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Evaluation of Intravascular Ultrasound Catheter Based Transducers using the Resolution Integral

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Abstract: Intravascular ultrasound (IVUS) catheters are a specialist imaging modality used in the assessment of cardiovascular disease. The ultrasound transducer may either be of single element mechanical or phased array design. Due to their design and operating frequencies (10-45 MHz), evaluation of the imaging performance is not possible with commercially available ultrasound test objects. A modification of an existing test object, the Edinburgh Pipe Phantom (EPP), was carried out to allow measurement of resolution integral (R), depth of field (Lr) and characteristic resolution (Dr) of IVUS catheters. In total seven IVUS catheters, from two manufacturers and of both single element mechanical and phased array design, were tested to provide a measure of performance over different frequencies and technologies. Measurements of R for the tested IVUS catheters ranged from 11.6 to 18.8. The modified EPP therefore allows catheter based ultrasound probes to be evaluated scientifically and their performance to be seen in relation to other similar ultrasound technologies such as preclinical ultrasound and endoscopic ultrasound.

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1 **Title Page**

2

3 **Evaluation of Intravascular Ultrasound Catheter Based Transducers using the**
4 **Resolution Integral**

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10

11 **Abstract**

12

13 Intravascular ultrasound (IVUS) catheters are a specialist imaging modality used
14 in the assessment of cardiovascular disease. The ultrasound transducer may either
15 be of single element mechanical or phased array design. Due to their design and
16 operating frequencies (10-45 MHz), evaluation of the imaging performance is not
17 possible with commercially available ultrasound test objects. A modification of an
18 existing test object, the Edinburgh Pipe Phantom (EPP), was carried out to allow
19 measurement of resolution integral (R), depth of field (Lr) and characteristic
20 resolution (Dr) of IVUS catheters. In total seven IVUS catheters, from two
21 manufacturers and of both single element mechanical and phased array design,
22 were tested to provide a measure of performance over different frequencies and
23 technologies. Measurements of R for the tested IVUS catheters ranged from 11.9 to
24 18.8. The modified EPP therefore allows catheter based ultrasound probes to be
25 evaluated scientifically and their performance to be seen in relation to other similar
26 ultrasound technologies such as preclinical ultrasound and endoscopic ultrasound.

27

28 **Key Words**

29 Intravascular Ultrasound, Phantom, Resolution, IVUS, Transducers, Evaluation,
30 Equipment, Resolution Integral, Imaging Performance

31 **Introduction**

32

33 The ever growing and diverse range of applications of diagnostic ultrasound has
34 led to the development of a range of specialised ultrasound transducers. Many of
35 these transducers do not undergo routine performance testing due either to
36 limitations of target location and size in commercially available performance test
37 phantoms or the inability to obtain useful and meaningful results from these
38 phantoms. One such technology is high frequency catheter based ultrasound
39 imaging. This includes: intravascular ultrasound (IVUS), transurethral ultrasound
40 (TUUS) (Ingrama et al. 2003) and intraductal ultrasound (IDUS) catheters (Levy et
41 al. 2002). Despite different clinical applications these transducers share design
42 similarities and are either single element mechanical or phased array micro
43 ultrasound transducers contained within catheters of diameters less than 3mm.
44 Catheter based transducers typically do not have adjustable operating frequencies.
45 Instead different models come at a stated centre frequency appropriate for the
46 anatomy of interest. A range of commercial IDUS and TUUS probes are available
47 between 10 MHz to 30 MHz, while those available for IVUS range from 10 MHz to 45
48 MHz. Clinically, IVUS is employed to give a 360° view of blood vessel structures for
49 the assessment of cardiovascular disease and stent placement.

50

51 The primary focus of test techniques and published guidance on grey-scale
52 imaging to date has been with respect to clinical ultrasound operating at frequencies
53 below 20MHz. These frequencies are typical of those used on clinical radiology and
54 cardiology ultrasound scanners. These test techniques are used for regular quality
55 assurance protocols (Dudley et al. 2014; Hangiandreou et al. 2013; Russell 2010)

56 and also to assess new technologies and provide a technical justification for
57 equipment purchases (Pye et al. 2011). The non-traditional shape of catheter based
58 transducers and their higher operating frequencies make the testing procedures for
59 the acquisition of meaningful performance data from commercial phantoms difficult
60 to obtain. Previous published work from Hoskins et al. (1994) and Elliott et al. (1996)
61 describes test procedures for assessing the performance of single element
62 mechanical transducers (axial, lateral and out of plane resolution and penetration
63 depth). Both employed high contrast targets; Elliott et al. (1996) used polyester
64 filaments in water to measure resolution via the Rayleigh criteria while Hoskins et al.
65 (1994) used nylon filament in water to measure resolution using the point spread
66 function. Hoskin et al. (1994) also utilised a tissue mimicking material to measure
67 low contrast penetration. Some studies have focused on the variability and reliability
68 of calliper measurements made with both single element mechanical and phased
69 array IVUS transducers. For example, Tardif et al. (2000) demonstrated that both
70 single element mechanical and phased array IVUS probes demonstrated low
71 variability in a variety of measurements within a veterinary/clinical setting. Artefacts
72 unique to IVUS transducers such as non uniform rotational distortion (NURD) have
73 been reported. This non uniform rotation of cable-driven single element mechanical
74 transducers leads to a misregistration of the received image lines (Kimura et al.
75 1996). The effect has also been assessed in comparison to optical coherence
76 tomography (OCT) (Kawase et al. 2007). When assessing accuracy and/or the
77 extent of NURD artefact, a stent phantom has typically been employed. These are
78 constructed of metal wires in a circular distribution and are analogous to the calliper
79 measurement test objects that will be familiar to many Medical Physics, Biomedical
80 and Clinical Engineering departments. The incorporation of a grey-scale test object

81 to characterise beam shape and the imaging capabilities of IVUS transducers would
82 complement the current methods for assessing image registration accuracy. The
83 single use nature of IVUS catheters reduces the advantages of a routine testing
84 protocol. Instead a strong technical evidence base would be of benefit as
85 justification of current equipment's capabilities and for equipment evaluation prior to
86 procurement. This is of particular importance as IVUS equipment is routinely
87 integrated into the catheterization laboratory potentially locking the hospital into a
88 single supplier.

