DESIGN AND DEVELOPMENT OF A PROTOTYPE ENDOCAVITARY PROBE FOR HIGH INTENSITY FOCUSED ULTRASOUND DELIVERY WITH INTEGRATED MAGNETIC RESONANCE IMAGING

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Running Title: Prototype endocavitary probe for MRI guided HIFU
ABSTRACT

Purpose: To integrate a high intensity focused ultrasound (HIFU) transducer with a magnetic resonance (MR) receiver coil for endocavitary MR guided thermal ablation of localized pelvic lesions.

Methods: A hollow semi-cylindrical probe (diameter 3.2cm) with a rectangular upper surface (7.2cm x 3.2cm) was designed to house a HIFU transducer and enable acoustic contact with an intraluminal wall. The probe was rounded distally to ease endocavitary insertion, and tapered proximally to a 1.5cm diameter cylindrical handle through which passed irrigation tubes (for transducer cooling) and electrical connections. MR compatibility of piezoceramic and piezocomposite transducers was assessed using gradient-echo sequences. Identical 6.5cm x 2.5cm rectangular receiver coils on the upper surface of the probe had their rf tuning adjusted to compensate for the presence of the conductive components of the HIFU transducers. A T1-W sliding window dual-echo gradient-echo sequence monitored phase changes in the focal zone of each transducer. High-intensity (2400Wcm\(^{-2}\)), short duration (<1.5s) exposures produced sub-therapeutic temperature rises.

Results: On T1-W images, signal-to-noise ratio (SNR) improved by 40% by quartering the conductive surface of the piezoceramic transducer. A piezocomposite transducer showed a further 28% improvement. SNRs for an endocavitary coil in the focal plane of the HIFU transducer (4cm from its face) were 3 times greater than from a phased body array coil. Local shimming improved uniformity of phase images. Phase changes were detected at subtherapeutic exposures.

Conclusion: Combining a HIFU transducer with an MR receiver coil in an endocavitary probe was achieved. SNRs were improved by quartering the conductive surface of the piezoceramic. Further improvement was achieved with a
piezocomposite transducer. A phase change on MR images during both subtherapeutic and therapeutic HIFU exposures was seen.

**Key Words:** Magnetic resonance imaging, High intensity focused ultrasound, endocavitary, thermal ablation, MR guided therapy
Thermal ablation with high-intensity focused ultrasound (HIFU) provides a method of immediate cell killing. It is increasingly being used in the repertoire of anti-tumor therapies, most notably for prostate cancer (1, 2). To be effective it must be delivered accurately to achieve selective destruction of a volume of abnormal tissue with minimal side-effects to surrounding normal structures. However, variations in the acoustic properties of tissue arising from normal heterogeneity can lead to an uncertainty of several millimeters in the position of the focal peak with consequent target sparing and unintended damage to structures surrounding the desired target. The exact position of the HIFU lesion will also depend on the intensity used, the tissue temperature and any tissue movement relative to the transducer during treatment. In addition, experience with laser ablation in the prostate has shown that delivery of an empirical thermal dose may not achieve the desired therapeutic effect (3). It is essential, therefore to monitor the location of the HIFU focus prior to exposure and the degree of local tissue destruction achieved precisely, with sensitive real-time imaging. Attempts have been made to use diagnostic ultrasound for this purpose (4) but ultrasound thermometry techniques are difficult to implement in real-time with HIFU ablation (5). Specialized ultrasound monitoring techniques such as radiation force imaging (6) and echo-strain imaging (7) are being explored. Exploiting the soft-tissue contrast of magnetic resonance (MR) imaging enables the use of techniques for delineating tissue changes in response to ablative therapy (8-11) but real-time information is still necessary for therapy guidance and exposure determination.

