S linear array centred at 40MHz.
3. Results

The $L$-$\alpha$ curve for each transducer typically comprised 5 to 7 data points. Figure 1 shows an example obtained using the MS550S 40MHz linear array. The area under the curve is equal to the Resolution Integral. Figure 2 illustrates two sets of results obtained from the RMV704 40MHz single element probe: one with the probe resting on the surface of the phantom, and one obtained using the tracking technique. The smallest diameter pipes (45µm) were not imaged by any of the pre-clinical transducers.

Values of $L_R$ and $D_R$ for each of the 10 transducers were calculated from their $L$-$\alpha$ curves and plotted on a diagram of $L_R$ vs $D_R$ as described in [4]. The results are shown in figure 3. The value of $R$ for each transducer is equal to the gradient of the line joining the point $(D_R, L_R)$ to the origin, since $R = L_R/D_R$. The values of $L_R$ and $D_R$ calculated from the three $L$-$\alpha$ curves shown in figures 1 and 2 are labelled in figure 3 as “40MHz (S)”, “40MHz(a)” and “40MHz(b)”.

4. Discussion

It can be seen in figure 3 that the 5 single element transducers all had values of $R$ close to 25 (range 18 - 25), with the exception of the tracked measurements for the RMV704 where $R=72$. For the linear array transducers, values of $R$ were close to 50 (range 49 - 58). The smallest Characteristic Resolution (93µm) was observed for the RMV704 tracked measurements. The points representing the two linear array transducers with centre frequencies of 40MHz (MS550D and MS550S) lie close together, with $R$ values of 55 and 56 respectively. The MS550D had a slightly longer Depth of Field (10.9mm) than the MS550S (10.5mm) which is consistent with the manufacturer’s recommendations that the MS550D transducer should be used for imaging slightly deeper structures. The RMV704 single element probe with the same nominal centre frequency of 40MHz had a significantly different performance, with a smaller Depth of Field (3.3mm) and Characteristic Resolution (135µm) compared to 188µm and
Figure 3. Measured values of \( L_R \) and \( D_R \) for 10 pre-clinical transducers. ▲ denotes Vevo 2100 linear array transducers [40MHz (S) = MS550S, 40MHz (D) = MS550D]. ○ denotes Vevo 770 single element transducers [55MHz(a) = RMV708, 55MHz(b) = RMV711, 40MHz(a) = RMV704, ● 40MHz(b) = RMV704 tracked through the phantom]. Lines of constant \( R \) are also shown (\( R = 25, 50, 75 \)). Expanded uncertainty in \( L_R \) and \( D_R \) (95% confidence) is approximately ±8%.

197\( \mu \)m for the MS550S and MS550D respectively. This is due to the greater degree of focussing of the single element transducer, yielding smaller values of Characteristic Resolution and Depth of Field for the same nominal centre frequency. The same comparison can be made between the single element and linear array transducers with a nominal centre frequencies of 30MHz: the RMV707B with \( D_R = 225 \mu \)m, \( L_R = 5.3 \)mm, and the MS400 with \( D_R = 269 \mu \)m, \( L_R = 13.2 \)mm.

Using the tracking technique with the RMV704 ensured that all the superficial aspects of the pipe could be visualised within the focal region, thus effectively lengthening the Depth of Field to 6.7mm, compared to 3.3mm when the tracking technique was not applied. In addition, since each superficial aspect of the pipe was visualised at optimal resolution in the focal region, the Characteristic Resolution was reduced from 135\( \mu \)m to 93 \( \mu \)m. The Resolution Integral increased from 25 to 72, which is the highest of all the pre-clinical transducers and comparable with values achieved by premium range clinical scanners [4].

The shapes of the \( L-\alpha \) curves for the pre-clinical linear array transducers (e.g. the MS550S shown in figure 1) were all close to linear. This is very similar to the shape of the \( L-\alpha \) curves reported for clinical scanners in [4], and indicates a weakly focussed ultrasound beam. In contrast, the results for the RMV704 in figure 2 show two different curve shapes. The lower curve, obtained with the probe
resting on the surface of the phantom, is distinctly concave. This is indicative of the relatively strong focussing of the single element transducer. The upper curve, produced by tracking the focal region through the phantom, is convex and approaches the rectangular “box” shape indicative of a collimated beam [2].

5. Conclusions
Measurements of $R$, $L_R$ and $D_R$ have been carried out on 10 pre-clinical ultrasound transducers, and the characteristics of single element and linear array technologies have been compared. The results demonstrate that the clinical and pre-clinical pipe phantoms, incorporating anechoic targets from 45µm to 8mm diameter, are able to quantify and differentiate the characteristics of transducers with centre frequencies up to at least 55MHz. The characteristics of the pre-clinical transducers reported here can also be compared with a range of clinical transducers reported in [4]. It may be possible to extend Resolution Integral measurements to transducers above 55MHz, given that the smallest pipes were not imaged by any of the pre-clinical transducers.

References
[1] Pye S D and Ellis W 2002 UK Patent GB2396213
[2] Pye S D, Ellis W and MacGillivray 2004 Journal of Physics: Conf. Series 1 187-92.
[3] Moran C M, Smart S, Ellis W and Pye S D 2008 Proc 2008 IEEE Ultrasonics Symposium pp1724-27.
[4] Pye S D and Ellis W 2010 Journal of Physics: Conf. Series (this issue)
[5] Brewin M P, Pike L C, Rowland D E and Birch M J 2008 Ultrasound in Med and Biol 34 1292-1306