89

90 The Edinburgh Pipe Phantom (EPP) has previously been used to demonstrate
91 improvements in imaging technology over time and to distinguish transducers suited
92 to particular clinical applications (MacGillivray et al. 2010; Pye & Ellis 2011; Inglis et
93 al. 2014). The EPP contains fluid filled anechoic pipe structures that are of various
94 diameters within an agar-based tissue mimicking material (TMM). Each pipe is
95 imaged to determine the depth range L over which the pipe is visible following the
96 procedure described in MacGillivray et al. (2010). The depth range L is plotted as a
97 function of the inverse of effective pipe diameter α . The effective pipe diameter is
98 taken as the geometric mean of the pipe diameter in the imaging and elevation
99 planes. Plotting $L(\alpha)$ generates a curve that is dependent on the specific
100 transducer's beam shape (Figure 1). The resolution integral (R) is defined as the
101 area under the curve and also corresponds to the ratio of penetration to lateral
102 resolution (MacGillivray et al. 2010). In addition to the resolution integral, two
103 additional metrics are derived from the $L(\alpha)$ curve: characteristic resolution and depth
104 of field. The depth of field (L_r) defines the axial extent of a region of optimum
105 imaging, analogous to the focal region. The characteristic resolution (D_r) is

106 representative of the beam width within the depth of field. These three parameters
107 are related by the equation $R=L_r/D_r$. Generated from one set of measurements, R,
108 D_r & L_r are used to characterise and compare different transducer types and
109 systems. Other test methods often generate large data sets from which it can be
110 difficult to determine the relative strength and weakness of different transducers and
111 systems (Madsen et al. 2000; Thijssen et al. 2007). The EPP employs anechoic
112 targets, the detection of which is critical in many clinical applications. This differs to
113 the high contrast filament methods used for example by Hoskins et al. (1994) and
114 Elliott et al. (1996). The key contribution of the work described here is that
115 measurements obtained from a range of depths using IVUS transducers are
116 combined in a physically meaningful way, folding in the effects of lateral and
117 elevation resolution, speckle, low contrast penetration, scan line density, electrical
118 noise and pixel interpolation. This allows them to be compared both with other IVUS
119 transducers and with other clinical and pre-clinical systems.

120

121 Modification of the EPP for high frequencies using small diameter pipe structures
122 as described in Moran et al. (2011a) would be appropriate for IVUS transducers.
123 The acoustic properties of the IEC agar-based TMM used to construct the EPP have
124 been well characterised over a frequency range of 2 to 60 MHz (Rabell et al. 2017;
125 Rajagopal et al. 2014; Sun et al. 2012) making it suitable for use at typical IVUS
126 frequencies. Some modifications to the EPP design were required to make it better
127 suited for use with IVUS transducers. For example the ten anechoic pipe structures
128 within the original EPP have diameters ranging from 0.42 to 7.9 mm which is an
129 appropriate range for probes at frequencies from 1-20 MHz. With IVUS frequencies
130 ranging from 10-45 MHz, smaller pipe diameters were required to adequately test

131 the resolution limit of these transducers. Similarly the depth of the phantom could be
132 reduced as penetration can be expected to be much less at these high frequencies.
133 Importantly, the ability to be able to hold and support the delicate catheters is a
134 necessary feature of a modified EPP for IVUS imaging. To that end the aim of this
135 study was to design and manufacture a modified EPP and use it for characterising
136 various intravascular ultrasound catheter based transducers.

137

138 **Methods**

139

140 Like the original EPP, the modified EPP was an agar based TMM which
141 incorporates anechoic pipes structures set at 50° to the scan surface. These
142 anechoic pipes structures were created from an initial mould using metal rods, wires
143 and monofilament sutures of appropriate diameter traversing a plastic container, into
144 which the TMM mixture was poured and set (Pye et al. 2011, Moran et al. 2011b).
145 Care was taken to ensure that metal wires and sutures were taut across the plastic
146 container, secured and sealed to the exterior of the container to prevent leaking of
147 the TMM before it set. The schematics for this are shown in Figure 2, with ten pipes
148 ranging in diameter from 0.044 mm to 1.5 mm spread across several small TMM
149 blocks (64 x 40 x 20 mm). When the TMM blocks had set, the rods, wires and
150 sutures were removed to leave two pairs of anechoic pipes within each block. The
151 blocks were then removed from the mould casing and small cuts made at the corner
152 or edge of each block to uniquely identify them. The blocks were stored in a water-
153 glycerol mix to allow the pipes to fill with fluid and prevent the TMM from drying out.
154 The pipe diameters were chosen to encompass a range of sizes, from diameters
155 large enough to be imaged down to the low contrast penetration (LCP) depth, to

156 those too small to be detected by the IVUS catheters. This provided a range of data
157 points covering the performance of any given IVUS catheter.

158

159 A plastic container was manufactured to accommodate the scanning of TMM
160 blocks, holding two at a time. Guide channels were cut into the container so that
161 plastic guide blocks could move freely along either side of the TMM blocks. These
162 guide blocks had a channel cut to allow a catheter to pass between them and be
163 held securely above the TMM. Confirmation of pipe patency was obtained by
164 scanning the TMM blocks with a VisualSonics Vevo 770 preclinical scanner using a
165 30 MHz RMV704 probe. The Vevo 770 allowed 3D scans of the pipes as shown in
166 Figure 3. All pipes down to 0.073 mm diameter were imaged using this scanner to
167 ensure that they were free from manufacturing defects.

168

169 Two TMM block groups were created from one batch of TMM. These were
170 labelled block groups A & B and used to perform tests on the seven probe models
171 shown in Table 1. To conduct each experiment the IVUS probe was inserted into a
172 channel cut into the guide blocks fixing it above the TMM blocks and allowing
173 positioning to be done via movement of the guide blocks. An example of this can be
174 seen in Figure 4. With the transducer placed over the top of the pipe to be imaged
175 the operator is free to optimise scanner settings to obtain the best image of the pipe.
176 For the Boston Scientific Opticross probe, four sets of measurements were obtained
177 on both group A and group B. All other probes had two sets of measurements
178 recorded from group A and two from group B. Due to the lower frequency of the
179 Volcano Visions PV.035 10 MHz probe, four sets of additional data points were
180 acquired by imaging the larger pipes within the original EPP (diameters: 1.0 mm, 2.0

181 mm and 3.0 mm). These data points were combined with data obtained from the 1.5
182 mm pipe and LCP measurements on the modified EPP, to provide an overall
183 measurement.

