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1 **Ultrasonic needle tracking with dynamic electronic focusing**

2

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21 **Abstract:**

22 Accurate identification of the needle tip is a key challenge with ultrasound-guided percutaneous
23 interventions in regional anaesthesia, foetal surgery and cardiovascular medicine. In this study, we
24 developed an ultrasonic needle tracking system in which the measured needle tip location was used
25 to set the electronic focus of the external ultrasound imaging probe. In this system, needle tip tracking
26 was enabled with a fibre optic ultrasound sensor that was integrated into a needle stylet, and the A-
27 lines recorded by the sensor were processed to generate tracking images of the needle tip. The needle
28 tip position was estimated from the tracking images. The dependency of the tracking image on the
29 electronic focal depth of the external ultrasound imaging probe was studied in a water bath and with
30 needle insertions into a clinical training phantom. The variability in the estimated tracked position of
31 the needle tip, with the needle tip at fixed depths in the imaging plane across a depth range from 0.5
32 cm – 7.5 cm, was studied. When the electronic focus was fixed, the variability of tracked position was
33 found to increase with distance from that focus. The variability with the fixed focus was found to
34 depend on the the relative distance between the needle tip and focal depth. It is demonstrated that
35 with dynamic focusing, the maximum variability of tracked position was below 0.31 mm, as compared
36 with 3.97 mm for a fixed focus.

37

38 **Keywords:** Ultrasound guided needle intervention, ultrasonic needle tracking, needle tip localization,
39 fiber optic ultrasound sensor, Fabry-Perot interferometer, interventional ultrasound

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43 **Introduction**

44 Ultrasound (US) image guidance is routinely used in clinical practice to guide needle-based
45 percutaneous procedures such as for the delivery of regional anaesthesia, tumour biopsies, central
46 venous access in cardiology and foetal medicine (Chin et al. 2008, Holm and Skjoldbye 1986, Khati et
47 al. 2011, Leibowitz et al. 2020, Morris and Weston 1998). A key challenge associated with ultrasound-
48 guided needle insertions is the efficient and accurate identification of the needle tip. In-plane
49 approaches require good hand-eye coordination to keep the needle in the imaging plane as it is
50 advanced to the target. These challenges are particularly acute when the needle tip is at high depths
51 (> 5 cm), when the insertion angle is steep and when the needle diameter is small (e.g., 22G or higher)
52 so that out-of-plane deflection takes place during insertions (Brambati et al. 1987, Ghi et al. 2016).
53 Even for experienced clinicians, the needle can readily stray from the imaging plane and damage a
54 critical structure (Smith et al. 1998).

55 Various techniques have been proposed to improve the visibility of the needle tip in ultrasound-
56 guided needle interventions (Beigi et al. 2021, McLeod 2021, Van de Berg et al. 2019). These include
57 the use of echogenic needles, camera-based tracking, electromagnetic tracking, and ultrasonic needle
58 tracking with the use of an integrated sensor, actuator or transmitter in the needle (Hebard and
59 Graham 2011, Levy et al. 2007, Nikolov and Jorgen 2008, Najafi et al. 2015, Mung et al. 2011, Guo et
60 al. 2014, Cheng et al. 2018, Xia et al. 2017). Ultrasonic needle tracking with the use of a fibre optic
61 ultrasound sensor (FOUS) positioned at the needle tip has shown good promise recently and has been
62 demonstrated for both 2D and 3D tracking (Xia et al. 2015, 2017). Previous implementations by our
63 group used two alternating sequences of ultrasound transmissions: one with electronic focusing for
64 US imaging and a second with transmissions by individual elements for US tracking. With the latter
65 sequence, the A-lines recorded by the FOUS are processed to generate a tracking image, in which the
66 only object is the needle tip. There are two main limitations of this paradigm. First, the use of two
67 sequences results in a two-fold reduction in effective imaging rate. Second, single-element

68 transmissions can have high divergence in-plane relative to electronically-focused transmissions,
69 which leads to lower hydrophone signal magnitudes in the tracking images, which could necessitate
70 computationally expensive algorithms to improve the SNR.

71 An alternative paradigm for needle tracking with a FOUS, which could overcome the
72 aforementioned limitations, involves the use of only one sequence of electronically focussed
73 transmissions that is used for both US imaging and tracking. Within this paradigm, if there is a fixed
74 electronic focus, a primary challenge that arises is that the lateral width of the needle tip in the
75 tracking image is highly dependent on the axial distance of the needle tip relative to that focus. It was
76 previously been demonstrated that as these widths increase, there is a corresponding decrease in the
77 accuracy with which the needle tip locations can be measured from the tracking images (Mari et al.
78 2014).

79 The solution we propose here is to perform US tracking with a single sequence of electronically
80 focused transmissions, and dynamically adjust the electronic focus based on the measured axial
81 location of the needle tip. We present a fully functional system with this capability, which includes a
82 real-time needle tracking system based on a FOUS. The sensor is integrated within the needle using a
83 novel withdrawable stylet design that can be tailored to suit different needle types and therefore
84 different clinical procedures. We study the impact of the US electronic focus on the reconstructed
85 image of the needle tip and estimate the variability in the tracked needle tip positions with fixed US
86 focus and with focus tracking in a water bath and a clinical phantom.

87

88 **Materials and Methods**

89 *Ultrasonic needle tracking system overview*

90 The US needle tracking (UNT) system was designed at the outset to be compatible with clinical B-
91 mode imaging. A schematic overview is shown in Figure 1. The system uses a clinical US imaging

92 system (Ultrasonix MPD, BK Medical, UK) with a linear array imaging probe (L14-5/38, 128 elements,
93 linear array transducer, 14-5 MHz) for 2D imaging. A primary element of the tracking system is a
94 custom-designed console as shown in Figure 1(a). The ultrasonic needle tracking system comprises
95 the medical device which is a needle with an integrated FOUS (Figure 1 (b)). The FOUS receives the
96 electronically focussed transmissions from the imaging probe, which are processed on the tracking
97 console to estimate the needle tip positions. When the needle tip is away from the US focus, the FOUS
98 receives a diverging US scan-line and receives the signal from several scan-lines adjacent to the direct
99 line-of-sight position. This results in the needle tip image having an extended span in the lateral
100 direction. On the other hand, when the needle tip is at the same depth as the US focus, the FOUS
101 receives the signal from only a few scan-lines that are directly in the line-of-sight and is likely to have
102 a localized needle tip image (Figure 1 (c)).

103 The system also captures the live B-mode US images from the US imaging system. The tracked
104 positions of the needle tip are then overlaid on the live US images on the monitor on the tracking
105 console system, where the user can see the US images together with the active position of the needle
106 tip.

