A Robotic Approach to HIFU Based Neurosurgery

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Abstract. The use of robotics in surgical interventions not only has the potential for minimally invasive surgical procedures but can improve performance and result in reduced operative time and post-operative trauma/recovery. This paper describes the concept of a robotic based High Intensity Focused Ultrasound system as a neuro-surgical tool for the destruction of subcortical lesions. A novel multi-transducer applicator system is proposed in order to minimise the effects of off-focal hot-spots and cavitation. Analytical models have been developed for simulating the acoustic field of the multi-transducer system. The models predict the interactive field effects from specific spatial configurations of the probes with respect to each other and to the target. Finally, the design aspects for a robotics based dedicated manipulator for HIFU-based brain surgery have been explored, together with those predicted from a laboratory system.
Keywords: Neurosurgery, Surgical Robots, Ultrasound Surgery, Treatment Planning

1 Introduction

Because of its complex anatomy and vital functionality, brain tissue surgery has always remained a formidable medical challenge. The need to devise precise and minimally invasive surgical methods is, therefore, more demanding for the brain than for any other organ or part of the human body. A widely used and well-established technique for neuro-surgery, particularly for point-in-space applications such as taking tissue biopsy, is stereotactic surgery. This method has been adopted by other treatment disciplines as well, (for instance in radiation therapy and surgery, in seed implantation in the brain, etc.) because it gives accurate registration of surgical tools. The disadvantage is that they are invasive and the reference base-frame needs to be clamped/screwed to the patient's skull, both in the pre-operative and intra-operative stages. Moreover, the frame may obstruct access to the surgical site or constrain the motion of the tools. To alleviate these problems, frame-less techniques have recently been introduced. These frame-less techniques however, can be less accurate and are presently in the developmental stage.

To avoid the risks involved in open brain surgery, efforts are being made world wide to develop minimally invasive surgical methods such as Computer Assisted Surgery

(CAS) and robotic surgery. Surgical targets in stereotatic surgery, and in most contemporary CAS procedures, are usually not under direct real-time vision. However, tool trajectories may be guided by registering and projecting their position onto 3-D reconstruction on the computer monitor screen during the operation. It is difficult to use these indirect vision methods to achieve precise micro-surgical manipulations with appropriate hand-eye co-ordination. Robotic techniques are more accurate, and require less operating time than human actions, particularly in tasks requiring repetitive actions.

In traditional open surgery, there is a need to create an access wound in the tissue. The surgical process involving opening the tissue may cause anatomical shifts, apart from making the procedure invasive. For brain tissue, in particular, these incisions may prove fatal in some cases. Also, the opening of the *dura* may cause severe loss of blood and cerebrospinal fluid, which in turn may result in large pressure changes and motion of the brain tissue. These risks are more severe if the lesions are bigger in size and/or are deep-seated. To minimise these complications, it is desirable to devise minimally invasive or non-invasive means of surgery.

Recently, High Intensity Focused Ultrasound as a non-invasive surgical modality has shown promising results, particularly in the areas of urology and oncology [1,2]. HIFU surgery is based upon the thermal effects of ultrasound and is beneficial if well-controlled thermal doses (without cavitation) can be administered in desired locations. It is necessary to study the requirements of each specific case and to decide upon the safety limits by considering the effects of all the exposure parameters involved in the treatment planning phase, prior to actual surgery/irradiation. In order to study the behaviour of the ultrasonic beam, as it traverses through the tissue, computer models have been developed and simulation results are presented in this paper.

In the case of brain tissue which is encased in the bony envelope of the skull, a direct application of the ultrasound energy is difficult because bone reflects a large amount of the incident energy. Therefore, it is necessary to create access craniotomies of considerable size in order to place the applicators directly on the *dura mater* (via an appropriate acoustic couplant). This however, should not introduce any risk because the external skull can be grafted and sutured and the whole procedure is still non-invasive to the sensitive brain tissue.