184

185 Following the method described in MacGillivray et al. (2010), each pipe was
186 scanned with depth and gain optimized so as to image the superficial portion of each
187 pipe, closest to the surface of the catheter. For pipes larger than the beam width the
188 visible portion was seen up to the surface of the TMM. On smaller pipes the
189 superficial portion of the pipe disappeared into the surrounding speckle before
190 reaching the surface of the TMM. The point at which this occurred was dependent
191 on the relation between the pipe diameter and the probe beam width. The images
192 were saved to allow measurements to be made off-line. With the aid of the guides,
193 the catheter was then moved to locate the deepest visible position of each pipe i.e.
194 the portion of pipe that was just distinguishable from the surrounding speckle. The
195 viewing depth and gain were adjusted as necessary, and the images saved for off-
196 line measurement. This process was repeated for the upper and lower portions of
197 each pipe, swapping out the pipe blocks until all the pipes had been imaged. The
198 LCP was determined by measuring the distance from the surface of the TMM down
199 to the deepest speckle identifiable from noise. This was done in real time in a
200 section of TMM where no pipes were visible in the field of view. For the majority of
201 the transducers tested the images could be exported to USB/Disc in DICOM format
202 to allow measurement on a review station. However to obtain images from the
203 Boston Scientific Clearview with Atlantis SR Pro and Pro² probes, an 8 bit, 20 MHz
204 image capture card (PicPort, CameraTec AG, Weisslingen, Switzerland) was used.
205 Measurements of the depth range L over which the pipe is visible were obtained by

206 subtracting the most superficial visible depth from the deepest visible depth (Figure
207 5). This was carried out using ImageJ (U. S. National Institute of Health, Bethesda,
208 Maryland). If required, images were rotated in ImageJ so that the pipe was
209 visualised below the centre point of the image. A piece of thin card into which a slot
210 had been cut to form a viewing aperture was moved along the length of the pipe to
211 allow the observer to compare the pipe with adjacent speckle and identify the most
212 superficial and deepest depths at which the image of the pipe could be confidently
213 detected from background speckle. The width of the slot size varied with probe
214 frequency and the scaling of the image on the screen and was calculated using the
215 protocol outlined in MacGillivray et al. (2010). For each pipe, measurements were
216 averaged and the visible depth range L of each pipe was calculated. A plot of L
217 versus α was then made (following Figure 1) and D_r , L_r and R were calculated using
218 the method given in MacGillivray et al. (2010). The reported 95% confidence interval
219 for D_r , L_r and R was calculated as the standard error of the sample mean multiplied
220 by a coverage factor based on the number of results obtained for that transducer
221 (UKAS, 2012).

222

223 **Results**

224

225 All pipes in the modified EPP were observed to be patent and the surrounding
226 TMM had no visible artefacts caused by solid or gaseous inclusions when assessed
227 on the Vevo 770 scanner. The resolution integral of all the IVUS probes tested using
228 the modified EPP are shown in Table 1 and range from 11.9 to 18.8, L_r and D_r are
229 also shown.

230

231 The Atlantis SR Pro and Atlantis SR Pro² probes on the Boston Scientific
232 Clearview are both 40 MHz single element mechanical probes and both detected
233 measurable lengths of five different pipe diameters, the smallest of which was 0.19
234 mm. The Boston Scientific Opticross on the Polaris system was also a 40 MHz
235 single element mechanical probe which imaged all the pipes down to the 0.19 mm
236 pipe. As can be seen from Figure 6, the Atlantis SR Pro and Pro² are almost
237 identical in terms of performance given the overlap of the measurement
238 uncertainties. The Opticross imaged less of the length of the 1.5 mm pipe compared
239 to the two Atlantis probes and had a smaller LCP resulting in a smaller Lr compared
240 to the Atlantis probes. The Volcano Revolution Rotational 45 MHz Imaging probe
241 was a single element mechanical probe that performed similarly to the Boston
242 Scientific single element mechanical probes imaging down to the 0.19 mm pipe. The
243 Volcano S5 system also offered three phased array probes to be tested. The Eagle
244 Eye platinum was a phased array design with a stated frequency of 20 MHz; it
245 imaged four pipes the smallest of which was 0.33 mm. The Visions PV .018 with a
246 stated frequency of 20 MHz was also a phased array design and imaged three pipes,
247 the smallest of which was 0.33 mm. The Visions PV.035 10 MHz probe detected
248 only the 1.5 mm pipe in the modified EPP. To provide suitable data points to
249 calculate the resolution integral additional pipes from the original EPP were scanned
250 with the Visions PV.035 probe. Compared to the single element mechanical probes
251 the phased array probes recorded the largest Lr and Dr values.

252

253 There was a greater uncertainty in the measurement of the resolution integral for
254 the Atlantis SR Pro and Visions PV0.18; $18.8 \pm 13\%$ and $14.2 \pm 14\%$ respectively.
255 This occurred from differences in resolution integral measurements made with TMM

256 block group A and TMM block group B. For the Atlantis SR Pro the resolution
257 integral was consistently higher for group B compared to group A, the Visions
258 PV.018 showed the reverse pattern. In each case group A was scanned first
259 immediately followed by group B. Other transducers showed no difference between
260 the TMM blocks and had 95% confidence interval typically $\pm 6\%$.

261

262 **Discussion**

263

264 The modified EPP design implemented for the evaluation of IVUS transducers in
265 this study proved to be effective. The size and range of pipes included in the
266 modified EPP design were best suited to testing catheter based transducers with
267 frequencies greater than 20 MHz. As the 10 MHz probe was only able to detect one
268 pipe the phantom could benefit from one or more larger pipes when low frequency
269 (10-30 MHz) catheter transducers such as those used in endoscopy are of interest.

270

271 The 10 MHz and 20 MHz phased array IVUS probes have greater depth of field
272 ($L_r > 5$ mm) and larger characteristic resolution ($D_r > 0.4$ mm) as one might expect
273 compared to the 40 MHz and 45 MHz single element mechanical IVUS probes. The
274 Eagle Eye Platinum 20 MHz phased array probe with a D_r of 0.39 mm and L_r of 5.9
275 mm places it in between the two aforementioned groups. The Atlantis SR Pro and
276 Atlantis SR Pro² displayed nearly identical results, evidence perhaps that these are
277 equivalent probes with small design/manufacturing differences.

278

279 The higher measurement uncertainty on the Atlantis SR Pro and Visions PV .018
280 probes is possibly the result of a lack in operator familiarity with the scanners, as

281 they were the first probes to be tested on the respective IVUS scanners, which
282 operate differently from clinical radiology scanners. No drop off in measured image
283 performance of the IVUS catheters occurred as might have been expected from
284 single use devices, although each probe's scanning time totalled up to 8 hours
285 across 2 to 3 days. A 95% confidence interval of $\pm 6\%$ is higher than the $\pm 2.6\%$
286 (MacGillivray et al. 2010) in measured value of R reported for linear and curvilinear
287 probe's using the original EPP, again potentially highlighting the challenges of testing
288 IVUS probes. Due to the design of the guide blocks, the process of scanning the
289 TMM blocks was easier to implement than manual scanning of the IVUS probe on
290 the original EPP. Manual scanning on the original EPP led to the catheters
291 becoming bent as a result of the constant handling and scanning angle.