107

108 *Ultrasound Receiver: Fibre Optic Ultrasound Sensor*

109 Reception of the ultrasound transmissions from the imaging probe within the needle was
110 performed by a high finesse Fabry-Pérot (F-P) cavity interferometer-based FOUS. The FOUS was
111 custom-built using methods previously described (Zhang and Beard 2011, Guggenheim et al. 2017).
112 Briefly, the ultrasound sensing element was fabricated on the tip of the fibre, with thin film reflective
113 coatings and a polymer spacer to form a plano-concave F-P cavity. The optical thickness of the F-P
114 cavity is modulated by impinging ultrasound waves, and detected as a change of the intensity of
115 reflected interrogation light. The FOUS, is particularly suited for needle tracking as it has been
116 demonstrated to have an omnidirectional frequency response (Zhang and Beard 2015) so that it is

117 independent of the needle insertion angle. The sensor used for the studies had a measured noise
118 equivalent pressure of ~ 350 Pa in a detection bandwidth of 20 MHz and a good frequency-dependent
119 directivity response for incidence angles ranging from -165° to 165° (Mathews et al. 2019).

120 *Needle with Integrated Fibre Optic Ultrasound Sensor*

121 In this study, we chose a commercial, 22G spinal needle (OD: 0.71 mm) (BD Medical, USA) with a
122 length of 90 mm that is widely used in percutaneous interventions across different clinical application
123 spaces. The FOUS was integrated inside a custom-built stylet to fit inside the cannula of the
124 commercial needle. The stylet was fabricated using a 27G (OD: 0.41 mm) hypodermic tube and was
125 designed to match the dimension of the stylet which is provided with the commercial needle. A
126 microscope image of the needle tip with the stylet and the integrated FOUS is shown in Figure 1 (b).

127

128 *Needle tracking console: data acquisition*

129 For real-time ultrasound needle tracking, we developed a tracking console that connects to the
130 needle with a FOUS and receives the US transmission from the imaging probe of the US system. The
131 tracking console registers the trigger signals from the US imaging system and uses these to process
132 the FOUS signals to form an image of the needle tip, which is used to estimate the needle tip
133 coordinates in the imaging plane of the US probe. A system-level block diagram of the real-time needle
134 tracking console is shown in Figure 2. The interrogation of the FOUS was done by an external cavity
135 wavelength-tunable laser (Santec TSL-550, Santec Europe Ltd, UK) with a tuning range from 1500–
136 1630 nm. The laser was coupled into the FOUS using a fibre optic circulator and the reflected light
137 from the F-P cavity sensing element of the sensor was directed onto a photo-receiver system.

138 The photo-receiver system comprises an InGaAs photodiode-transimpedance amplifier unit that
139 generates low (V_{lf}) and high frequency (V_{hf}) output voltage signals. These were acquired in the needle
140 tracking workstation, where the V_{lf} was acquired using a multifunction I/O device (PCIe-6323, National

141 Instruments, UK), which was digitized at 16 bits with a sampling rate of 250 kS/s. The received
142 ultrasound signal, V_{hf} , was acquired using a high-speed digitizer (PCI-5114, 125 MHz Bandwidth,
143 National Instruments, UK), and was digitized at 8 bits at a sampling rate of 250 MS/s. The FOUS
144 interrogation software uses the low-frequency signal to measure the transfer function of the F-P cavity
145 and determine the interrogation wavelength of the tunable laser (via GPIB control) that provides the
146 maximum sensitivity (Zhang et al. 2008).

147 The high-speed digitizer uses the trigger outputs from the US imaging system (grey blocks on the
148 top). The acquisition of the digitizer was triggered by the output frame trigger which is the start of
149 each B-mode US frame. The digitizer then synchronously acquired both the line trigger (start of
150 individual A-lines) from the US system and the high-frequency FOUS signal from the photo-receiver.
151 These FOUS signals were processed through a parsing algorithm that applies a band-pass filter and
152 computes the signal envelope using a Hilbert Transform. The parsing was based on the reception of
153 the line triggers and the estimated relative delays in receiving the US signals from the elements of the
154 US imaging probe by the FOUS. The algorithm converts the 1-D FOUS signal (a series of A-lines in one
155 imaging frame) to a 2-D reconstructed image of the needle tip. This was done by a custom-built
156 tracking application software developed in LabVIEW. The position of the needle tip was estimated in
157 the coordinate system of the US imaging probe for overlay and passed on to the graphical user
158 interface (GUI) of the application software. The rate of the overlay on the tracking system was
159 approximately 10 Hz.

160

161 *Needle tracking console: image capture and dynamic focusing*

162 The console acquires the live stream of US images from the US imaging system through an Ethernet
163 port on the tracking workstation. The acquisition of the US images was enabled using a custom
164 LabVIEW compatible extensible C++ application programming interface (API) (Shakir and Mathews
165 2019), that is interfaced with the commercial software development kit (SDK) of the Ultrasonix MDP

166 imaging system (Figure 2; grey blocks on top). The API allows for real-time capture of the US frames
167 at an acquisition rate of over 60 frames per second. The US frames acquired as 1-D low-level metadata
168 was passed on to the tracking software, which converts it into 2-D B-mode US images based on the
169 image size set on the US imaging system. The US images were then passed on to the GUI as shown in
170 Figure 2.

171 In addition to real-time frame capture, the API can control the imaging parameters on the US
172 imaging system provided by the SDK. This allows the API to set the focal depth of the probe through
173 the tracking application software. The real-time focus tracking was achieved by the software by first
174 estimating the depth of the needle tip using the signals acquired by the high-speed digitizer and then
175 using the estimated depth of the needle to dynamically reset the focal depth of the US imaging probe
176 to the depth of the needle tip.

177

178 *Impact of US focus and needle tip variability*

179 To study the impact of the position of the US electronic focus on the needle tip image, the needle
180 was inserted into a clinical training phantom (Adam, Rouilly Ltd, UK) and needle tip images were
181 acquired at various focal depths. The depth of the needle tip was fixed at ~ 4.02 cm and the US focal
182 depth was varied from 0.5 cm – 7.5 cm. For each focal depth, the needle tip images were recorded by
183 the acquisition system and post-processed in MATLAB. This post-processing comprised maximum
184 intensity projections (MIPs) along the lateral (scan axis) and axial (depth axis) directions. Gaussian
185 fitting was applied to the MIPs in each direction; the full width at half maximum (FWHM) was
186 calculated from those fits and used as measures of the lateral and axial spread of the needle tip image.
187 The SNR was estimated from needle tip images by taking the ratio of the maximum amplitude of the
188 FOCUS signal and the estimated standard deviation of the noise floor from an empirically chosen region
189 of the 2-D needle tip image.

190 The needle tip positions were estimated based on both the coordinates of the maxima of the
191 reconstructed needle tip image and also on the centre of mass (CoM) of a region of interest (ROI)
192 around the position of the maxima. To estimate the CoM, a region of interest with 100×120 pixels
193 was selected from the needle tip image based on the position of the maximum. Prior to the calculation
194 of CoM, background subtraction was applied and then thresholding was applied to the ROI such that
195 coordinates with less than $\sim 70\%$ of the maximum intensity are zeroed, to obtain an accurate
196 representation of the shape of the needle tip image. The CoM was then estimated based on the
197 average of all positions weighted by the intensity at each position in the ROI.