1.1 A New Multiple Probe Approach

It is necessary to minimise the effects of cavitation at high intensity levels as desired for tissue ablation and to avoid the off-focus hot-spots that are inherent in conventional HIFU systems and electronic arrays. To achieve this, a method is proposed in this paper for using separate multiple probes to simultaneously ablate the target from different angles. This ensures that the total required power is divided into low intensity beams. The probe selection, its dimensions and the applicator power play a crucial role in making the system minimally invasive. The concept of providing the required thermal dose from well spaced multiple probes favours minimal access. Thus two or three small craniotomies will be required, which can be planned in safe regions on the skull surface and for which appropriate focal lengths can be chosen. This will be advantageous only if the surgery system is designed to avoid constructive interference outside the composite focal region of the participating beams, thus avoiding hot-spots in the normal tissue regions. Moreover, the dimensions of the focal region should be planned in such a manner that it is confined well-within the abnormality. This may be particularly difficult to achieve in the case of small lesions. Alternatively, the probe may need to be scanned over the whole volume of the lesion when the lesion is larger than the composite focus. The development of such a system, which requires robotic techniques to simultaneously move and focus the probes, is the underlying aim of the present research and will be described in the following sections.

Ultrasound provides an indirect form of surgery in which the cutting formats are not straightforward (in contrast to the case of well defined surgical cuts using routine clinical tools e.g. knives, scalpels, scissors etc.). It is thus necessary to study the beam characteristics in terms of power, size and shape when it propagates through the tissue. For this purpose, mathematical models have been developed. These predict the interactive field resulting from the joint effect of multiple transducers in a variety of spatial configurations and excitation conditions. The models developed at Imperial College are summarised below. Validation work and parametric studies of transducer properties and positions along with the field modifications by introducing a diagnostic probe coaxially at the center of one of the surgical probes (for online feedback) will be reported elsewhere.

2 Computer Simulations

Models for predicting the beam patterns generated by a focused transducer are well established [3]. They typically make use of integrals of the fields due to point sources or plane waves. In our models, we exploit the point source approach which is formalised by Huygen's principle and the Fresnel Kirchoff's diffraction theory [3] and we extend the concept to incorporate the interactive field of multiple transducers. Field characteristics of an ultrasonic transducer such as the radial and transverse power amplitude, intensity profiles, beam plots at any point in the field both for two dimensional (cylindrical wave equation) and three dimensional (spherical) coordinates can be studied by using these models. The results in this paper were obtained using the two dimensional model. The software is written in such a way that almost all the field parameters can be entered as desired (for instance, the size and shape of the probe, the frequency of operation, focal length, velocity in the sound medium, axial and radial distance, number of Huygen's pts per mm etc.). This flexibility is necessary in order to understand the effects of individual variables on the beam characteristics. The co-ordinate system, showing three probes in a plane for the 2-D calculations are illustrated in fig.1. The model assumes plane strain in the plane of fig.1, so that each point in this plane represents a line source. The probes are arranged so that they produce a confocal superimposed focus within the target region. The multi-transducer model is initially developed for three probes only, but can be extended to any desired number within the defined co-ordinate system. The overall field distribution obtained from the interference of individual probes is given by complex summation of individual potentials and then plotting the absolute value of the sum, solving the wave equation for cylindrical 2-D calculations for the constituent probes. The acoustic pressure amplitude p(r,t) at any arbitrary point P(r,θ) in the field at a distance r from the source and time t is given by [3]:

$$p(r,t) = A\sqrt{\frac{2\lambda}{\pi r}}.e^{i(\omega t - kr)} \tag{1}$$

where, λ is the wavelength of sound, ω is the angular frequency, k is the wave number and A is an arbitrary constant. Only the wave travelling in the positive z-direction has been considered (fig.1). For large *r*, the cylindrical wave can be approximated as a plane wave with pressure amplitude decreasing as $1/\sqrt{r}$. The intensity, in terms of pressure amplitude is calculated as:

$$I = \frac{p^2}{\rho . c} \tag{2}$$

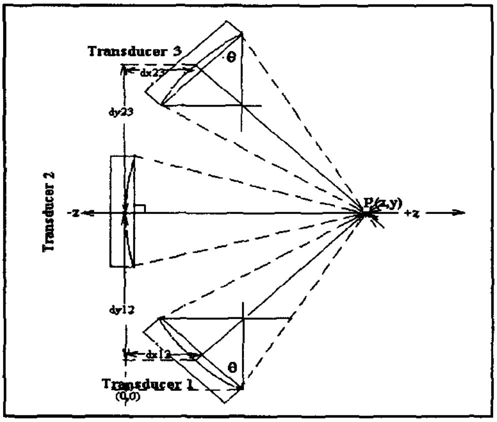


Fig.1. The planner co-ordinate system used for modelling the three transducer system.

The spatial configurations (location & orientation) of all of the probes can be separately entered and thus the corresponding effects of individual probes on the composite intensity/beam profiles can be studied. Initially, a linear propagation of sound waves is assumed in a homogenous and loss-less medium for the purpose of calculating the field parameters. In practice, biological tissue is not homogenous and the ultrasound energy is attenuated due to reflection, refraction, scattering and absorption during its passage to the target. The fraction of the incident energy that is absorbed gives rise to the desired focal hot-spot for ablation. The original model has been extended in two ways: first to incorporate the attenuation effects ($e^{-\alpha x}$) as a function of propagation distance(x) and second to calculate tissue temperature rise by solving Pennes' bio-heat equation (expressed for one dimension):

$$\rho . c (\frac{dT}{dt}) = \kappa (\frac{d^2 T}{dx^2}) + \omega . c_b (T_a - T_b) + q \tag{3}$$

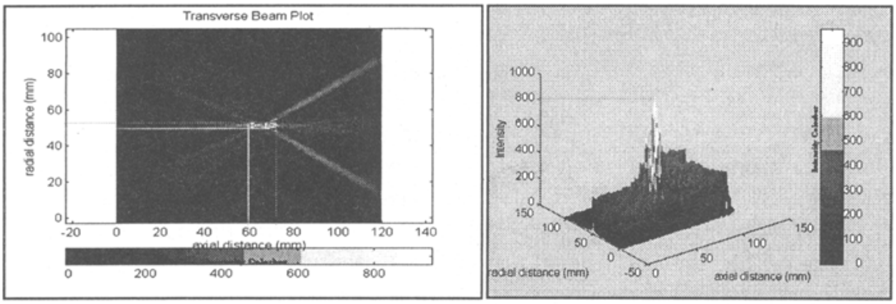
where T is the tissue base temperature, T_a is the local arterial blood temperature, T_b is the local venous temperature, c_b is the Sp. heat of blood, c is the Sp. heat of the tissue, ρ is the tissue density, ω is the blood perfusion rate, q is the volumetric heat generation rate and κ is the thermal conductivity of the tissue. The effects of blood perfusion during short ablation periods are assumed to be negligible ($\omega = 0$). With this assumption, the temperature elevation in the exposed area was computed by adding the effects from individual Huygen's sources, similar to the calculation of field intensities in the previous models.

The model was further extended to include the effects of heterogeneity of the brain tissue as seen by the propagating ultrasound wave. The acoustic properties of grey and white matter are found to be widely different [4]. The interface of various layers has been considered as a straight line for the given size of the probes. This is a preliminary approach to the practical *in vivo* situation and only the differences in propagation velocity and attenuation characteristics of the constituent media have been considered. Initially, it is also assumed that the beam propagates from one medium to another following refraction and attenuation only. It is assumed that there is no acoustic internal reflection back from the skull, which is quite reasonable since the individual low energy beams, in the domain of selected focal length, becomes very feeble after the joint focus, as will be shown in the simulation results. The number of absorption layers can be inputted as desired. Further development of this model is planned to include curved interfaces together with the effects of reflection and refraction at the boundaries.