There are significant challenges in integrating HIFU therapy with MRI and considerable research effort has been invested in experimental systems (12) which are
being translated to clinical use. HIFU may be delivered using extracorporeal or
docavitary devices (2, 13, 14, 15). These may be single element or multielement
arrays (16, 17), which offer the advantages of control of the focal position by
changing power and phase to the individual elements enabling focusing and
electronic steering over a 3-D volume. As endocavitary MR imaging is a well-
established, routine diagnostic investigation particularly in prostate cancer (18), the
development of an endocavitary device for lesion detection on MR imaging that can
subsequently deliver HIFU treatment under real time MR imaging guidance is
warranted. The purpose of this study therefore, was to develop an endocavitary probe
(for endorectal or endovaginal placement) comprising an MR compatible HIFU
transducer integrated with an MR receiver coil for MR guided thermal ablation.
MATERIALS AND METHODS

Probe Design

The dimensions of the probe casing were constrained by the need for endorectal or endovaginal insertion. The design was based on a previously described solid endorectal MR receiver coil (19). The hollow outer casing was semi cylindrical (diameter 3.2cm) with a rectangular upper surface (7.2cm x 3.0cm) and housed a HIFU transducer (Figure 1) which could be placed in acoustic contact with an intraluminal wall. The distal end of the probe was rounded to ease intraluminal insertion, and the proximal end tapered to a 1.5cm diameter cylindrical handle. Cable connections for the HIFU transducer and MR receiver coil and tubes for circulating cooled irrigation fluid around the HIFU transducer (in order to prevent the high-power device from overheating) passed through this handle (Figure 1). The smaller diameter of the handle meant that there would be no significant discomfort from distension of the anal sphincter or vaginal introitus during the therapeutic procedure.

MR Compatibility of HIFU Transducers

The MR characteristics of a piezoceramic material routinely used in HIFU transducers (PZT-8; Morgan Electro Ceramics Ltd., Southampton, UK) were compared with a MR compatible piezocomposite developed by Imasonic SA (Besancon, France) (20, 21). The piezoceramic transducer contains small quantities of nickel which, due to its ferromagnetic properties, is a well-known cause of magnetic field distortion. However, the nickel is necessary to enable the transducer to handle the high levels of electrical excitation and mechanical stress required to produce the high ultrasound intensities required for therapeutic ultrasound. The same is true for the bismuth sulphate paste used to bond the silver flake coating necessary
for electrical connection to the ceramic element. MR image artefacts may also occur as a result of eddy current generation within the conductive silver coating when placed within a magnetic field. The induced eddy currents cause local magnetic field inhomogeneity and lead to significant image artefact. In order to investigate the significance of this effect, the silvered surface of the ceramic was initially divided into 4 equal-sized electrically isolated elements, thus quartering the large conducting area. If this proved beneficial the device could be segmented into smaller elements to further reduce artefact. The piezocomposite consisted of numerous piezoceramic rods embedded in a polymer matrix and was developed as an MR compatible material. Modification was therefore not required.

All imaging was carried out on a 1.5 T Vision (Siemens, Erlangen, Germany) using an 11cm ring surface coil for artefact measurement and the endocavitary coil (see below) for SNR measurement. Gradient echo sequences (GRE, TR=800ms, TE=26ms, Flip angle 30°, 256X256 matrix, 210mm field of view (FOV) were used as well as sequences planned for the clinical studies (T1-weighted and phase temperature-sensitive Sliding Window Dual-echo Gradient-Echo (SW-dGRE) sequences, TR=35ms, TEs=9.1ms, 31.8ms Flip angle 5°, FOV 200mm, single slice 8mm thick) (22, 23). Images transverse and coronal to the transducer element were obtained with both sequences and magnitude and phase reconstructions done.

**HIFU Transducer Beam Characteristic**

The HIFU transducers were calibrated using standard techniques: complex impedance measurement, pressure field characterisation and acoustic power measurement. Low voltage impedance measurements were made as a function of frequency on a Network Analyser (HP 8712ES, Paolo Alto, California, USA) to assess the need for electrical
impedance matching between the transducers and the drive system in order to optimise acoustic power delivery. The real component of the complex electrical impedance of the 1.46MHz piezocomposite transducer was \( \sim 33\Omega \) and was close enough to that of the drive electronics (50\( \Omega \)) to not necessitate a matching network. Similar results were found for the quartered piezoceramic transducer.

A 0.5mm diameter element GEC-Marcony PVDF membrane hydrophone (Y-34-3598) was used to measure the acoustic pressure profile in three orthogonal directions (one along the sound axis and 2 perpendicular to this). For the “truncated circle” shaped HIFU transducers (43mm diameter x 21mm) the focal region, which is determined by the transducer geometry and frequency and thus was similar for all the transducers tested, was found to be ellipsoidal in shape and approximately 10.2mm x 3.0mm x 1.4mm (-6dB beam width), which compared well with simulated acoustic pressure fields produced using a linear ultrasound propagation model in water (24).