292

293 In Figure 7, results for the single element mechanical IVUS transducers are
294 displayed alongside results from previous studies that used the EPP method to
295 characterise single element mechanical transducers used in preclinical work (Moran
296 et al. 2011b) and endoscopic ultrasound (Inglis et al. 2014). Given the physical
297 constraints of placing transducers within the body of a catheter it is not surprising to
298 find the magnitude of the resolution integral of the IVUS catheters (range 15-19) to
299 be slightly less but comparable to values for single element preclinical probes (range
300 19-25). A common imaging artefact in IVUS, non uniform rotation distortion (NURD)
301 was observed on single element mechanical catheters. It was most prevalent on the
302 Atlantis SR Pro and Pro², and although not always present, it was very noticeable in
303 some instances (Figure 8). Despite efforts to minimise the curvature of the catheter
304 during testing, the impact of the artefact could not be controlled. Due to the

305 inconsistent nature of this artefact it was difficult to assess the impact on the
306 measurements made.

307

308 The results from this study demonstrate that IVUS imaging performance may be
309 characterised using the EPP. While the EPP can be used to track the development
310 of faults in clinical ultrasound scanners, the single use nature of the catheters and
311 higher uncertainty make it unlikely that the technique would be used to detect faults
312 in IVUS systems. The successful use of this phantom with IVUS probes indicates it
313 should also be suitable for testing of Transurethral Ultrasound and Intraductal
314 Ultrasound. Further work is required to broaden knowledge of the imaging
315 performance of these other catheter based ultrasound transducers.

316

317 **Conclusions**

318

319 With a lack of published information on the performance testing of catheter based
320 ultrasound transducers it has been shown in this study that modifications to an
321 existing imaging test technique and phantom can be used. The described
322 modification to the design of the Edinburgh Pipe Phantom made it suitable for testing
323 catheter based transducers. Experience with the testing of IVUS transducers on
324 both the modified and original EPP found the new phantom to be easier to use.
325 Values of resolution integral (R), depth of field (Lr) and characteristic resolution (Dr)
326 allowed the probes tested to be evaluated scientifically and their performance to be
327 seen in relation to other similar ultrasound technologies. This will be of particular use
328 to those interested in characterising IVUS equipment as either part of the
329 procurement process or to support research and clinical studies.

330

331 **Declarations**

332

333 This research was conducted with support from an Institute of Physics and
334 Engineering in Medicine (IPEM) Innovation and Research Award granted to SDP.

335

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337

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396

397 **Figures**

398 Figure 1. a) Schematic diagram of L against α for the ultrasound beam generated by
399 an IVUS catheter shown in image b) which illustrates how depth of field L_r and
400 characteristic resolution D_r are defined. The plot of L vs α for the IVUS catheter
401 extends from the low contrast penetration L_0 on the y-axis to $1/D_0$ on the x-axis,
402 where D_0 is the minimum beam width. The area under the curve $L_0 - 1/D_0$ is R . L_r
403 and $1/D_r$ are the coordinates of O' , the vertex of a rectangle with area R and in
404 which the diagonal OO' bisects the area under the curve $L_0 - 1/D_0$. Thus $R=L_r/D_r$. L_r
405 and D_r can also be considered as the characteristics of a perfectly collimated
406 ultrasound beam with low contrast penetration L_r and beam width D_r .

407

408 Figure 2. a) Rectangular plastic containers act as moulds for the Tissue Mimicking
409 Material (TMM) blocks. The different diameter pipes and sutures were threaded
410 through and held in place with sealant. b) The diameter of each pipe and the block
411 of TMM in which it was located. Each block contained 4 pipes, two of each size. An
412 image of the resulting anechoic pipe structure is shown in a).

413

414 Figure 3. 3D scans were obtained for all pipes except the 0.044 mm using a
415 RMV704 probe from a VisualSonics Vevo 770 preclinical ultrasound scanner. This
416 particular pipe shown is 0.33 mm in diameter. The RMV704 probe has a 30 MHz
417 centre-frequency making it suitable for imaging such small targets. The 3D scans
418 provided confidence that there were no pockets of gas and that the pipes were patent.

419

420 Figure 4. Set up of the IVUS phantom: two TMM blocks, each with anechoic pipe
421 structures, can be held within the Perspex container. The container also houses the

422 Perspex guide blocks that allow the catheter to be slotted through a channel cut into
423 them so that the catheter is suspended just above the surface of the TMM. Via
424 movement of the guide blocks the catheter can be positioned over the pipe to be
425 imaged and the length of pipe which is distinguishable from background speckle can
426 be measured. This set up places minimal strain on the catheter and allows the
427 operator to adjust imaging settings as required to optimise the image.

428

429 Figure 5. Measurement of (a) the superficial visible section of pipe, and b) the
430 deepest visible section of pipe. The depth range L over which the pipe is visible is
431 given by $d_2 - d_1 = 3.7 \text{ mm}$

432

433 Figure 6. Graph of depth of field (L_r) versus characteristic resolution (D_r) showing the
434 IVUS results for both single element mechanical and phased array transducers.
435 Error bars show the 95% confidence interval for the L_r and D_r measurements made
436 on each transducer.

437

438 Figure 7. A depth of field (L_r) versus characteristic resolution (D_r) plot; shown in
439 black squares are the IVUS single element mechanical transducer results obtained
440 from the modified EPP. Results for preclinical single element mechanical
441 transducers are shown as reported by Moran et al. (2011). Single element
442 mechanical endoscopic ultrasound probes reported by Inglis et al. (2014) are also
443 shown.

444

445 Figure 8. Example of the Non Uniform Rotation Distortion (NURD) artefact. a) 1.5
446 mm pipe imaged with the scan surface horizontal. b) From the same probe an

447 image of the 0.23 mm pipe. In this image the scan surface has become distorted
448 and appears to slope away from the transducer. This artefact can vary in severity
449 and was seen on the single element mechanical catheters, the example shown is the
450 Atlantis SR Pro. There is also a mirror image and shadowing artefact in the upper
451 half of the image due to reflection from the surface of the TMM fluid.
452

453 **Tables**

454

Manu.	Scanner	Probe	Transducer Type	Nominal Centre Freq. (MHz)	R	Lr (mm)	Dr (mm)
Boston Sci.	Clearview	Atlantis SR Pro	M	40	18.8	4.91	0.26
Boston Sci.	Clearview	Atlantis SR Pro ²	M	40	17.2	4.88	0.28
Volcano	S5	Eagle Eye platinum	PA	20	15.3	5.94	0.39
Volcano	S5	Revolution Rotational Imaging	M	45	14.7	4.44	0.30
Boston Sci.	iLab	Opticross	M	40	14.5	3.84	0.27
Volcano	S5	Visions PV .018	PA	20	14.2	5.95	0.44
Volcano	S5	Visions PV .035	PA	10	11.6	14.0	1.21

455 Table 1. The resolution integral (R), characteristic resolution (Dr) and depth of field

456 (Lr). Transducer type noted as M=Single element mechanical, PA=Phased Array.