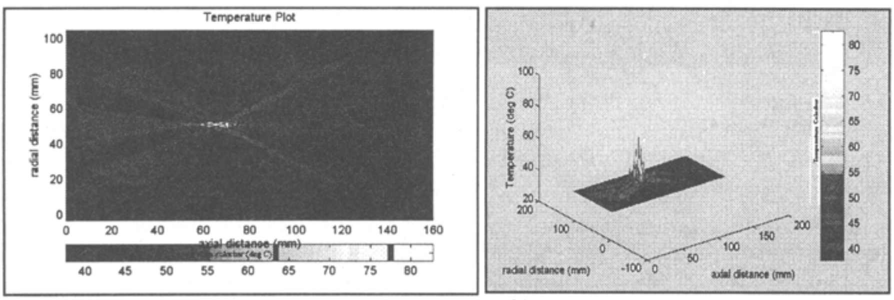
3 Simulation Results

The simulation results obtained by using the specifications of transducers operating at a resonance frequency of 1.88 MHz, 25.4 mm active aperture and focused at 63.5 mm have been presented in fig.2 and 3. The transducer specifications are those used in an experimental study to validate the acoustic field model, which will be reported elsewhere. Fig.2 shows the field computations for a particular spatial arrangement. Here the probes are considered to be positioned on an arch of radius equal to the focal length and oriented in the plane of the arch at an angle of 35^0 with respect to the middle probe, such that they produce a superimposed confocal area. The intensity is plotted as a dimension-less ratio, with maximum intensity at the focus normalised to unit intensity at the source. In fig.2(a), intensity variation along axial and radial dimensions in a loss-less medium has been illustrated. Fig.2(b) shows the temperatures produced in the exposed area by considering power absorption in the medium. For temperature estimations, the base temperature is taken as 37.6^0 C considering an absorption coefficient of 0.7 dB/cm/MHz$^{1.1}$ as suggested for human soft tissue [4,5]. The plots clearly show that for a two sec pulse, a highly focused region of 65^0 C average is produced whilst the off-focus regions are at about 42^0 C. These results are ideal for very localised ablation of tumours etc.. Thus, it is possible to predict the temperature elevation in the focal region as well as throughout the irradiation field (region of interest) of the probes.

The simulation results of fig. 3 show the effects of absorption of the acoustic beam during its passage through biological media. Fig.3 (a) and (b) depict a comparison of the axial beam intensities in a loss-less medium and with those in an absorbing medium (with a=0.7 dB/cm/MHz$^{1.1}$) for a single probe. As is evident from these results, the joint focus is smaller in size, the overall intensity is lower and the focal peak shifts towards the surface (transducer face) when absorption effects are introduced. The axial beam characteristics through multi-layered tissue are shown in fig 3(d). Fig. 3(c) shows the schematic of the field composed of three layered tissue and the corresponding axial beam trace is shown in 3(d) for typical layer thickness and absorption coefficients lying in the range of grey and white matter (1.08 - 1.93 dB/cm at 1.88 MHz). Depending upon the ultrasonic characteristics of the medium in the path of constituent probes, the focal region may change its geometrical characteristics (location, size and shape) and field intensity values as compared to that predicted in either a loss-less medium or homogeneous absorbing medium. By pre-operative estimations of the type and depth of tissue layers, in the path of individual acoustic beams emanating from constituent probes, it is possible to predict field modifications. Further developed model would be used to determine optimum positions and treatment according to location of tumour.



(a)



(b)

Fig.2. Simulation results: axial and radial intensity profiles for the multi-probe system in a loss-less medium (a); and temperature predictions (b) for a medium with α=1.19dB/cm at 1.88 MHz, base temp. =37.6 for 2s exposure.