198 The impact of a fixed focus on the variability of the needle tip position along the axial and lateral
199 directions in the imaging plane was studied. To measure the variability, the needle with a FOUS was
200 positioned inside a water tank directly under the US probe in the imaging plane. The needle tip images
201 were recorded sequentially for over 100 US frames, with the needle tip fixed inside a water bath. The
202 variability in the needle tip coordinate (for both axial and lateral) is defined as the absolute value of
203 the difference between the mean of the coordinate over the 100 US frames and the estimated
204 coordinate for each frame. The maximum value of the variability along both the lateral and axial
205 direction are taken as the maximum variability. Here we make the distinction between maximum
206 variability and absolute accuracy of tracking. The former quantity is taken to be a relative measure
207 that describes how the needle tip position estimates from the UNT change across frames; the latter,
208 which is not measured here, relates to a ground truth position of the needle tip.

209 The variability estimation was performed on both the needle tip positions based on maxima and
210 the CoM and the maximum variability in these positions were estimated with both fixed US focus and
211 with focus tracking. For the fixed focus case, the US focal depth was fixed at 4.18 cm, and the needle
212 was translated inside the water bath to have it positioned at various depths from ~ 1.5 cm to 6.5 cm.
213 For the focus tracking case, the needle was translated and positioned in the same depth range.

214

215 **Results**

216 *Impact of US focus on the image of the needle tip*

217 The 2-D needle tip images acquired for a fixed position of the needle tip (~ 4.02 cm) in a focal depth
218 range from 1.0 cm – 7.0 cm are shown in Figure 3(a). For shallower (1 cm – 3 cm) and deeper (5 cm –
219 7 cm) positions of focal depths relative to the depth of the needle tip, the image of the needle tip had
220 an extended lateral span. When the focal depth (4 cm) was coincident with the needle tip, the 2-D
221 needle tip image was a localized spot in the imaging plane.

222 Figure 3(b) shows the FWHM in the lateral and axial directions for the needle tip images recorded
223 at depths ranging from 0.5 cm – 7.5 cm, in increments of 0.5 cm. The FWHM in lateral direction varies
224 significantly with the axial position of the US focal depth. The values were highest for focal depths 0.5
225 cm (FWHM = 11.06 mm) and 7.5 cm (FWHM = 10.02 cm) and was estimated to be the least at a focal
226 depth of 4 cm (FWHM = 1.02 mm), where the needle tip (fixed at 4.02 cm) is coincident with the
227 position of the US focal depth. The FWHM in the axial direction is marginally impacted by the position
228 of the US focal depth. The estimated values were in the range from 0.487 mm – 1.11 mm. Figure 3(b)
229 also shows how the signal-to-noise ratio (SNR) of the 2-D needle tip image varies with the focal depth.
230 A maximum SNR of ~ 165 was measured when the focal depth was at ~ 4.0 cm. The SNR reduces when
231 the focal depth was set above or below the depth of the needle tip. A minimum SNR of ~ 20 (a relative
232 drop of 88% from maximum) was measured at the focal depth of 0.5 cm.

233

234 *Variability of needle tip position with US focal depth*

235 For the case where the electronic focus was fixed (4.18 cm; Figure 4a), the needle tip images
236 obtained far from the focus (1.73 cm and 6.61 cm) had extended lateral spans. The lateral span was
237 considerably smaller when the needle tip was positioned at the US focus. For the case with focus
238 tracking, a small lateral span was consistently obtained across this depth range (1.69 cm, 3.95 cm and

239 6.49 cm; Figure 4b). Figure 4(c) and (d) summarise the maximum variability in estimated needle tip
240 positions (axial and lateral) based on both the maximum and CoM, with fixed US focus and with focus
241 tracking. Along the lateral direction (Figure 4(a)), the variability of needle tip positions was the highest
242 in the case with fixed US focus and when the maximum was used as the position of the needle tip. The
243 highest value for maximum variability was estimated as 3.97 mm at a depth of ~ 1.73 mm and the
244 lowest value of ~ 0.33 mm was estimated at a depth of ~ 4.17 mm. For needle tip positions based on
245 the CoM, the variability was lower in comparison. The estimated values ranged between a maximum
246 of 1.33 mm to a minimum of 0.22 mm. The variability was maximal when the needle was further away
247 from the position of the focal depth on either side of the fixed US focus. The variability was the least
248 with the focus tracking feature on as can be seen in Figure 4(a). The estimated values were found to
249 be in the range from 0.12 mm – 0.31 mm for the depth range from 1.73 cm – 6.61 cm.

250 Figure 4(b) shows the variability with the needle tip position in the axial direction for all three cases
251 (maximum, CoM for fixed focus and with focus tracking). For the coordinates based on the maximum
252 of the needle tip image, the maximum variability was estimated to be in the range from 0.21 mm –
253 0.38 mm. For the CoM based needle tip positions the maximum variability in the axial direction was
254 estimated to be in the range from 0.046 mm – 0.106 mm. With focus tracking, the maximum variability
255 was approximately 0.05 mm across a depth range from 1.73 cm – 6.61 cm. In the axial direction, the
256 variability did not have any dependency on the US focal depth.

257

258 **Discussion**

259 One prominent challenge for clinicians with procedures that involve deep target insertions is the
260 fixed position of the US electronic focus of the imaging probe during the procedure. It is quite well
261 known that the dynamic US focus feature provided by the modern US imaging system improves the
262 resolution of the US image (Manes et al. 1988, Powers and Kremkau 2011). For deep target needle
263 insertions, clinicians set the US focal depth based on the depth of the target and/or the envisaged

264 needle insertion path and is normally not changed during the procedure, particularly because of the
265 requirement to manually reset the focus on the US system. During the needle insertion, this can lead
266 to a poor resolution in the US image at depths above and below the set depth of the US focus.
267 Although advanced US imaging systems can provide multiple US transmit foci, this feature results in a
268 reduction of the frame rate (Powers and Kremkau 2011). The focus tracking feature, which has been
269 demonstrated with the system here, can provide dynamic focusing without reduction in imaging frame
270 rate together with real-time needle tip tracking.

271 The major advantage provided by the UNT system presented in this paper, is the reduction in
272 frame-to-frame variability of the tracked positions of the needle tip. The FOUS in the needle receives
273 the signals from the elements of the US imaging probe with relatively varied delays depending on
274 where the electronic focus was set. For shallower and deeper focal depths relative to the depth of the
275 needle, the 2-D reconstructed images of the needle tip were spread laterally across the US imaging
276 plane. When the needle tip was at the US focus, the FOUS receives the signals from the elements of
277 the US imaging probe with the least relative delays and the 2-D needle tip image was more like a
278 localized spot in the US imaging plane as observed in figure 3(a) with the needle in the phantom. In
279 the axial direction, the shape of the needle tip image primarily depends on the frequency response of
280 the FOUS and is mostly found to be similar with changing depth and the position of the ultrasound
281 focus (Figure 3(b)).

