Phys. Med. Biol. **53** (2008) N437–N445 doi:10.1088/0031-9155/53/23/N03

NOTE

Evaluation of alignment error due to a speed artifact in stereotactic ultrasound image guidance

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Received 29 May 2008, in final form 13 October 2008
Published 12 November 2008
Online at stacks.iop.org/PMB/53/N437

Abstract

Ultrasound (US) image guidance systems used in radiotherapy are typically calibrated for soft tissue applications, thus introducing errors in depth-from-transducer representation when used in media with a different speed of sound propagation (e.g. fat). This error is commonly referred to as the speed artifact. In this study we utilized a standard US phantom to demonstrate the existence of the speed artifact when using a commercial US image guidance system to image through layers of simulated body fat, and we compared the results with calculated/predicted values. A general purpose US phantom (speed of sound (SOS) = 1540 m s^{-1}) was imaged on a multi-slice CT scanner at a 0.625 mm slice thickness and 0.5 mm × 0.5 mm axial pixel size. Target-simulating wires inside the phantom were contoured and later transferred to the US guidance system. Layers of various thickness (1–8 cm) of commercially manufactured fat-simulating material (SOS = 1435 m s^{-1}) were placed on top of the phantom to study the depth-related alignment error. In order to demonstrate that the speed artifact is not caused by adding additional layers on top of the phantom, we repeated these measurements in an identical setup using commercially manufactured tissue-simulating material (SOS = 1540 m s^{-1}) for the top layers. For the fat-simulating material used in this study, we observed the magnitude of the depth-related alignment errors resulting from the speed artifact to be 0.7 mm cm^{-1} of fat imaged through. The measured alignment errors caused by the speed artifact agreed with the calculated values within one standard deviation for all of the different thicknesses of fat-simulating material studied here. We demonstrated the depth-related alignment error due to the speed artifact when using US image guidance for radiation treatment alignment and note that the presence of fat causes the target to be aliased to a depth greater than it actually is. For typical US guidance systems in use today, this will lead to delivery of the high dose region at a position slightly posterior to the intended region for a supine patient. When possible, care should be taken to avoid imaging through a thick layer of fat for larger patients in US alignments or, if unavoidable, the spatial inaccuracies introduced by the artifact should be considered by the physician during the formulation of the treatment plan.

(Some figures in this article are in colour only in the electronic version)

1. Introduction

Stereotactic ultrasound (US) image guidance for alignment of prostate (Chandra *et al* 2003, Fuss *et al* 2003, Lattanzi *et al* 1999, Morr *et al* 2002, Serago *et al* 2006, 2002, Feigenberg *et al* 2007, Patel *et al* 2003, Poli *et al* 2007), as well as other sites such as pancreas and liver (Fuss *et al* 2004), is now well established as a viable image guidance tool for delivery of conformal radiotherapy. Common to all commercially available in-room US-guided alignment approaches are the acquisition of room-calibrated US images of the relevant target and critical structure anatomy which is to be aligned. This in-room-acquired US information defines the pre-treatment location of target and sensitive structures and is then compared with treatment-planning CT-contour representations of the same anatomy at the time of simulation. Deviations of the target and critical structure positions from their treatment-planned positions are subsequently quantified by user and/or software interaction, and corrective shifts are then performed to return the relevant anatomy to its position at the time of simulation. In this manner, the in-room US-guided approach facilitates the accurate delivery of conformal radiation therapy treatments.

Fundamental to the production of dynamic B-mode scanning images, the technique used most widely in modern US imaging, is the mapping of A-mode acoustic impedance variations into a two-dimensional image for display. The A-mode amplitude signal is mapped to a depth by the general relationship, depth $= t{*}C_{\mathrm{m}}$, where $2{*}t$ is the round trip echo time and C_{m} is the speed of sound in the media. It is clear from this relationship that an accurate understanding of the speed of sound in the media being imaged is essential to an accurate assignment of depth to a particular location in a B-mode US image.