The acoustic power output was determined for each transducer in vacuum degassed water. Calibration information is required over the lifetime of the devices to verify the output stability required for safe clinical use. Measurements were made using a radiation force balance, with reflecting target. For each transducer, free field spatial peak intensities up to 2600 W/cm\(^2\) were obtained, ie above the 1500 W/cm\(^2\) needed for tissue ablation; an in situ intensity of 1500 W/cm\(^2\) (ie accounting for depth in tissue and ultrasound absorption) is required for thermal necrosis (25). It is also generally accepted that HIFU exposures are perfusion independent for exposure times up to 3s.
**Cooling System**

Cooling this internally placed HIFU transducer is critical to avoid temperature rises that may lead to surface burns or device damage. The rate of transducer cooling using an irrigation system was investigated. The probe containing water-filled plastic tubing (internal diameter 0.86mm, outer diameter 1.27mm) mounted through the handle and housing and sited 6 mm above the transducer’s concave surface (Figure 1) and covered in a sheath containing 150ml of degassed water was placed in a water bath maintained at 70°C. A thermocouple on the front of the transducer recorded temperature equilibration. On transferal to a second water bath at 37°C, temperature measurements were made during circulation of cooled water (~12°C) circulated through the plastic tubing. Cooling rates between 10 and 100 ml/min were investigated in 10ml/min increments.

**MR Receiver Coil**

Identical purpose-built, MR receive only endocavitary coils were constructed (6.5cm x 2.5cm rectangular design surface coils) for use with the piezoceramic transducer (SRA) and the piezocomposite (Imasonic) transducer (3). The coil was located on the rectangular upper surface of the probe, outside and above the edge of the ultrasound transducer, (Figure 1) so as to minimize any disturbance of the ultrasound field. Each coil was tuned and matched to a 50Ω output impedance (when positioned adjacent to the HIFU transducers with a typical ex vivo tissue load) during manufacture. The 50Ω impedance matching ensures minimum signal loss in the output cable whilst providing an optimum noise match for the preamplifier input stage. Coil tuning was adversely affected by the presence of the conductive surface coating of the
transducer. The effective inductance of the coil was reduced and the tuning and matching capacitors were adjusted to compensate.

To reduce disturbance of the local B1 field during the body coil excitation period of the MR sequence, the rf current in the receive coil was minimized by decoupling, utilizing an rf trap circuit controlled by a PIN diode switch. This switch also protects the preamplifier input circuitry from high rf voltages. The trap circuit is disabled during the MR receive period. The rf decoupling switch operation is constantly checked by “fail-safe” safety circuitry within the scanner.

A purpose-built MR phantom consisting of a water-filled 13.4cm diameter Perspex cylinder with a hollow 3.4cm diameter central channel along its length representing the rectum was used (Figure 2). The phantom was filled with 0.9% saline (1.1L) and 1ml of Gadolinium DTPA. Each endocavitary probe (coil and transducer in outer casing) in turn was inserted into the central hole. Spin-echo images (TR 1000ms, TE 20ms, Flip Angle 90°, Matrix 256x256, FOV 160mm, NEX=2, slice thickness 2mm) were obtained in a plane transverse to the probe (Figures 3 and 4). For comparison, after removing the endocavitary probes, a body phased array coil (Siemens, Erlangen, Germany) was positioned around the phantom and the MR measurement sequences repeated. Comparable images were chosen at the centre of the phantom and the signal to noise ratio (SNR) was calculated by drawing 2cm diameter circular regions of interest (ROIs) at a point where the HIFU focus would be expected (4cm from the center point of the transducer surface, and at 90° to it) and also in a background ROI (a non-signal producing air region outside the phantom). Identical ROIs were used for all measurements. SNR was defined as being the average image intensity in the ROI at the expected HIFU focus divided by the standard deviation in the background ROI. In addition, for SNR comparison of the
endocavitary probe with the body phased array coil, entire phantom signal measurements also were obtained by drawing a 134mm diameter ROI to match the circumference of the circular phantom. This included the central channel containing the coil.