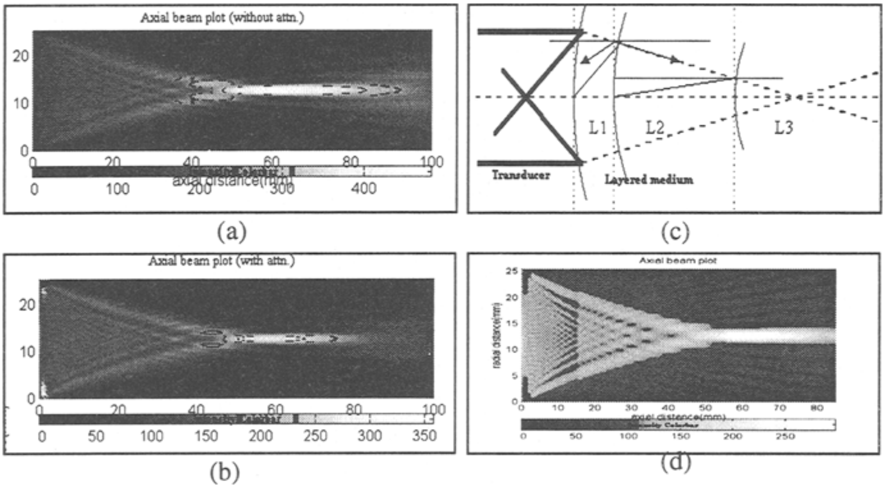


Fig.3. Predicted axial beam plots: in a loss-less medium (a) and in a homogenous absorbing medium (b) with a = 0.7 $dB/cm/MHz^{1.1}$; 3-layered medium schematic (c) and beam plot calculated for L1=20mm, α=1.93dB/cm; L2=40mm, α=1.09dB/cm; L3=30mm, α=1.19dB/cm.

4 Instrumentation

The size of the superimposed foci of a multi-probe system may not always cover the desired target. The frequency and intensity selection govern the foci dimensions but they cannot be varied indefinitely. By considering various spatial configurations, power inputs and frequencies of individual probes, it is possible to diffuse the focal region for slight modifications in shape and size with an overall effect on intensity generation. This may suffice for the ablation of some small tumours. For ablating larger lesions, it therefore, becomes necessary to mutually scan all the individual beams together, over the area of interest, either mechanically or electrically. Electrical phased arrays have several advantages over traditional single focused probes such as: electronic scanning involves no physical moving parts, the focal point can be changed dynamically to any location in the scanning plane and the system can generate a wide variety of scan formats for heating any shape of target. However, the most critical disadvantage is the formation of constructive interference zones which produce pseudo foci beyond the actual focal region. Other inherent disadvantages include: increased complexity of scanning electronics (particularly for large arrays), higher cost of transducers and scanners. Also the requirement of a large number of elements and greater complexity will result in large apertures to achieve high quality images and the required high intensities.

The procedure of mechanical scanning of the focused beams over the abnormality, however, will become more complex in the case of multiple probes. The problem can be segregated into two steps: Firstly, it is required to exactly match the foci of multiple beams such that they intersect within the target boundaries and secondly to maintain the co-ordinated focusing while they are being scanned. The second stage will be more difficult to implement, because of the motion constraints (the probes are not free to move as in the case of extra-corporeal methods but have to be seated and scanned within fixed craniotomies). This would require complex and precise manipulation strategies through the use of robotic techniques. A schematic diagram of the ultrasound surgical system is shown in fig.4.