Typically, clinical use of a stereotactic US guidance system for imaging of normal patient pelvic and abdominal anatomy entails the imaging of materials of variable cellular composition, with variable thickness layers of soft tissue, muscle and fat likely to be present. It is a well-appreciated fact of US imaging physics that an artifact, typically referred to as a speed artifact, is manifested due to the variability of sound propagation speed in different tissues, particularly fat (speed = 1450 m s^{-1}, versus 1540 m s^{-1} for soft tissue) (Anderson *et al* 2000, Napolitano *et al* 2006). The speed artifact can lead to edge discontinuities of distal organ borders after passing through a fatty tissue layer and, most relevant to this work, range and/or distance ambiguities. In short, US systems which are distance/depth calibrated for soft tissue applications will introduce errors in depth/distance-from-transducer representation when used in media with a different speed of sound propagation (e.g. fat).

Spatially calibrated systems, such as those used for stereotactic US image guidance, must assume a standard speed of sound for the media on which they will be used. Not surprisingly, this is typically selected as soft tissue, with a speed of sound of 1540 m s^{-1}. Standardized calibration phantoms are available for validation and calibration of depth accuracy and are typically filled with a water/alcohol solution formulated to have a sound propagation speed of 1540 m s^{-1}. Rigorous, phantom-based calibration processes allow the verification of the proper performance of an US-guided system for stereotactic alignment. The fundamental test employed for US guidance system calibration in our clinic employs a soft tissue-simulating phantom supplied by the US guidance system vendor.

In this study, we utilized a standard soft tissue-simulating US phantom to demonstrate the existence of the speed artifact when using a commercial US image guidance system to image through (simulated) layers of body fat. Furthermore, we quantify the magnitude of the depth-related alignment errors resulting from this artifact and compare the results with calculated/predicted values for clinically relevant situations in which stereotactic US image guidance might be used to align prostate and abdominal targets.

2. Methods and materials

2.1. US image guidance system

The specific stereotactic US guidance process used in our clinic employs the Nomos BAT system (Best–Nomos Inc., Cranberry Township, PA). Similar to other vendors of stereotactic US guided systems, the BAT system utilizes a standard, off the shelf, diagnostic US unit to generate the images necessary to the US guidance process. For the purposes of characterizing the US imaging system inherent phenomena of the speed artifact, the system should, therefore, serve as a valid representation of US guidance systems in general. A rigorous set of calibration processes are utilized in our clinic, as recommended by the vendor, to assure the spatial integrity of the US alignment process. One of these tests makes use of a prostate/soft tissue-simulating phantom to test the spatial calibration of the system. The calibration process entails, essentially, the use of the phantom as a controlled surrogate for the patient.

The BAT system utilizes either a stereotactic arm or a reflective array of passive infrared markers to convey the position and orientation of the curvilinear phased array transducer to the BAT software. Armed with room-coordinate information regarding the plane from which the US image is generated, the software is able to overlay the isocenter-relative, CT-derived contours of target and critical structures onto a real-time, room-coordinate US image. By software manipulation of the CT-derived contours on the in-room computer screen, the target and critical structure definitions from both CT simulation and US images can be made to agree. The software subsequently tracks the X, Y, Z coordinate shifts required to cause the two image sets to agree, thus enabling the user to perform the shifts necessary to properly return the target to its simulated position, relative to isocenter.

2.2. Experimental setup and procedure

Figure 1 depicts the experiment setup of an US phantom and BAT SXI system with two layers (3 cm and 2 cm) of the fat-simulating material stacked on top of the phantom. The General Purpose Multi-Tissue Ultrasound Phantom Model ATL 040 with a speed of sound (SOS) = 1540 m s^{-1} (CIRS, Norfolk, VA) was imaged on a LightSpeed RT CT scanner (GE Health Care, Waukesha, WI) at a 0.625 mm slice thickness and 0.5 mm × 0.5 mm axial pixel size. Target-simulating wires inside the phantom were contoured and the contours were transferred to the in-room BAT SXI unit, as would happen for a patient who was to be treated with US guidance. To simulate different patients with differing thickness of overlying abdominal fat, layers of various thicknesses (1–8 cm, in 1 cm increments) of fat-simulating material (SOS = 1435 m s^{-1}), manufactured by the Ultrasound Division of CIRS, were placed on top of the phantom. A thin layer of optical coupling gel was applied to all interfaces of the phantom, fat-simulating material and the US probe to facilitate the acquisition of optimal