Figure 1
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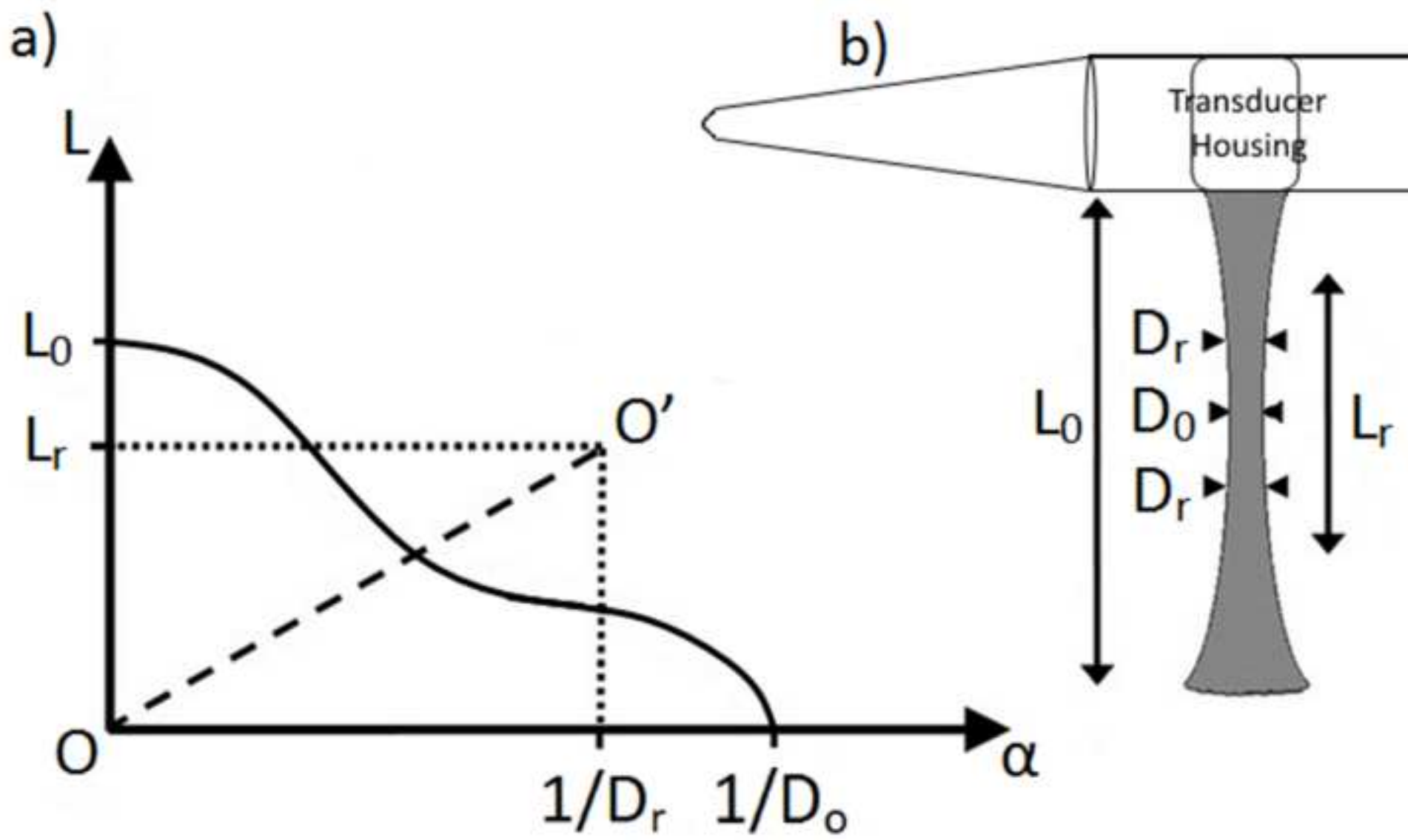
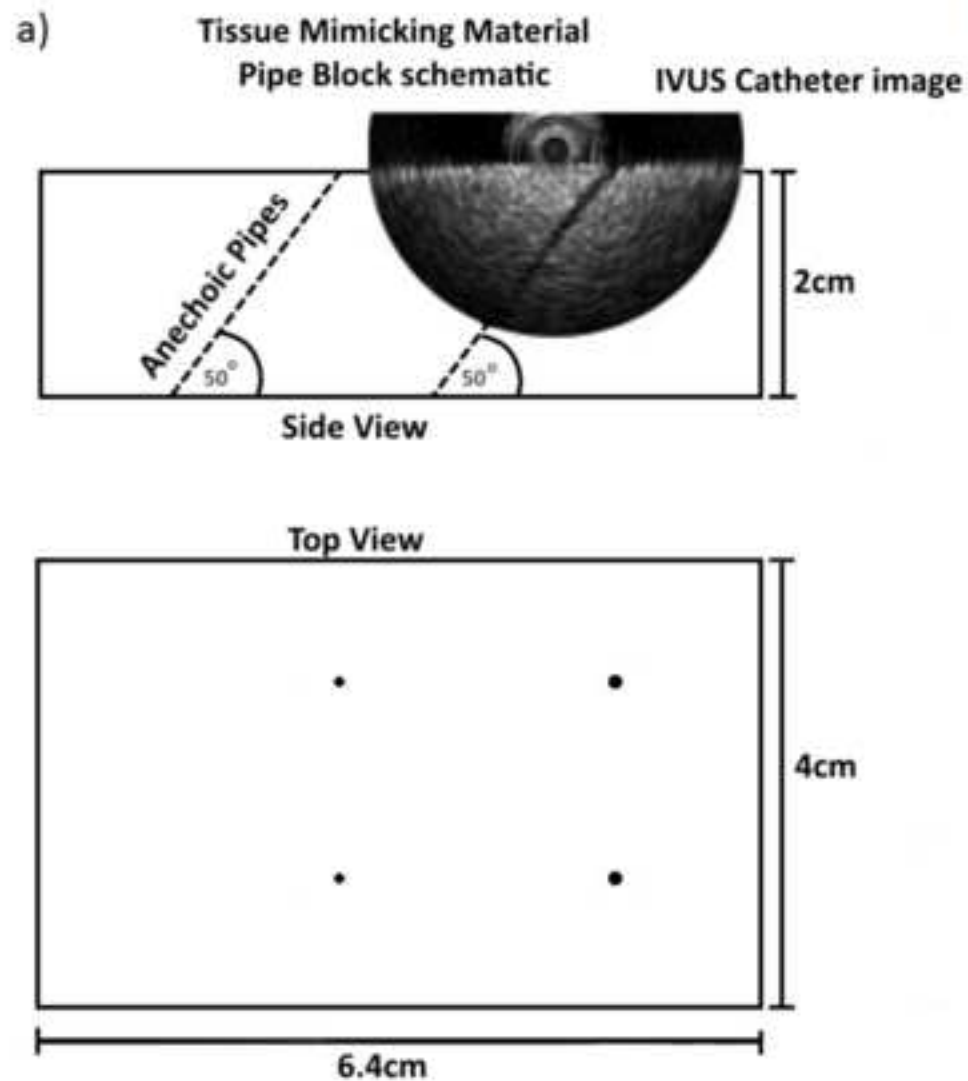