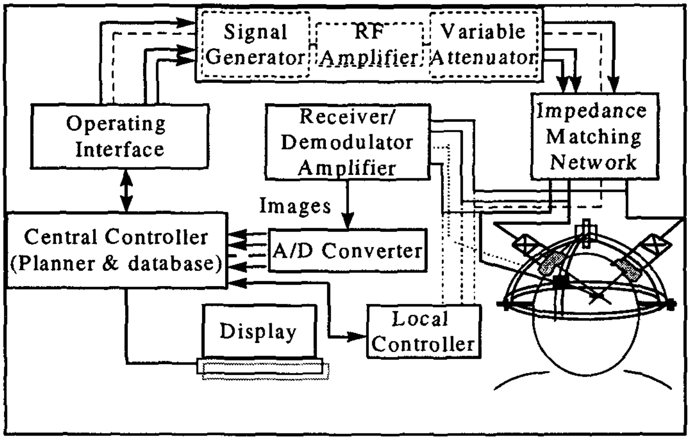


Fig.4. Robotic HIFU Neurosurgery System

Time gated, high intensity continuous sinusoidal bursts are applied by a radio-frequency power generator and are then fed to the transducers via a variable attenuator. Depending upon the specific application and tissue under investigation, the attenuation levels and time settings are varied. The transducers are also impedance matched with the r.f-generator circuitry for an efficient energy transfer. The transducers are coupled to the tissue of interest by the use of a couplant assembly which comprises fluid infusers/diffusers to regulate and control the amount of couplant in order to minimise the effect of abnormal pressure build-up of couplant over the sensitive tissue, while mechanically scanning the target.

4.1 Design aspects of the manipulator system

Surgical Robots. The Mechatronics in Medicine Group at Imperial College (IC) has a long-standing research activity in Robots for Surgery. Special-purpose systems have been produced for Prostatectomies and for Prosthetic Knee Replacement [6,7] It is felt that industrial robots, designed for their large reach envelope, are less suitable than special-purpose systems which can be designed to be simple, small and light to suit the specific task. Robots in Industry are required to be caged away from human contact for safety, which is clearly not practicable for surgery. For this reason, industrial robots have often been used in neurosurgery to position a fixture next to the head and then be locked off, unpowered, whilst the surgeon used the fixtures to carry out manual interventions. This was felt to be safer than using the industrial robot to actively remove the tissue and it was quite clear that the surgeon, and not the robot, was in-charge of the procedure. The only special-purpose neurosurgery robot to be used clinically (Minerva) was a large structure, which, although very accurate, was very complex and expensive [8]. The IC general approach to surgery robots is to use a small special-purpose active robot mounted upon a larger " gross positioning" robot which can be locked off, unpowered, once the active robot is in approximately the right location. By this means, the active robot can be made simple and low cost, with a small applied force and motion capability to enhance safety.

Motorised HIFU. In order to sweep the focus of a HIFU probe and ablate a volume, it is common to mount the probe on a motorised, three Cartesian axes, system. Alternatively, an electronic array may be used to adjust the focal depth, whilst the probe is scanned in a plane by a two-axes system. A hydraulic powered, plastic 3-axes manipulator has also been used in a closed MR system to provide real-time imaging of ablated tissue during HIFU ablation [9].
The IC approach to multiple-probe HIFU can be achieved in its most versatile form by use of a 3-axes system for each probe, and using a robotic control system for co-ordinating the probe motions. However a simpler, lower cost system can be used for a number of procedures if the probes shown in fig. 1, are rigidly mounted together in a single block. Thus the block geometry provides accurate focusing of the probes, whilst a simple 3-axes system sweeps the combined focus through a volume. In practise the probes typically require an offset angle of 35 degrees to each other and so the combined block becomes large. Whilst in some aspects of neurosurgery a large skull flap can be removed, which can allow access for the block via a flexible couplant bag interposed between the dura and the probe. However, the range of these applications is limited in neurosurgery, although there are potential benefits for many other areas, such as tumour ablation in the liver or the prostate.