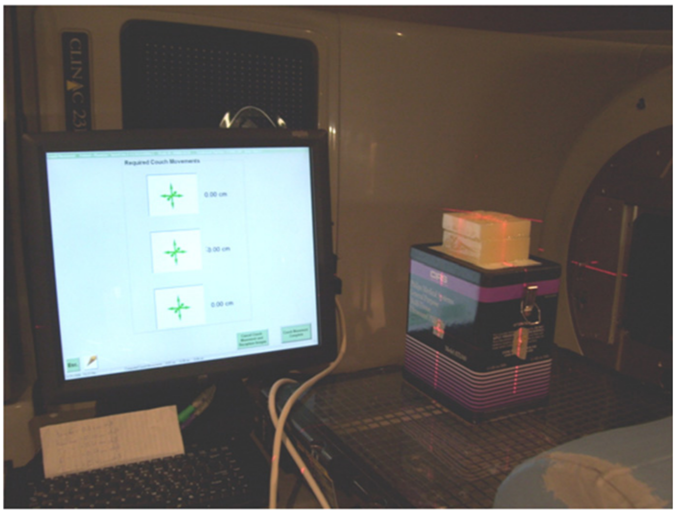

Figure 1. Experimental setup of the US phantom and BAT SXI system. Two layers (3 cm and 2 cm) of fat-simulating material are shown stacked on top of the US phantom.

quality US images of the phantom. In order to make clear that the speed artifact observed here is not caused by the experimental addition of various layers of material on top of the phantom, we repeated these measurements using layers of identical thicknesses of the tissue-simulating material (SOS = 1540 m s^{-1}), which were also manufactured by the Ultrasound Division of CIRS. Technically, this step is not necessary because the software 'understands' the isocenter-relative, room-coordinate position of the US transducer within the room, so the addition of extra layers/depth on top of the phantom is accounted for by the software in its spatial representation of the image. Nevertheless, we chose to precisely replicate the experiment with both fat and tissue to eliminate potential concern that the results were affected by the addition of fat layers alone. Data collection entailed performing 12 independent measurements using four different ultrasound operators, for a total of 96 independent measurements.

2.3. Predicted alignment errors by calculation

Given the linear mapping of echo time to depth, we expect the misrepresentation of depth (or alignment error) to be given by equation (1):

$$\delta = (C_{\text{tissue}}/C_{\text{fat}} - 1)^{*}d \qquad (1)$$

where δ is the misrepresentation of depth caused by the speed artifact; C_{tissue}, C_{fat} are the US speed in tissue and fat, respectively; and d is the thickness of the fat-simulating material.

3. Results and discussion

3.1. System calibration

Figure 2 shows the result of alignment of the phantom's target-simulating wires, without added fat-simulating layers, immediately following the calibration of the system. As would be expected, the CT-derived contours align perfectly with the US image of the structures. This merely demonstrates that the system was properly calibrated, as shown by solid arrows in