**Thermometry Sequences**

A SW-dGRE sequence, (TR=35ms, TEs=9.1ms, 31.8ms Flip angle 5°, FOV 200mm, single slice 8mm thick) (23), which reconstructs images every 1.4 seconds, and uses 2 main methods of temperature measurement (phase difference and T₁-weighted imaging) was used for HIFU monitoring.

**Ex vivo Phase Maps**

The endocavitary probe was covered with a rubber sheath (PalMedic®, Paltax bv, Lichtenvoorde, Netherlands) and placed flush with a core of ex vivo tissue in a tank of degassed water. Due to lack of available power from the amplifier used to drive the endocavitary transducer, a second 1.7MHz, 5.6cm diameter prototype MR compatible HIFU transducer (designed in collaboration with, and constructed for research by, Siemens, Erlangen, Germany) was placed on the other side of the tissue and positioned so that its focus (at 7cm from its front face) lay within the tissue, at the focal point of the endocavitary HIFU transducer (4cm from the centre of its front surface). The experimental set-up was positioned in the transaxial (x axis) plane of the 1.5T MR scanner. The endocavitary MR coil was attached to a pre-amplifier and interfaced to the MR scanner. The 1.7MHz HIFU transducer was connected to a power generator and amplifier located outside the magnet room. Connecting cables were appropriately low pass filtered (<2MHz) and passed through the Faraday cage of
the scanner room. The HIFU transmission line incorporated a 63MHz notch filter. This has previously been shown to remove HIFU induced image artefacts (in both GRE and SE images) (26).

Baseline T1 measurements and a T1-W image and phase data were obtained every 1.4seconds during HIFU heating and subsequent cooling. An MR phase image at the end of a HIFU exposure (1750 W/cm² for 1.5s) with the probe adjacent to the liver sample is shown in Figure 5. Initially subtherapeutic “siting” shots were investigated in which high intensity (2400 W/cm²) short duration (<1.5s) exposures were used. In liver tissue ex vivo (n>10), exposures at this level did not produce any macroscopic tissue change. Such shots have previously been used to establish the position of the focus in tissue prior to a HIFU exposure (26). At a separate location, lesion formation was also monitored using high intensity (1750 W/cm²) for longer durations (6s). In liver tissue ex vivo, such exposures produce macroscopic lesions ellipsoidal in shape (Fig 6) and have previously been calibrated in ex vivo experiments to produce specific temperature rises (22).
RESULTS

Transducer MR Compatibility

Piezoceramic Effect of Quartering

As expected, a dramatic reduction in the magnitude of the rf eddy currents was achieved by quartering the conducting area of the piezoceramic transducer. This was evidenced by a reduction in image artefact. A comparison of artefact before and after quartering is shown in Figure 3. Corresponding transverse image slices for each sequence selected for both transducer configurations and analysed using the scanner’s software showed that the SNR on the T1-weighted images at the 4cm focal plane of the transducer was 40% greater using the quartered transducer (Table 1) compared to the non-quartered single element transducer. (SNR for the quartered transducer = 95.9, SNR for the non-quartered transducer = 68.1). This difference in SNR was sequence dependent. With the SW-dGRE sequence SNR for the quartered transducer was 143.9 and for the non-quartered transducer was 108.5 (32.6% improvement on quartering).

Piezoceramic versus Piezocomposite

MR compatibility, as shown by variations on the phase maps of the piezocomposite device, was demonstrated to be significantly better than that of the original piezoceramic HIFU transducer. A direct comparison using T1-weighted sequences is shown in Figure 4 with an 11cm diameter surface coil. The extent of the artefact from the unquartered piezoceramic transducer on the gradient echo images was 1cm compared to 3mm for the piezocomposite (Table 1). The improvement in SNR was 28% compared to the quartered piezoceramic (SNR piezocomposite =122.4) (Table 1).


Cooling System

Results showed that a flow rate of 50 ml/min could be achieved in the tubing built into the transducer casing, and that this produced a substantial cooling (up to 11°C in 10s). In vivo the 150 ml water reservoir would not initially exceed body temperature, and with pre-circulation this could be kept substantially lower than body temperature to prevent thermal damage both to patients and the HIFU source. Furthermore the intended duration of HIFU exposure (<10s) should not lead to temperatures as high as the 70°C that was tested in our developmental set-up. Similar water cooled rectal applicators have been used successfully in other endorectal thermal therapy devices (27, 28).