282 With the focusing of ultrasound to the depth of the needle tip, the FOUS receives a stronger signal
283 due to constructive interference of US signals from the maximum number of elements of the imaging
284 probe that are in the line-of-sight. In the case of the needle in the clinical training phantom, the SNR
285 of the FOUS signal was highest when the US focus is set axially closer to the depth of the needle tip.
286 The studies into the impact of US focus demonstrate that the FOUS signal will have good SNR and a
287 spatially localized 2-D needle tip image when the needle tip coincides with the US focal depth.

288 The use of CoM for the estimation of needle tip coordinates reduces the randomization of the
289 position of the maximum caused due to noise in the acquisition system. As a result, the variability in
290 estimated needle tip positions based on CoM are consistently lower in comparison to the needle tip
291 position based on maxima. However, with a fixed US electronic focus, the variability in the lateral
292 direction (figure 4(a)) is highly dependent on the position of the focal depth. This is due to the
293 prominent broadening feature of the reconstructed needle tip image in the lateral direction (figure
294 3(a)), caused by the variations in the delays in receiving the signals from the elements of the US
295 imaging probe by the FOUS. With a laterally-broadened image of the needle tip, the variability of the
296 lateral position of the maximum from frame-to-frame will increase, although the CoM based
297 estimation helps in reducing this variability, it is still dependant on the shape of the needle tip image.
298 With real-time focus tracking, the needle tip images are locally confined in the US imaging plane due
299 to the focus of the ultrasound probe being reset to the depth of the needle tip. The variability
300 estimates with the needle in the water bath show the maximum variability were the lowest (for both
301 lateral and lateral and axial directions) with the real-time focus tracking. Lower frame-to-frame
302 variability is useful to improve tracking accuracy. However, to estimate the tracking accuracy, a
303 measure of the ground truth for the needle tip position is required for comparison. There are several
304 methods for estimating this ground truth, including the use of an additional imaging modality (for
305 instance, multi-axis fluoroscopy) or with linear translation stages, as described in Xia et al. 2017.

306 Several improvements to the system presented here can be envisaged. The effective frame rate
307 (ca. 10 Hz) was limited by the digitizer used that had a single trigger input. Data acquisition level
308 parsing of the FOUS signals can be implemented by using a digitizer card with two or more trigger
309 inputs, which will eliminate the need to apply a parsing algorithm post signal acquisition. In future
310 implementations, the processing and visualizing of US and tracking image data could be more
311 efficiently handled with the use of high-speed image acquisition and tracking image processing using
312 a graphical processing unit (GPU). In this way, the effective frame rate could be improved substantially
313 to match the frame rate of the US imaging system (25 Hz - 30 Hz).

314 **Conclusions**

315 In conclusion, we have demonstrated for the first time a fibre optic ultrasound sensor based real-
316 time US needle tracking system with dynamic electronic focusing. With US focus tracking, the
317 electronic focus of the US imaging probe can be set based on the estimated needle depth, which allows
318 for dynamic focusing based on the point of interest of the medical device. The SNR is shown to improve
319 with focus tracking and the variability in estimating the needle tip position along the lateral direction
320 is shown to have reduced with the maximum variability estimated to be below 0.31 mm. Both these
321 factors will reduce the uncertainty in estimating the absolute needle tip positions and improve the
322 absolute accuracy of needle tracking with FOUS integrated needles in ultrasonic needle tracking
323 systems.

324 Focus tracking ensures dynamic focusing can be achieved without a reduction in the imaging frame
325 rate, which could avoid the need to set a fixed focus or multiple US foci prior to the start of the clinical
326 procedure, and also avoid the need to reset or change the focal depth during the procedure. Dynamic
327 focus tracking also ensures that the transmit focus of the US probe is near the physical depth of the
328 needle tip, which would provide the best possible US image quality for the anatomy near the needle
329 tip and in the region of interest. The real-time focus tracking feature of the system is attractive in the
330 space of foetal medicines with longer needles (≥ 8 cm) for deep target insertions, for procedures such
331 as amniocentesis, chorionic villus sampling and foetal heart surgery.

332

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344

345 **Conflict of interest statement**

346 The authors declare that there are no conflicting interests

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433 **Figure Captions List**

434 **Figure 1:**

435 Schematic with the overview of the ultrasonic needle tracking system.

436 (a) Showing the needle with fibre optic ultrasound sensor (FOUS), tracking console, ultrasound
437 imaging system and its imaging probe, (b) Microscope image of the 22G needle tip, with the 27G stylet
438 and the sensing element (green) at the end of the FOUS, and (c) Schematic showing the reception of
439 electronically focused transmission from the US imaging probe, by the needle with the FOUS at the
440 needle tip. Two scenarios are shown, first with the needle tip away from US focus (top) and second
441 with the needle tip at US focus (bottom). The corresponding needle tip images formed by the signals
442 received by the FOUS are shown on the right in both cases.

443

444 **Figure 2:**

445 System level block diagram of the ultrasonic needle tracking system.

446 The hardware components are shown on the left and the software components are shown on the
447 right. The electronic connections are shown using black solid lines and optical connections are shown
448 with red solid lines. The associated software communications and data flow are shown using dashed
449 lines. Yellow blocks correspond to the needle, the FOUS console and the associated FOUS
450 interrogation software; Blue blocks correspond to the needle tracking workstation, software and GUI;
451 Grey blocks correspond to the US imaging probe, US imaging system and the application programming
452 interface. The green dotted lines show focus tracking.