Figure 2
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b)

Modified Edinburgh Pipe Phantom (IVUS Phantom)			Original Edinburgh Pipe Phantom
TMM Block	Mould Material	Diameter (mm)	Diameter (mm)
1	Metal Rod	1.5	7.9
1	Metal Rod	0.55	6.0
2	Metal Rod	0.33	4.0
2	Metal Rod	0.23	3.0
3	Monofilament Suture	0.19	2.0
3	Monofilament Suture	0.14	1.5
4	Metal Wire	0.11	1.0
4	Monofilament Suture	0.094	0.72
5	Monofilament Suture	0.073	0.55
5	Metal Wire	0.044	0.42

Figure 3
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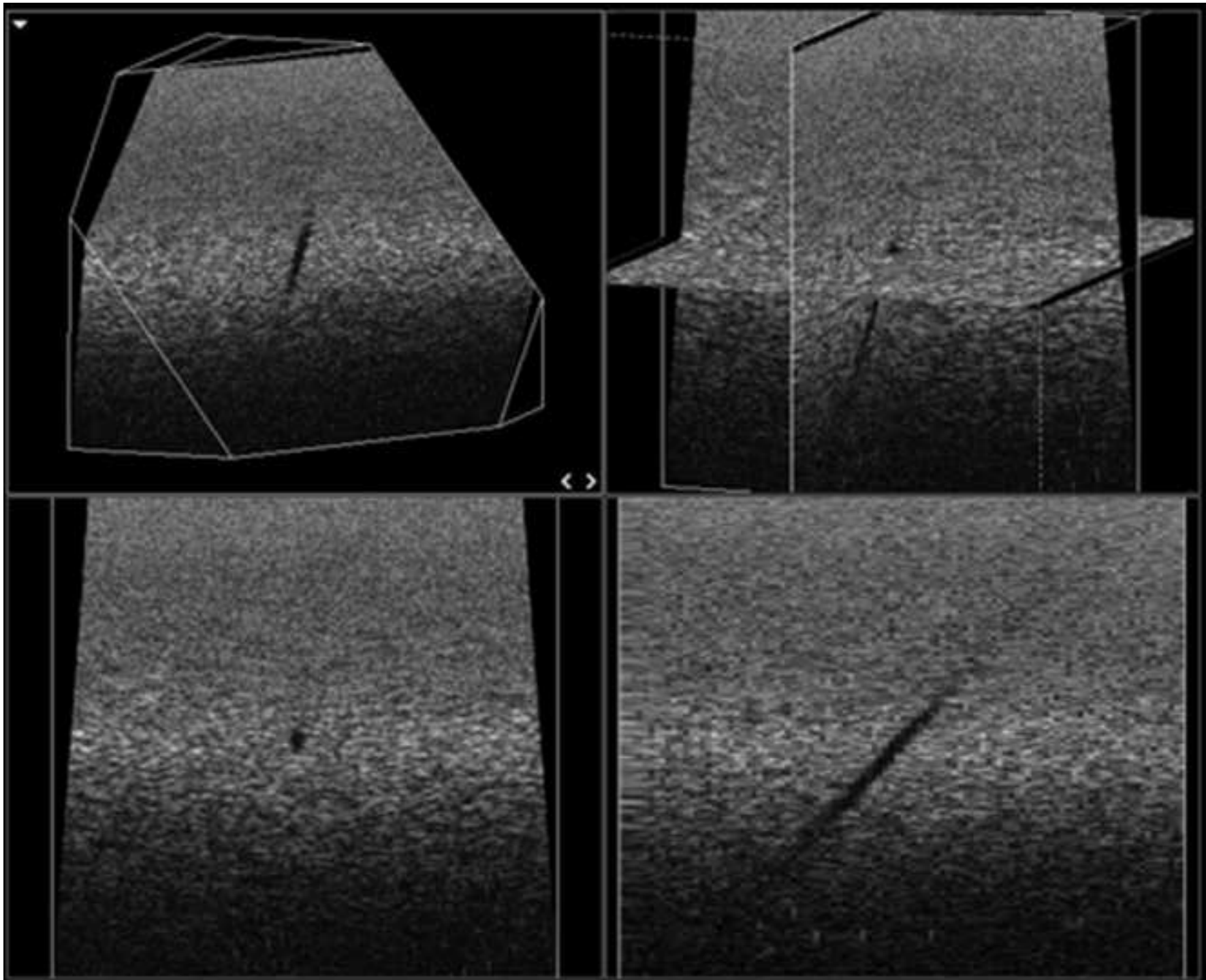


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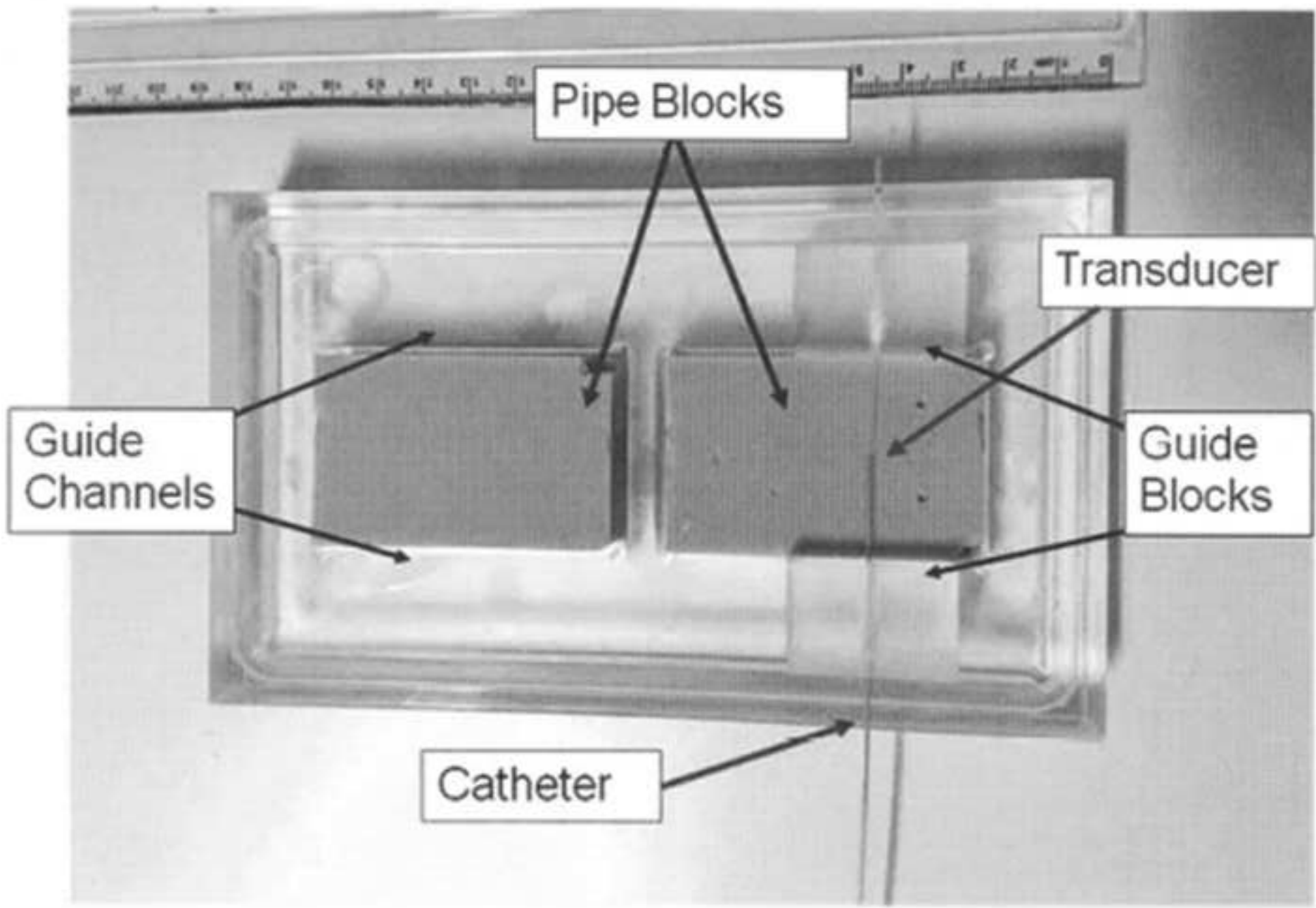