4.2 The Neurosurgery HIFU Robot

In its most widely applicable form, the neurosurgery robotic system has been designed to position each of the 3 probes independently at any of 3 burr (entry) holes on the skull. This is achieved by a modified form of stereo-tactic frame (Fig. 5) in which the normal diametral arch has been replaced by 3 quarter-arches, joined at the top swivel-point so that each is motorised and free to rotate independently around the base-ring. The 3 probes are located on a probe- mount which can be driven along each quarter-arch. Thus any probe-mount can be located at the computer co-ordinates of any selected burr-hole on the skull, where it can be locked in position to facilitate safety. The probe-mount also serves as a fixture to facilitate burr-hole drilling.

Each probe-mount has 3 powered axes of motion: a pitch, a yaw and an in/out relative to the pitched/yawed mount (Fig 6). The tip of the mount carries the HIFU probe, to which is attached a flexible bag containing the coupling medium. The 3 axes of each probe motion allow its focus to be located anywhere within the target. A robotic control system co-ordinates the 9 motors of the 3 probes, to allow the 3 foci to overlap at a precise position which can also be swept anywhere in the brain. Alternatively the foci can be juxtaposed to give a larger region of heating. This 3 probe system requires the small, low-powered motors to have co-ordinated motions, but can provide total versatility in positioning and sweeping the probe focus. As commented in the previous section, however, simpler, cheaper systems are possible for particular procedures which require a less flexible approach.

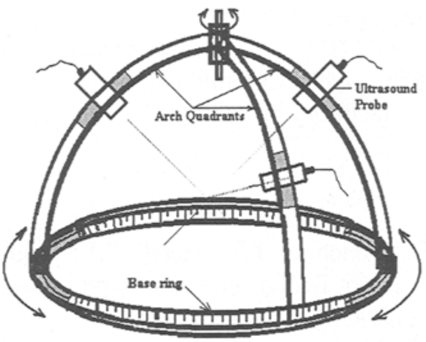


Fig.5. Modified Head frame.

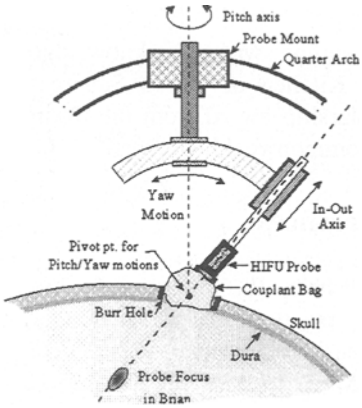


Fig.6. Schematic diagram of motions of a single probe.

5 Conclusion

The use of High Intensity Focused Ultrasound as a surgical modality is highly promising for minimally invasive procedures e.g. as required in neuro-surgery and may also result in a totally non-invasive treatment for other disciplines, which do not require any access wounds e.g. for surgery requiring abdominal access. However, it is necessary to undertake detailed studies in order to understand the interaction of highly intense ultrasound beams with biological tissues, normal and abnormal, in terms of a well-defined mathematical understanding. Numerical models enable the prediction of the after-effects of pre-defined exposures leading to planning of safer treatments. Simulation studies using a multi-transducer approach have been presented in this paper. The aim with this configuration is to minimise the effects of hot-spots in the off-focal areas in comparison to a single transducer.

Further experimental investigations are being carried out in order to study the attenuation effects in mammalian tissue *in vitro*. It is recognised that these are preliminary predictions and it is required to further explore the physical phenomena such as non-linear propagation, scattering effects from tissue heterogeneities & curved tissue boundaries and dynamic changes in tissue characteristics during the process of irradiation in order to simulate actual *in vivo* situations. The design and development of appropriate instrumentation for the precise, accurate and complete irradiation of a defined tissue volume is equally important. In the case of the multi-probe system, mechanical scanning of the joint focus in the target area (for ablating bigger tumours) is a crucial aspect of the instrument design. To achieve this a mechanical mechanism for robotic based HIFU system as described in this work, has been designed and developed at Imperial College. Further improvements with appropriate safety considerations are being carried out.

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