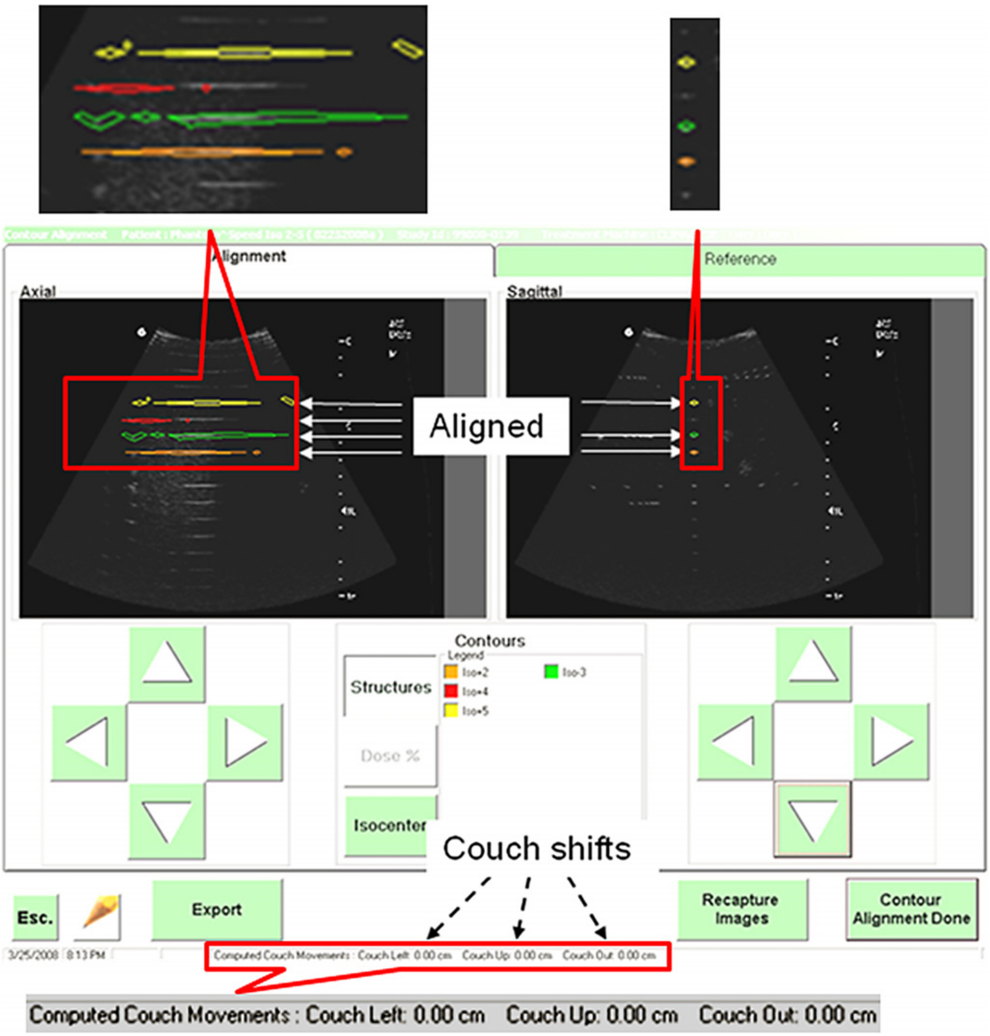

Figure 2. The alignment of the phantom target-simulating wires without any fat- or tissue-simulating material. The zero couch shifts indicated a perfect system calibration.

the figure. The fact that this alignment was achieved without manipulating the contours in software, in the usual way that an improperly aligned patient would be returned to the correct position, can be seen in the screen-captured image of the BAT software by observing that the displayed result is for the POST alignment values and that couch movements required to achieve this alignment were couch left 0.00 cm, couch up 0.00 cm, couch in 0.00 cm, as indicated by dashed arrows in the figure.

3.2. Fat versus tissue layers

Figures 3(a) and (b) depict the results of the alignment performed for the phantom with 3 cm layers of either fat- or tissue-simulating material on top, prior to software manipulation of the CT-derived contours. Enlargements of the solid box area are shown for a better appreciation