453

454

455 **Figure 3:**

456 Impact of US focus on needle tip image.

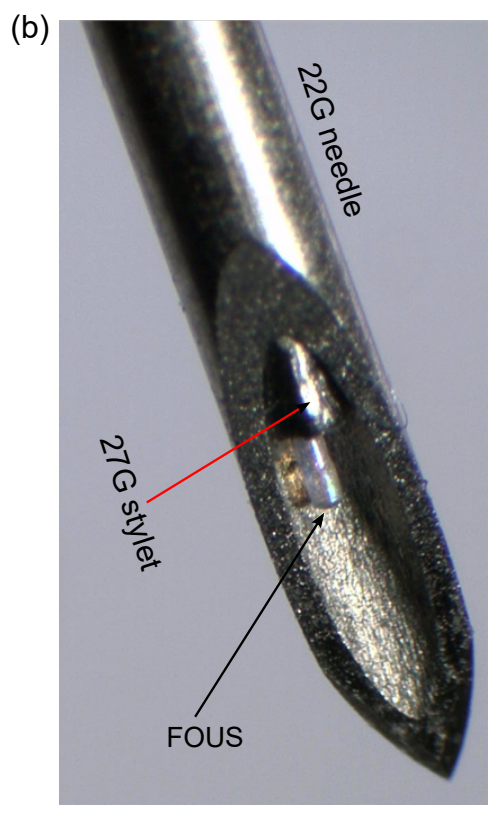
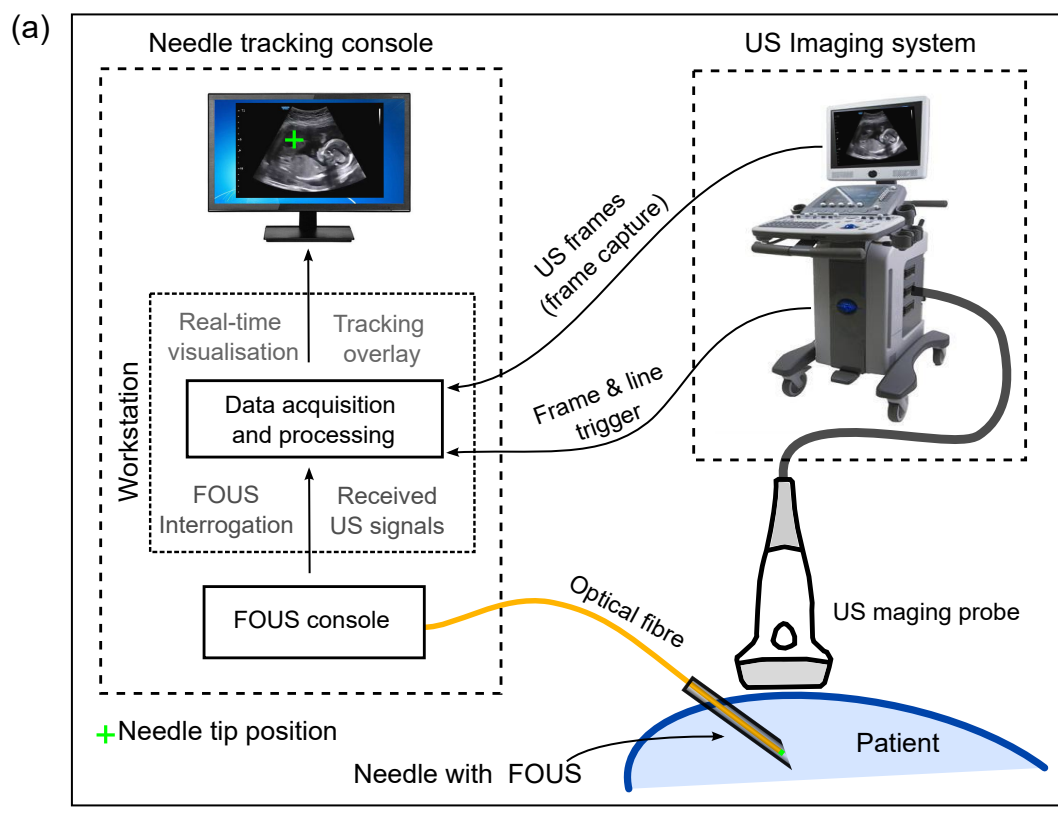
457 (a) 2-D needle tip images acquired with fixed US focus at depths ranging from 1.0 cm – 7.0 cm. (b) A
458 plot showing the FWHM of the needle tip images along the axial and lateral direction, together with
459 SNR for the FOUS signal. The US focal depth was varied from 0.5 cm – 7.5 cm. The needle tip was fixed
460 at a depth of 4.02 cm in the imaging plane.