MR Receiver Coil Performance

Each coil alone had an unloaded Q factor of 332 and a loaded Q factor of 150 at 62 MHz. With the introduction of the transducer element, the unloaded Q fell to 220 (piezoceramic non-quartered, 177 piezoceramic quartered), and 109 (piezocomposite) (Table 1). The piezocomposite figure is likely to be related to a higher resistance of its conductive copper coating. Tissue loading reduced these Q differences, although this reduction was less for the piezocomposite indicating a greater loss due to the transducer. However these Q differences would not translate to significant SNR differences.

As it was decided to proceed with the use of the piezocomposite transducer in vivo, because of smaller artefacts SNR measurements made with a piezocomposite transducer combined with the endocavitary coil were compared with a body phased array. SNR calculated for the endocavitary coil at the focal point (4cm) of the HIFU
transducer were 3 times greater than those for the body phased array coil (endocavitary coil SNR=637; body array SNR=205). A ROI encompassing the entire phantom yielded an SNR value for the endocavitary coil that was nearly twice that of the body phased array coil (endocavitary coil SNR=301; body array SNR=172). Thus, the endocavitary coil produced a SNR significantly greater than that achieved by the body phased array coil routinely used at our hospital for pelvic imaging. When the endocavitary coil was combined with the piezocomposite transducer, the SNR gains normally expected from such a coil alone were not realized (19), but still offered significant SNR advantages over a body phased array coil.

Since each coil was individually tuned to each transducer, it was not possible to obtain useful SNR measurements without the transducer present.

Ex Vivo Heating Experiments

The ability to produce phase images capable of detecting changes equivalent to subtherapeutic temperature rises (22) was clearly demonstrated (Figure 7). Changes could be clearly seen on the phase images without the need for post processing of the data. Implementation of local shimming in the volume of the image slice produced phase images with greater phase uniformity (Figure 7). Additional shimming, achieved less phase uniformity (light to dark variation) across the tissue sample. Thus this approach is recommended for future use.
DISCUSSION

We have shown that combination of a HIFU transducer with an MR receiver coil in an endocavitary probe is achievable and that phase changes were visible on MR images at the focal point of the transducer during HIFU exposures known to produce both subtherapeutic and therapeutic effects. Quartering the piezoceramic transducer’s surface successfully resulted in a 40% improvement in SNR because of reduced induced currents and hence a reduction both in the receiver coil losses and in B1 uniformity. Additional sub-division probably would improve SNR further, but other problems arise, such as the reliability of the increasing number of electrical connections. The further improved SNR of the piezocomposite transducer allows better visualization of the region in which tissue will be ablated and permits higher spatial resolution imaging of locally sensitive tissues near the probe surface, such as the rectal mucosa and neurovascular bundles in the case of endorectal use. Newer piezocomposite materials are comprised of numerous ceramic elements mounted within a polymer matrix: standard PZT material is diced and a polymer added (20). These effectively form individual isolated conducting units and reduce eddy currents. The “silver” coating on the PZT is removed in the dicing process and replaced on the composite with a layer of copper (by chemical deposit). This is likely to account for the improvement in MR compatibility.

Artefacts measured on gradient echo images even with the unquartered piezoceramic did not reach the focal zone of the transducers (4cm from their surface). However, despite quartering the piezoceramic transducer it produced a larger artefact than the piezocomposite transducer indicating more local field inhomogeneities. The resulting phase shifts can make local shimming difficult and can lead to poor or non-visualization of small temperature rises achieved at sub-ablative exposure levels. We
therefore did not pursue the use of a piezoceramic transducer for this application. With the piezocomposite transducer, artefacts were smaller, which indicates less local field inhomogeneity, and makes the resulting phase shifts easier to correct with additional local shimming. In routine clinical endorectal imaging without the confounding presence of a HIFU transducer, local shimming is not usually affected by tissue heterogeneity or even the presence of calcification.