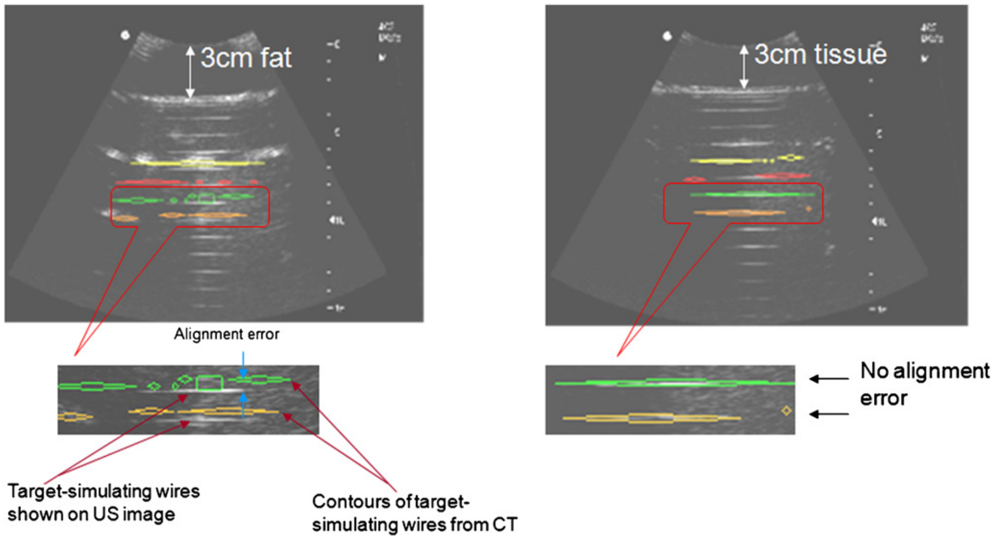

Figure 3. US axial images of the phantom with either 3 cm fat- (a) or tissue-simulating material (b) on top. No software manipulation of the contour alignment has been performed. (a) The US image of the target wires and the contours are not correctly aligned for the case of fat-simulating material and would require a 2.5 mm vertical shift to correct. (b) In contrast, the US image of the target wires and the contours are perfectly aligned for the case of tissue-simulating material.

of the misalignment of the CT-derived contours with the target wires. Figure 3(a) shows that the US image and the contours were not correctly aligned for the case of fat-simulating material. The POST aligned contours (i.e. the contour alignment achieved following software manipulation of the CT-derived contours) require a 2.5 mm vertical couch up shift for the CT-derived contours to align with the US image of the targets. Such a shift demonstrates that the US image of the target has been incorrectly represented as being 2.5 mm deeper than it actually was or, in other words, that the phantom/target would need to be raised by 2.5 mm to cause the US image of the target and the CT-derived contour to agree. In contrast, the US image of the target wires and the target contours were perfectly aligned, without any need for corrective shifts, for the case of the 3 cm thick layer of tissue-simulating material (figure 3(b)). Figure 4 depicts an example of a screen snapshot of BAT, post-alignment, for the phantom with 8 cm fat material on top. Both axial and sagittal US images of the target-simulating wires were correctly aligned with the contours in the depth direction (A/P) after a 5.7 mm couch up shift.

3.3. Alignment error magnitude and direction

Figure 5 depicts a comparison between measurements and calculations of the alignment errors caused by the speed artifact for seven different thicknesses of fat-simulating material. The measurements for 1, 2 and 3 cm were carried out with a single piece of 8 cm × 8 cm × (1 or 2 or 3 cm) fat-simulating material lying flat, respectively, while the measurements for 4 and 5 cm were done with a combination of different pieces (i.e. (1 + 3), (2 + 3), respectively). The 6 cm measurement was achieved by using the 8 cm × 6 cm × 3 cm thick piece positioned on its 6 cm side or edge and the 8 cm measurement was achieved by using the 8 cm × 6 cm × 3 cm thick piece positioned on its 8 cm side. The 7 cm measurement was achieved using the 8 cm × 6 cm × 3 cm thick piece on its 6 cm edge combined with a 1 cm slab. As shown in

Figure 4. A screen snapshot of BAT, post-contour alignment, for the phantom with 8 cm fat material on top. Both axial and sagittal US images of the target-simulating wires were correctly aligned with the contours in the depth direction (A/P) after a 5.7 mm couch up shift.

figure 5, the mean measured value for each fat thickness agreed with calculations for each thickness within one standard deviation for all points, or a maximum deviation of 0.47 mm. We note that the measurements for 4, 5 and 6 cm deviated slightly more from the calculated values than the other measurements. We attribute this to the fact that the image quality was degraded when passing through the interfaces between the multiple fat-simulating layers, thus causing more uncertainty in the measurement. This is further supported by the fact that the deviation for 6 cm is slightly larger than those for 4 or 5 cm because of two interfaces rather than one.

