MODELLING BEAM DISTORTION DURING FOCUSED ULTRASOUND SURGERY IN THE PROSTATE USING K-WAVE

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Introduction

Prostate cancer is one of the most commonly occurring cancers for men in Europe and the US, and a leading cause of cancer-related death [1]. For patients with early-stage localised disease, the cancer is often treated using external beam radiation therapy (EBRT) [2]. This procedure usually involves implanting a small number of gold fiducial markers into the prostate to verify the position of the prostate gland between treatments [3]. For many of these patients, their cancer will recur. In those cases, further treatment using a salvage therapy is necessary, which does not expose them to additional levels of radiation. HIFU is currently offered in hospitals as a minimally invasive salvage therapy for treating prostate cancer in patients whose cancer recurred after failed EBRT [2].

The efficacy and safety of salvage HIFU treatment after failed EBRT has been investigated in clinical studies. These studies reported good local cancer control but with some cases having high complication rates, comparable to other salvage therapies [2]. However, none of these studies consider the impact of implanted fiducial markers on the delivery of the HIFU treatment.

The objective of this work was to systematically investigate, using computational simulations, how the fiducial markers affect the delivery of HIFU treatment. The impact of the marker was studied through a series of large-scale simulations modelling the propagation of ultrasound pressure waves in the prostate with a single gold marker obstructing the beam's path. In each simulation a single spherical or cylindrical marker was included at different positions and orientations. For each marker configuration, a set of metrics (spatial-peak temporal-average intensity, focus shift, focal volume) was evaluated to quantify the distortion introduced at the focus in comparison to the corresponding metrics of a homogeneous simulation without a marker.

Results and Methods

The simulations were performed using the open-source k-Wave acoustic simulation toolbox developed by our group [4]. The toolbox solves a generalised version of the Westervelt equation which accounts for the combined effects of nonlinearity, heterogeneous material properties and acoustic absorption following a frequency power law. K-Wave is designed and optimised to fully utilise the computational resources offered by the IT4Innovations' Salomon and Anselm supercomputers on which the simulations were executed.

The simulated domain corresponded to a physical volume with dimensions 44.7x29.4x60.0 mm^3. The background medium was assigned the material properties (density and sound speed) of prostate tissue. For the heterogeneous simulations a spherical or cylindrical volume corresponding to the marker was also included and assigned the material properties of gold. All simulations were non-linear and accounted for absorption using a power law. The simulations were performed using a regular Cartesian mesh with a 1536x1024x2048 pt^3 grid-size. The large grid-size used, determined after a series of convergence tests, is necessary so that higher harmonics are supported in order to increase the accuracy with which acoustic non-linearity is captured. Such simulations have extreme computational and memory requirements. For example, a single heterogeneous simulation (with a marker) using 128 physical cores on Salomon requires approximately 332 GB of RAM and 8.5 days to complete, has a 226 GB input file and generates a 446 GB output. In total this study has consumed approximately 5 million core-hours.

The transducer model used in the simulations was derived from the transrectal probe of the Sonablate 500 (SonaCare Medical) clinical HIFU system used for treating prostate cancer. More specifically, the transducer geometry was assumed to be a single-element spherical cap with 22 mm width, 35 mm length and a fixed 40 mm focal length. A cross-section of the transducer model can be seen in Figure 1. The driving parameters of the transducer model were adjusted so that the spatial-peak temporal-average intensity at the focus of a homogeneous simulation (without a marker) is similar (1.1 kW/cm^2) to the values reported for the Sonablate 500 [5].

The distortion introduced by the fiducial markers to the HIFU beam was investigated by including a single spherical or cylindrical gold marker positioned at different coordinates in each simulation. The position of the transducer was kept fixed across all simulations. The spherical marker had a 3 mm diameter, whereas the cylindrical had a 3 mm height and 1 mm diameter. In total, 143 marker positions were simulated: 113 with a spherical marker and 10 with a cylindrical marker at 3 orientations. All the simulated positions are shown in Figure 1.

To quantify the effect of a single marker on the focusing of the HIFU beam, four metrics were evaluated using the simulation results for each marker position. These metrics were compared to the corresponding quantities obtained from a homogeneous simulation without a marker. In Figure 2 the effect of a single marker is shown in comparison to a homogeneous simulation.

The first quantity evaluated was the focal shift, calculated from the coordinates of the transducer's geometric focus and the coordinates of the maximum pressure point. The next two metrics were the spatial-peak temporal-average intensity evaluated at the geometric focus (I_{focus}) of the transducer and at the coordinates of the maximum-pressure point (I_{max}). These two quantities provide an indication of how much energy is redistributed due to the marker. The fourth metric evaluated the -6 dB focal volume for each simulation, which provides an indication of how the size of the focal region changes in comparison to the homogeneous simulation due to the presence of the marker.

The four metrics evaluated for each position of the spherical marker are shown in Figure 3. Each metric is plotted with respect to the marker distance from the transducer's focus with the distance increasing from the focus when the marker is positioned closer to the transducer. The metrics demonstrate that the marker acts as a strong reflector of the HIFU beam. When the marker is positioned in the pre-focal region, it causes reflections which induce a decrease in the focal intensity and focal volume, and a shift of the maximum pressure point away from the transducer's focus. These effects become more pronounced as its distance from the transducer's focus decreases, with the distortion introduced by the marker greatly increasing when placed within approximately 5 mm of the focus. The analogous simulations performed for the cylindrical marker also demonstrate that these effects depend on the shape and orientation of the marker.