A loop design was chosen for the endocavitary receiver coil since it provides the best SNR/FOV performance compromise within the anatomical constraints. Other designs have been investigated, particularly in the field of intravascular imaging (29). The ‘opposed-solenoid’ design has been shown to approach, but not equal, the SNR performance of the loop design, but its FOV is limited to the region between the two solenoid structures, confining imaging to a restricted number of slices (depending on slice thickness) transverse to the coil axis. Imaging of other areas of the prostate would require the coil to be moved. The loop design allows imaging in any slice plane, covering most, if not all, the prostate without moving the coil.

Current clinical therapy with endocavitary HIFU is limited to prostatic applications. Though few incidences of major morbidity have been reported (30), localisation currently is achieved only with ultrasound imaging of the whole prostate gland prior to HIFU exposure. Using a HIFU device combined with a MR receiver coil allows localization of a subtherapeutic HIFU focus and enables a strategy of “siting exposures” during a procedure whose overall time means that there is a very high likelihood of patient and hence target movement. It also avoids relying on assuming the focus will lie at the geometric focus, or the focal peak measured in water. In addition, temperature measurement away from the focus is possible and is particularly important in avoiding inadvertent damage to sensitive surrounding tissue,
such as the genitourinary diaphragm, neurovascular bundles, and rectal mucosa. Also, as the treatment progresses, there is significant tissue edema (3, 31) which may move the ‘target region’ away from the initial prescribed treatment volume. Reimaging to establish the change in lesion position with a “siting” exposure in the required area should result in a more effective and accurate destruction of the lesion. Endocavitary probes may also be placed transurethrally and with highly directional energy deposition and rotational control can achieve controllable coagulation of a 270° contiguous section of the prostate extending to the capsule boundary (28, 32).

Other potential real-time methods besides MR for monitoring HIFU exposures involve specialized ultrasound techniques. For example, use of the radiation force of the therapeutic ultrasound beam as an elastographic push to detect relative stiffness changes (6) required for lesion formation have been investigated. Echo-strain imaging is another technique being explored for monitoring formation of thermal lesions (7). Ultrasound elastography requires compressions to achieve a measurement, which are performed with an endorectal balloon, but data so far show that although there was a statistically significant correlation between elastographic and MRI measurements of HIFU lesion volume, elastographic measurements were unable to predict MRI measurements of HIFU lesion volume in a single individual (33).

The recently presented results of the European Multicentre Study based on 400 patients with a follow-up of more than 1 year show that transrectal HIFU for prostate cancer is a valid alternative for the management of well differentiated and moderately differentiated localized prostate cancer (with an initial PSA ≤15 ng/mL in men with a life expectancy >10 years (34)). The best indications for HIFU currently are men over the age of 65, those who are not candidates for radical prostatectomy or external
beam radiation treatment, or patients with co-morbidities likely to make surgery more difficult. Also, HIFU can be repeated in cases of localized recurrence (unlike any ionizing radiation based treatment) or to re-treat a prostatic site; it involves no radiation, and patients do not suffer from long-term urinary symptoms. The main clinical anxiety surrounding transrectal HIFU ablation is the potential for damage to the interposed rectal mucosa, which could result in a urethra-rectal fistula (35). Reports suggest this is rare and although more likely to heal spontaneously with simply urinary diversion (catheterization) in the absence of radiation damage or residual tumour, the requirement for invasive interventions to repair any damage remains. Poor localization of the ultrasound focus has the potential to cause other clinically relevant damage. Depending on the extent of an inaccurately located focus, damage could include urethra (with later stricture), urinary sphincter (with possible incontinence), ureteric, neurological (e.g. cavernosal nerves affecting erection), vascular and even bowel injury.

In gynecological use, indications for HIFU would primarily include vaginal and cervical dysplasia, and small recurrent neoplastic lesions (36). Recto-vaginal endometriosis forms another clinical entity where endovaginal HIFU may usefully be employed. The close proximity of the ureters to the cervix would pose a risk and benefit from precise image guidance.

Safety issues in this combined endocavitary HIFU device and MR receiver coil are paramount. Cooling system leaks could potentially damage coil electronics and lead to coil failure. This is addressed by coating the coil and its circuitry in epoxy resin, but means that any electrical faults would entail coil replacement rather than repair.
A clinically useful device would be required to produce multiple lesions at several locations within the prostate gland. Transducer arrays with electronic beam steering are increasingly investigated to address this problem (37). In addition, the SNR of the images may also be improved by the use of endocavitary array coils (38). Finally the short bore tunnels of most MR scanners provide restricted access to the patient. The use of robotic arms to manipulate endocavitary probes in situ within the MR scanner that can be controlled remotely by an operator, are also in development.

