Review of Imaging for Brachytherapy

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Introduction

Almost every form of imaging can be used in brachytherapy, and many clinics may employ different imaging for similar tasks. Most of these uses are similar to how these modalities would be used in external beam or in other functions within the radiation oncology clinic. An example of this would be that CTs are acquired similarly for treatment planning of high dose rate (HDR) brachytherapy as they would be for planning of an external beam treatment. Two topics which will not be covered in other talks in this session and have a much more brachy-specific niche in how they are used will be discussed here: brachytherapy specific issues in MR, and ultrasound.

Magnetic Resonance Imaging

MR can be very useful in brachytherapy treatment planning due to the fact that many current brachytherapy treatment planning algorithms are TG-43 based and assume a homogeneous density of water within the body for dose calculation. As with other uses in the clinic, T1 and T2 weighted MR imaging can be useful for greater soft tissue contrast visualization compared to CT. Other MR sequences and techniques may have uses which are more specialized for brachytherapy issues. Two examples presented here are proton density weighted imaging and diffusion weighted imaging:

Proton Density Weighted MR:

- Proton density weighted images produced with spin echo sequences using long relaxation times (TR=2000 - 4000 ms) to minimize differences in signal between tissues due to differences in T1 relaxation and short echo times (TE = 3 – 30 ms) to minimize differences in signal due to T2 differences in tissues.
- Larger proton density tissues have higher signal.
- Proton density images have very higher signal and contrast to noise ratio (CNR) compared to T1, T2 images.
- Proton density MR has less soft tissue contrast in general than T1 or T2 weighted imaging.
- However contrast is much greater between tissues and MR-compatible applicators (made from materials such as titanium) which have low contrast relative to the surrounding tissues in T1 or T2 images.
- Proton density weighted images can be used in gynecological brachytherapy planning for visualization and digitization of applicators; these images can be registered with the T1 or T2 images used for soft tissue visualization and treatment planning.

Diffusion Weighted Imaging (DWI):

- DWI measures differences in intensity and directionality of the random motion of water molecules in various tissues.

- DWI sequences employ very strong MR gradients applied symmetrically around a refocusing gradient in a spin echo or other sequence.
- Water diffusion properties are quantified on DWI through the use of apparent diffusion coefficient (ADC) maps.
- ADC maps are determined from two or more gradients through relationships such as:

$$S_{DW} = S_o e^{-b^* ADC}$$

  Where SDW is the S_{DW} and S_0 are signal intensities measured with and without diffusion-weighted gradients, respectively, and b is the diffusion factor (in s/mm^2) that characterizes the strength of the diffusion gradients.
- Tissues with restricted diffusion appear darker on DWI images.
- It has been shown in the literature that cervix tumors are more cellularly dense than healthy tissues, restricting diffusion, and resulting in lower ADC values (appearing darker in DWI) compared to healthy tissue.
- Registration of DWI with T2 weighted images can aid in the delineation of targets for brachytherapy treatment planning.

Ultrasound

In the past, ultrasound imaging had more broad uses in radiation oncology (example: used for daily setup for prostate patients receiving external beam), but has lost favor in a number of applications with the implementation of other imaging technology in the clinic. For brachytherapy treatments such as prostate LDR seed or HDR needle implants, gynecological implants, and episcleral eye plaque brachytherapy, however, ultrasound remains a widely used imaging modality for a preponderance of clinics for treatment planning and/or image guidance and it is thus important to understand the fundamentals of ultrasound imaging and production.

Basics

- Ultrasound is sound waves > 20 kHz. Medical ultrasound is typically in the 2-20 MHz range.
- In ultrasound imaging, emitted sounds waves are reflected at boundaries between material with differences in acoustic impedance. Assuming a known speed of sound (for water or tissue), the time it takes for an emitted wave to be reflected back can determine how deep an object or boundary is. This is combined with information regarding the intensity of the reflected wave to generate images.
- Scattering from objects or surfaces with that have variation the size of the sound wavelength are smaller are "non-specular reflectors" and reflect in all different directions. The varying non-specular structure of different tissues gives rise to changes in texture and brightness between different tissues in ultrasound images.
- Tissue / air interfaces are highly reflective, meaning much of the wave cannot penetrate into the air cavity.
- Bone is highly attenuating, meaning ultrasound typically cannot transmit through bone.