Figure 5
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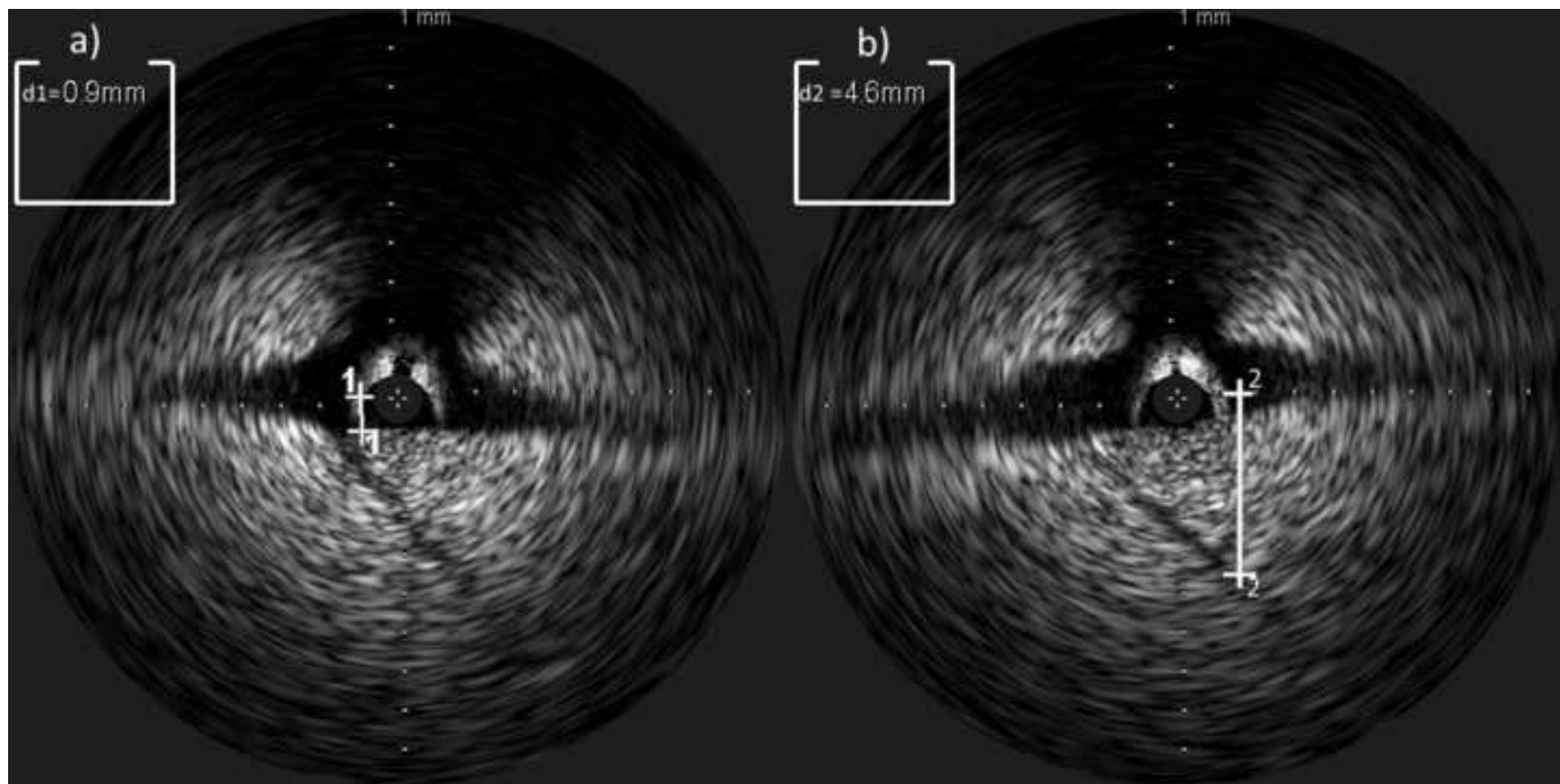