461

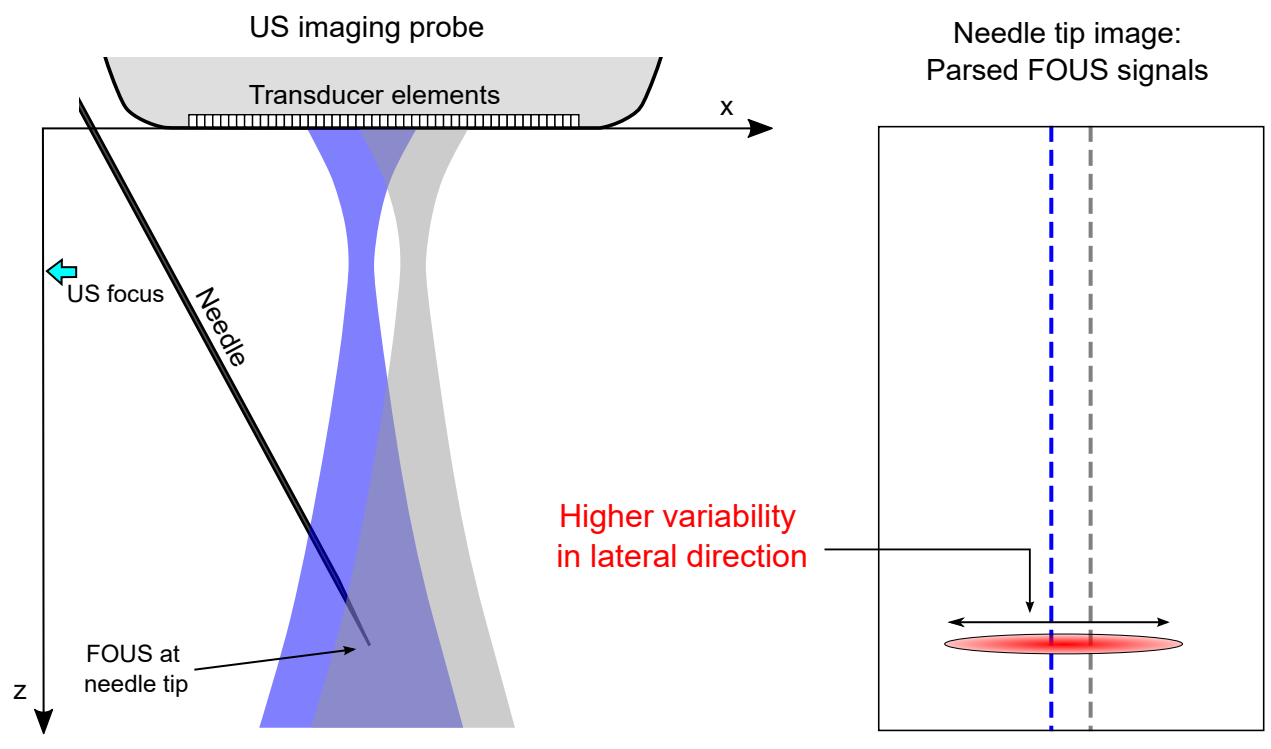
462 **Figure 4:**

463 Variability of the needle tip position with US focal depth.

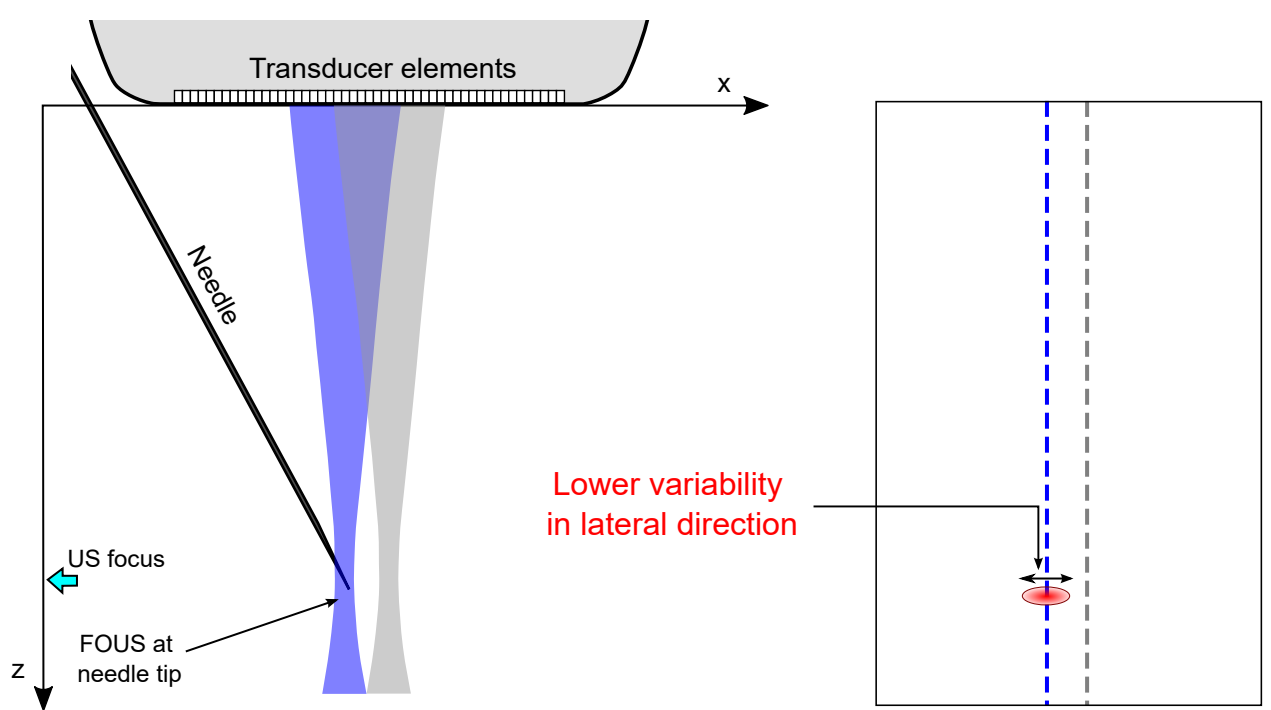
464 (a) Needle tip images at depths 1.73 cm (a_1), 4.17 cm (a_2) and 6.61 cm (a_3), with fixed US focus at 4.18
465 cm, (b) Needle tip images with focus tracking at depths 1.69 cm (b_1), 3.95 cm (b_2) and 6.49 cm (b_3).
466 The needle tip images are normalized to their individual maximum intensities. (c,d) Maximum
467 variability for fixed focal depth (at ~ 4.18 cm) and focus tracking, estimated using the position of
468 maximum and the centre of mass:(c) along the lateral direction (c), and along the axial direction (d).
469 The datapoints corresponding to needle tip images (a_1), (a_2), (a_3), (b_1), (b_2), and (b_3) are indicated.