In future, the understanding of tumor targeting and tissue changes derived from MR studies may result in a combined diagnostic and therapeutic out-patient package designed to achieve control of localized disease using MR guided HIFU without recourse to major surgery in an elderly population. Further advantages of the technique are the repeatability of the treatment and the potential for low toxicity to normal tissues.
REFERENCES


Table 1  Characteristics of HIFU transducers and MR receiver coil used in the endocavitary probe design

<table>
<thead>
<tr>
<th>Artefact measured using 11cm surface coil on T1W GRE</th>
<th>Q of coil combined with</th>
<th>Q of coil ‘loaded’ with tissue and combined with</th>
<th>SNR using endocavitary coil (4cm from focal plane on T1-W SE (1000/20/90⁰))</th>
</tr>
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<tbody>
<tr>
<td>Piezo ceramic non-quartered</td>
<td>1cm</td>
<td>220</td>
<td>141</td>
</tr>
<tr>
<td>Piezo ceramic quartered</td>
<td>1cm</td>
<td>177</td>
<td>114</td>
</tr>
<tr>
<td>Piezo composite</td>
<td>3mm</td>
<td>109</td>
<td>87</td>
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FIGURES

**Figure 1** Endocavitary probe consisting of a piezocomposite HIFU transducer (1.7 MHz, 43 mm X 21 mm, focal length 4 cm) mounted in an inner casing with MR receive only coil (tuned to the transducer load) mounted around it in an outer casing. A sheath (not shown) fitted over the HIFU transducer would be clamped around the outer edge of the inner casing using a rubber o-ring and filled with cooled degassed water through the inlet pipe whilst ensuring that air is evacuated down the outlet pipe. The route of the water pipes, located between the inner and outer casings, is shown schematically. This arrangement separates the MR coil from the water compartment required for coupling the HIFU transducer to body tissue.

**Figure 2** Photograph of MR phantom showing 3.4 cm diameter central cavity (arrow).

**Figure 3** Effect of quartering the piezoceramic transducer: Gradient echo MR images (TR=800 ms, TE=26 ms, Flip angle 30°) showing the effect of quartering the conductive surface of the piezoceramic transducer. In (a) the single-element transducer shows a “blooming” effect of the signal void (arrow) around the transducer. In the quartered transducer (b), this effect is much reduced (arrow), and the signal to noise ratio is improved.

**Figure 4** Artefact from piezoceramic compared to piezocomposite transducer elements: T1 weighted images (GRE TR=800 ms, TE=26 ms, Flip angle 30°) obtained with an 11 cm surface receiver coil showing the artefact around a
piezoceramic transducer (a) compared to a piezocomposite transducer (b). Artefact around the transducer is much greater in a than b.

**Figure 5**  Experimental set-up used for monitoring ex vivo lesions:  MR image in longitudinal section through the endocavitary probe (piezocomposite HIFU transducer and MR receiver coil, medium arrow) mounted in a degassed water bath with a section of bovine liver (short arrow). A separate MR compatible HIFU probe (1.7MHz, 5.6cm diameter, 7.3cm focal length, long arrow) positioned confocally with the endocavitary piezocomposite transducer was used to induce temperature rises in the tissue, which were imaged using the MR receive only coil.

**Figure 6**  Ex vivo liver tissue following 6 sec exposure at 1750W/cm². The ellipsoidal whitish lesion with clearly demarcated edges (arrow) is easily identified.

**Figure 7**  Ex vivo heating experiments with MR monitoring: Phase images without shimming (a), with shimming (b), with shimming and during HIFU exposure (c) using a Sliding Window Dual-echo Gradient-Echo sequence, (TR=35ms, TEs=9.1ms, 31.8ms). There is a marked improvement in homogeneity in b compared to a. In c the area of elevated temperature appears in cross-section as a small white spot (arrow) slightly off centre within the sample. The ring is the Perspex holder used to contain the ex vivo liver sample.