Figure 6
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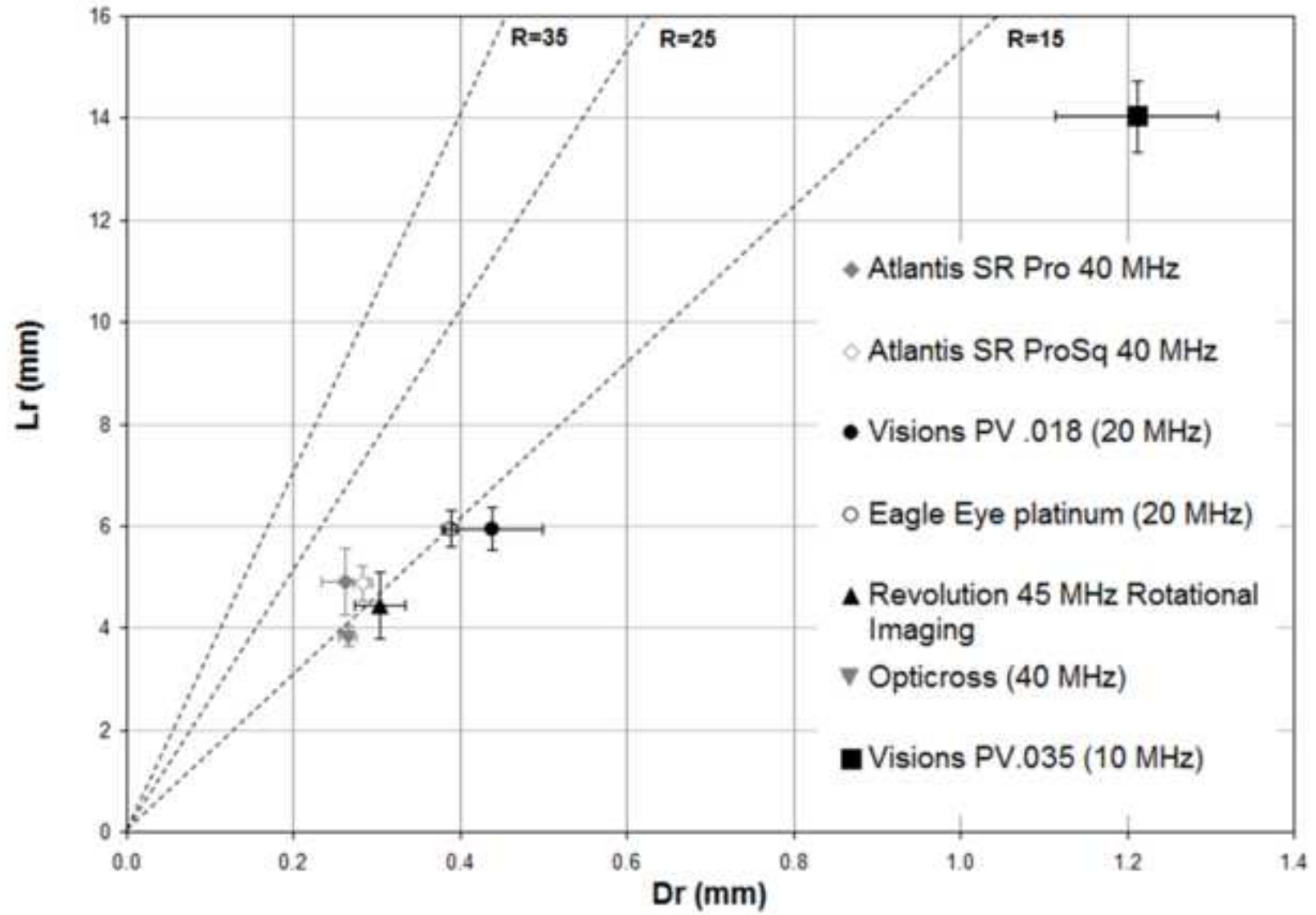


Figure 7
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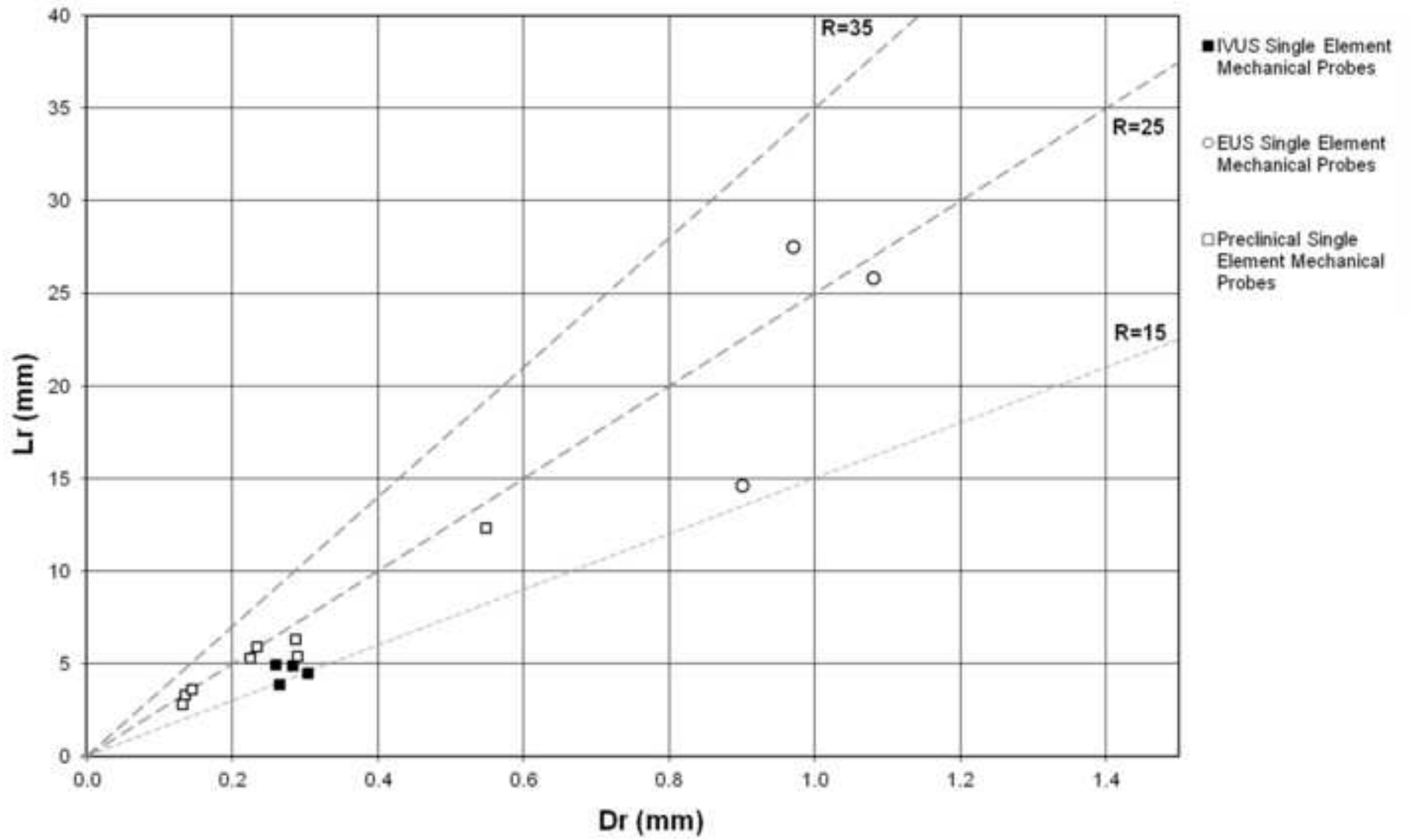


Figure 8
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