(c) **Needle tip away from US focus**

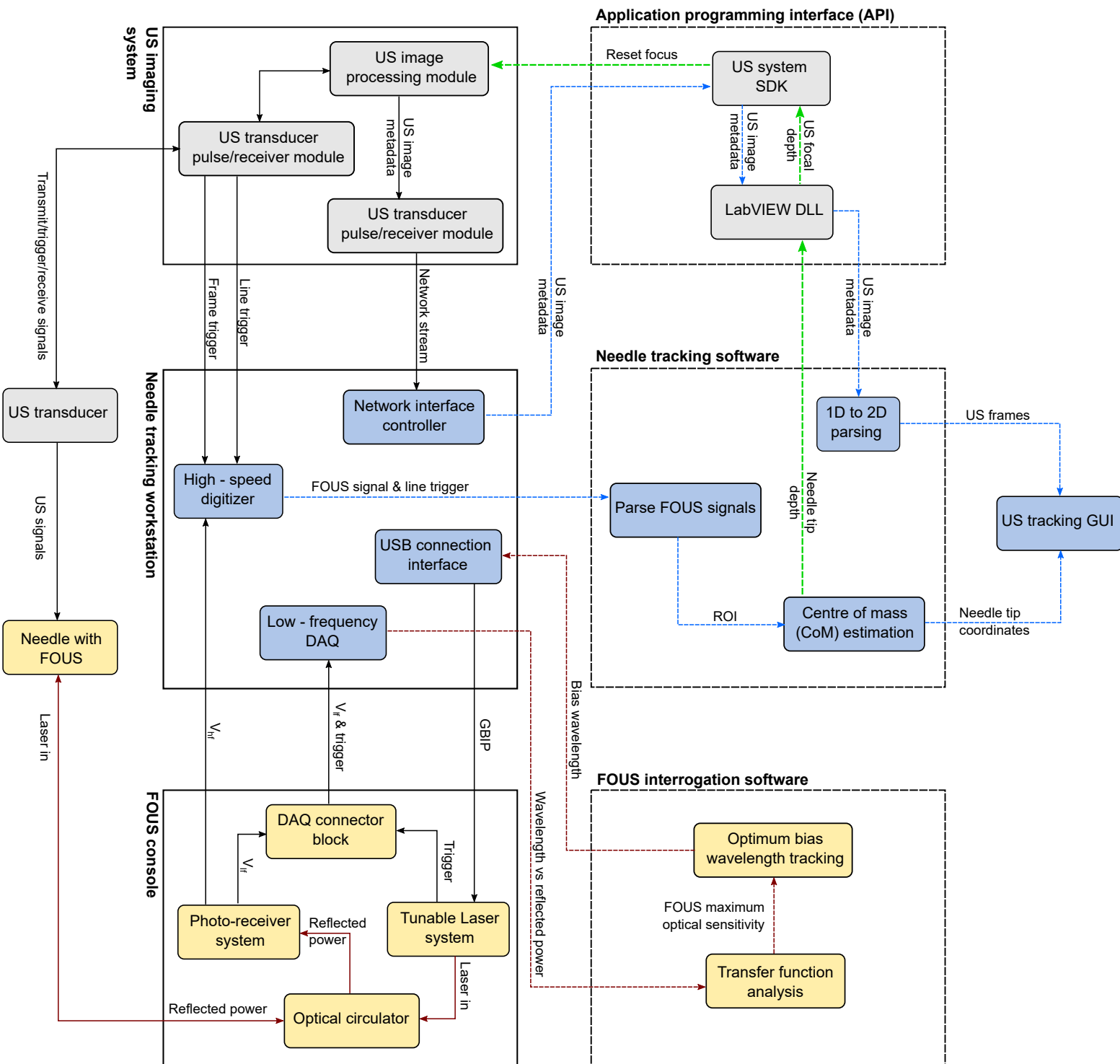


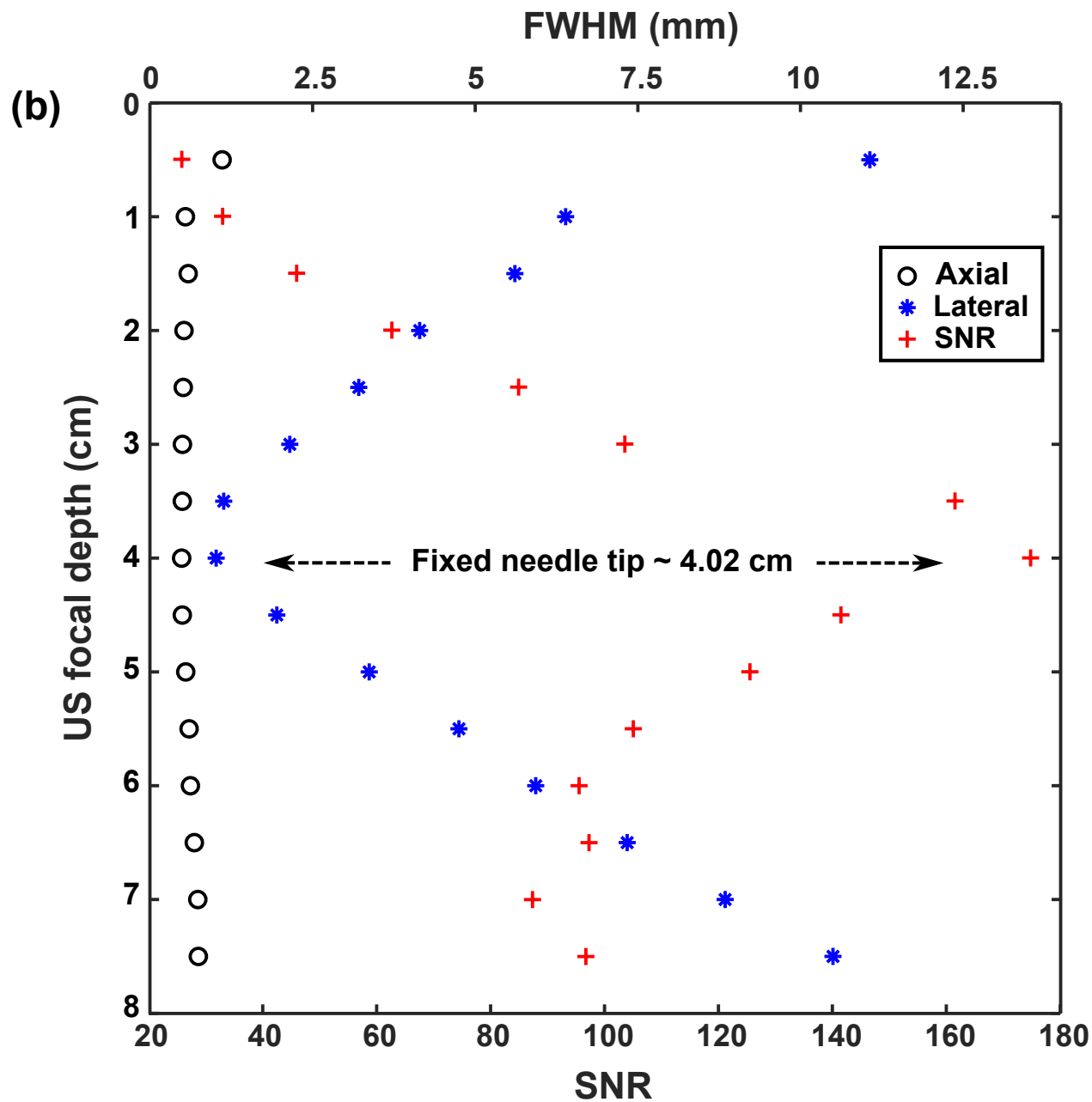
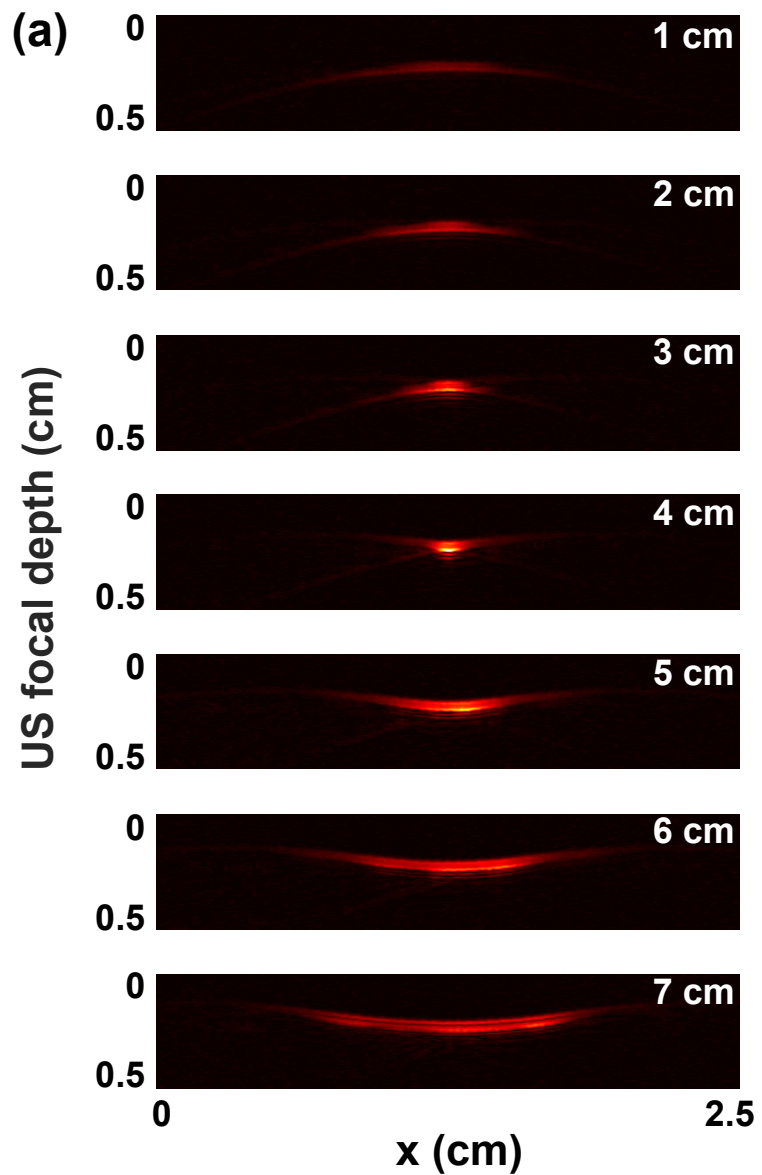
Needle tip at US focus