For the fat-simulating material ($C_{\text{fat}} = 1435$ m s^{-1}) used in this study, we observed the magnitude of the depth-related alignment errors resulting from the speed artifact to be 0.7 mm cm^{-1} of fat imaged through. As for the direction of the alignment error, the lower speed of sound in the fat material aliased the position of the target-simulating wires in the US image deeper, or further from the transducer, than they actually were. This is due to the fact that the reduced speed of sound in the fat layer causes the round trip time to be greater than when passing through soft tissue, and thus, causes the system to interpret the echo as having come from deeper in the phantom. This direction is confirmed by the fact that all of

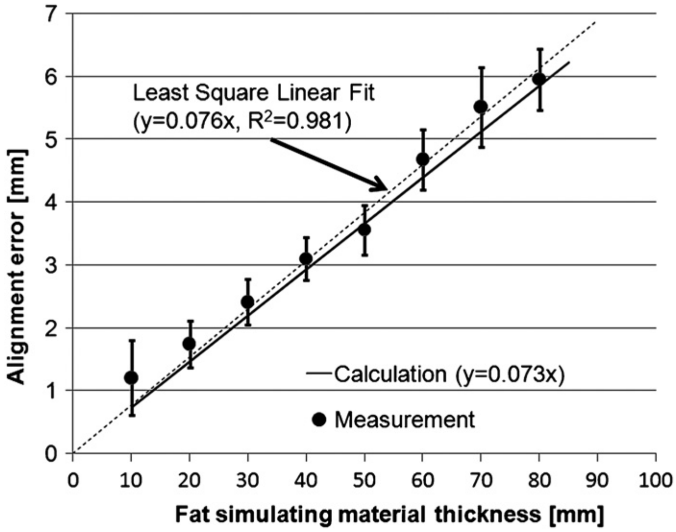

Figure 5. Comparison of measured alignment errors (caused by speed artifact) with the calculated values for eight different thicknesses of the fat-simulating material. The measurements and the calculations one agree within standard deviation or a maximum deviation of 0.47 mm.

the BAT alignments for fat required a couch up shift. The systematic introduction of a small, but erroneous, upward shift of the couch will subsequently cause the high dose region to be delivered slightly posteriorly to its intended position. The speed artifact error demonstrated here as a systematic error, which is a function of the amount of fat imaged through, would also be expected to be present (to varying degrees) in the data published by previous authors characterizing clinical comparisons between US guidance and other, 'gold standard' alignment methods.

4. Conclusions

We have demonstrated the depth-related alignment error due to a speed artifact when using US image guidance to image through clinically relevant thicknesses of fat, for radiation treatment alignment. For the commercially manufactured fat-simulating material ($C_{\mathrm{fat}} = 1435$ m s^{-1}) used in this study, we observed the magnitude of the depth-related alignment errors to be 0.7 mm cm^{-1} of fat imaged through. As expected, the lower speed of sound in the fat material aliased the position of the target-simulating wires in the US image deeper, or further from the transducer, than they actually were. This aliasing can subsequently cause the high dose region to be delivered slightly posteriorly to its intended position. We also calculated the expected alignment errors based on the ratio of speed of sound in fat versus tissue and saw the calculated values to agree with the measurements within uncertainties. Our study suggests that care should be taken to avoid imaging through a thick layer of fat for larger patients in US alignments whenever possible. If this is unavoidable, the inaccuracy in alignment which is introduced by the speed artifact should be considered by the physician as it relates to the dose actually delivered to target and critical structures or might possibly be corrected by CT measurement of the fat thickness traversed with subsequent correction of the US image guided shift. This is a topic that we hope to explore further in future work.

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