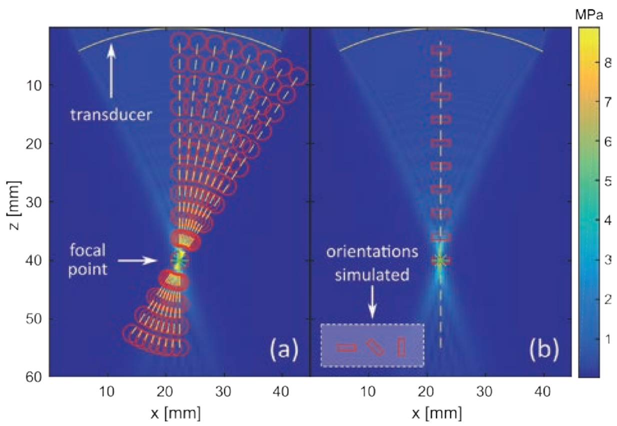

Figure 1.
The positions and orientations simulated for (a) the spherical and (b) the cylindrical markers. The background is the maximum pressure field from a homogeneous simulation.

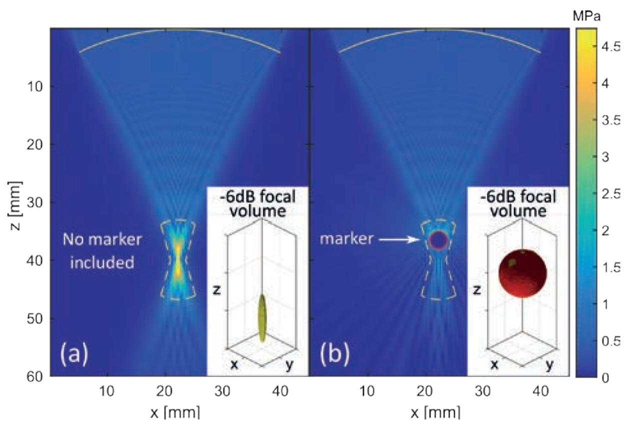

Figure 2.
Change in the maximum pressure field (b) due to the presence of a single spherical marker close to the focus compared to (a) the homogeneus simulation without a marker. The insets show the change in the -6 dB focal volume due to the marker.

Conclusion

A series of large-scale simulations was performed in order to study and quantify the impact of a single spherical or cylindrical gold fiducial marker on the HIFU beam. The four metrics evaluated for each marker position have shown that the distortion introduced by the marker increases as its distance from the transducer's focus decreases and depends on the marker's shape and orientation. The distortion when the marker is positioned within 5 mm of the focus significantly increases. This may result in an undertreated region beyond the marker due to less energy arriving at the focus, and an over-treated region due to reflections. Both effects may be undesirable depending on the location of the marker relative to the targeted cancerous region or other organs at risk.

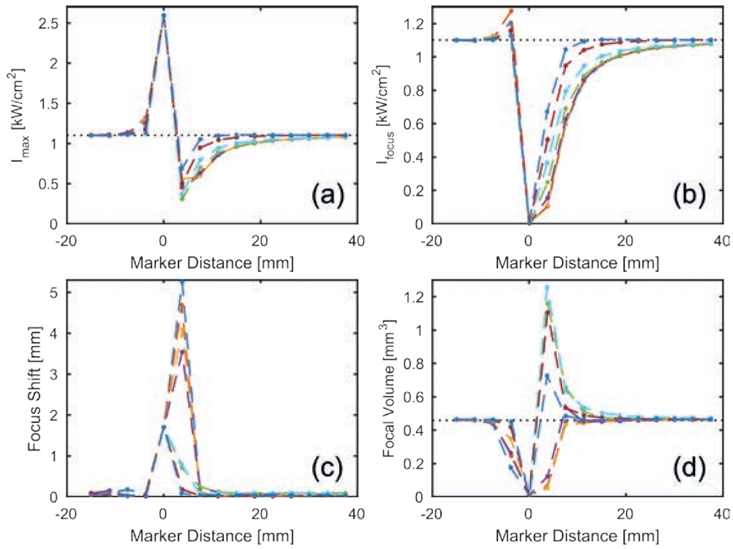

Figure 3.
The four metrics used to quantify the effect a fiducial is shown here for the spherical marker case.

On-going Research / Outlook

While investigating the impact of the markers, some additional factors have been omitted which may affect the distortion introduced by the marker to the treatment. Some of these factors are: the imaging element of the transducer, which was not included in the model, additional heating occurring due to absorption within the marker or viscous relative motion between the marker and surrounding tissue and the effect of multiple sonications during such treatments. In order to confirm the results observed from the acoustic simulations while taking into account the impact of many of the omitted factors above, our team is currently using experimental measurements on ex vivo tissue phantoms with implanted markers. The experimental work will also be extended to include measurements on ex vivo human prostate specimens.

It is also interesting to extend the results therein for salvage-HIFU treatment after failed (low-dose) brachytherapy. In this scenario, a large number of marker-like elements are introduced in the prostate. The material properties of these elements are similar to those of fiducial markers, and thus, expected to cause significant distortions to the HIFU beam.

References

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Publication

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Project website: bug.medphys.ucl.ac.uk