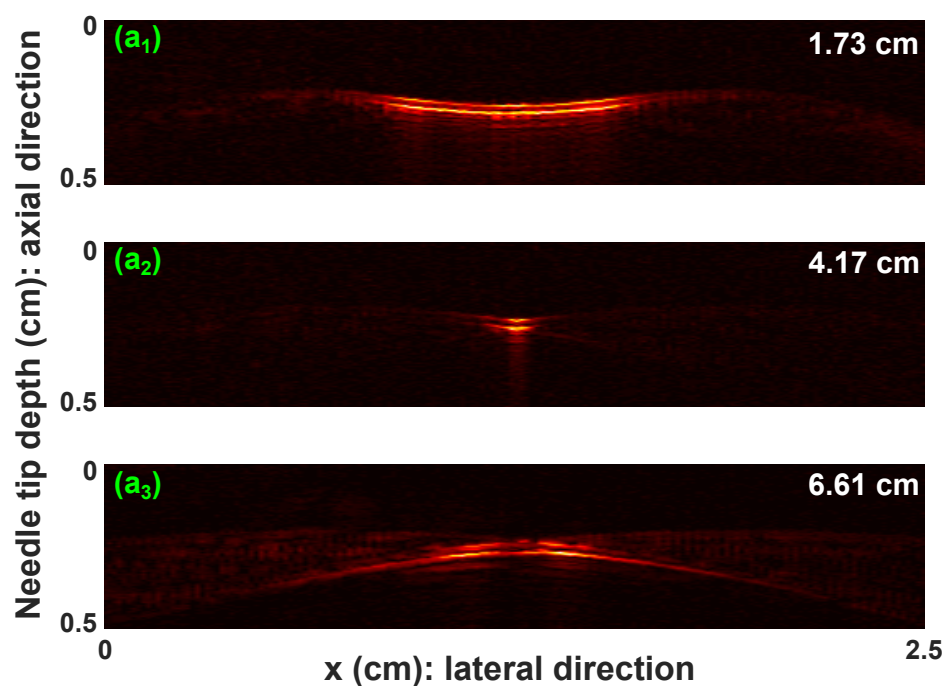
Hardware components

Software components

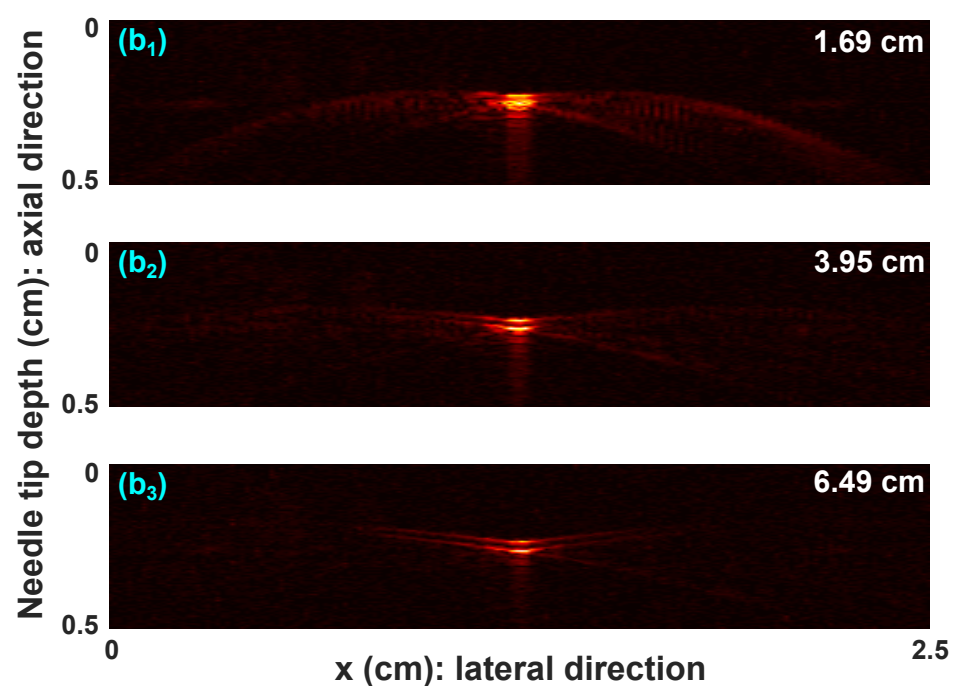




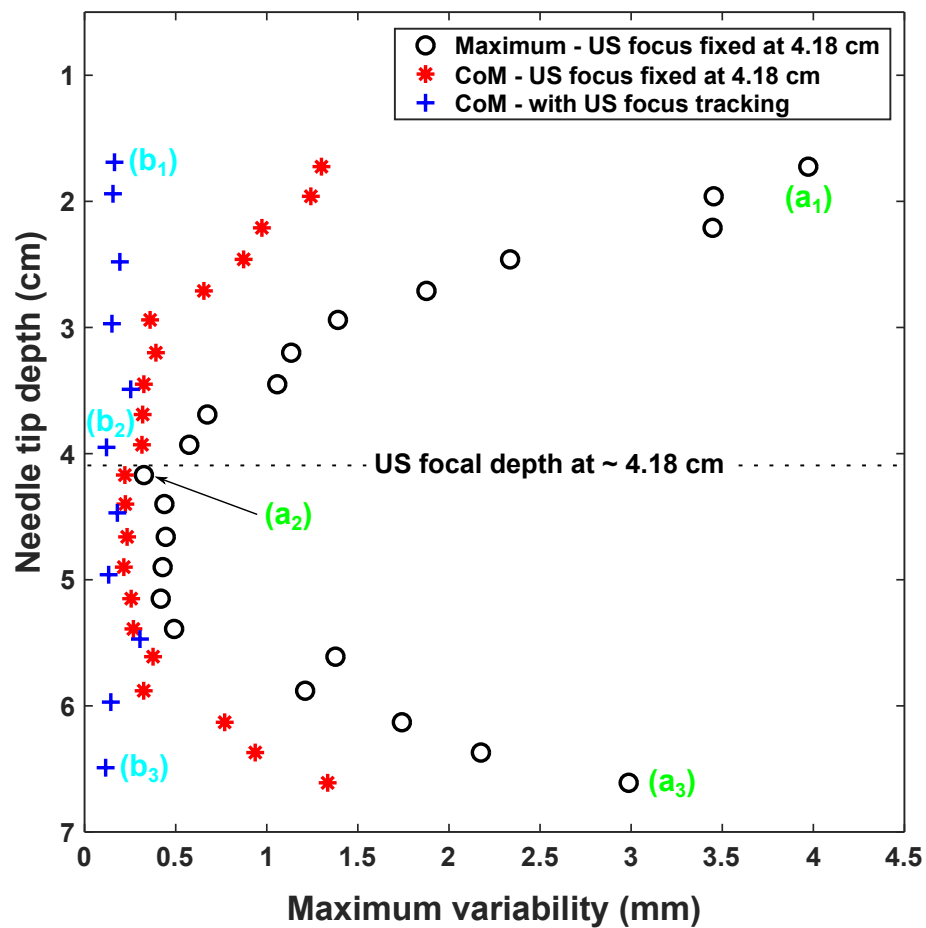
(a) Needle tip images with fixed focus at 4.18 cm



(b) Needle tip images with focus tracking



(c) Max variability along lateral direction



(d) Max variability along axial direction