- Higher frequency ultrasound waves give images with greater axial (depth) resolution as they can discriminate between more closely spaced reflectors but are attenuated more quickly (0.5 dB/cm/MHz for soft tissue).
- Lower frequency waves are more penetrating, allowing for imaging to greater depths.
- For prostate brachytherapy, 6-10 MHz can be used while for breast or eye a higher resolution, less penetrating beam may be more useful (10-18 MHz).

Ultrasound Beams

- Ultrasound transducers produce and receive sound waves using piezoelectric ceramic or crystals which change shape with applied electrical potentials, converting digital signals into a mechanical pressure waves and vice versa.
- Damping blocks coupled to the transducers give the ultrasound beam a pulse length.
- Typical transducers are arrays 128 or more rectangular piezoelectric elements.
- Linear arrays use small groups of transducers at a time to determine image information along those elements.
- Phased arrays steer the focal area by using all elements together with time delays between when they emit the ultrasound waves.
- The ultrasound beam converges for a depth (the "near field" or "Fresnel zone") and then begins to diverge after a certain depth (the "far field" or "Fraunhofer zone"). This leads to a focus spot at a depth of greatest convergence where the resolution is the best.
- Additional focus and broader focal zones can be generated by adding and modulating time delays to the listening mode of the transducers.
- Axial (depth) resolution is dependent on the pulse length (resolution = 1/2 pulse length).
- Lateral resolution dependent on depth and transducer width. Best lateral resolution occurs at focal spot/zone.
- Elevational resolution (slice-thickness) dependent on transducer height.

Image Formation and Artifacts

- A-(Amplitude) mode shows processed information from the receiver versus time along a certain line of imaging which shows the relative depth and intensity of the signal. A-mode or A-line imaging is used in measuring structure and tumor depths in the eye for plaque brachytherapy planning.
- B-(Brightness) mode changes the information along a line from A-mode into grey-scale or colored dots, where the color of the dots is preportional to the intensity of the signal at those points.
- 2D ultrasound images are created from combining B-mode image lines acquired throughout a space. Most standard imaging applications (for brachytherapy and other applications) are created this way.
- M-(Motion) mode displays B-mode dots from one specific line moving with time. M-mode is of lesser importance today.
- Most modern ultrasound devices used phased arrays to create 2D images where time delays steer the beam at different angles from the transducer face. This allows the device to sweep the beam through the imaging volume without moving.
- The rapid rate of at which the sweeping of both transmitting and receiving can be done allows ultrasound to be essentially a "real time" imaging modality.

- "Reverberation" or "ringing" artifact occurs when a highly reflective object is in the beam path (such as a needle containing LDR seeds for prostate treatment), characterized by multiple equally spaced bright objects visible along the line of the object.
- Highly or lowly attenuating objects can cause artificial shadowing or enhancement of structures.
- Refraction of the beam at an oblique angle from boundaries not perpendicular to the beam can cause misplacement or distortion of objects.

Ultrasound Devices and QA

- Single arrays in a linear formation or curvilinear formation lead to rectangular or trapezoidal fields of view, respectively and are used in brachytherapy applications such as breast and pelvis (for guiding implant placement) and in the eye.
- QA for such arrays is outlined in AAPM Ultrasound TG-1.
- Prostate LDR/HDR implants and some gynecological applications are done using endorectal transducers which generally have two perpendicular arrays for performing both axial and sagittal imaging of the patient with the same transducer.
- Some devices have one crystal which translates back and forth between axial and sagittal planes.
- QA for these endorectal devices is outlined in AAPM TG-128.

References / Further Information

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