Imaging Techniques to Aid IR Treatment of Musculoskeletal Malignancy

A. Kyle Jones, PhD¹ Steven Yevich, MD, MPH²

¹ Division of Diagnostic Imaging, Department of Imaging Physics, The University of Texas MD Anderson Cancer Center, Houston, Texas

² Division of Diagnostic Imaging, Department of Interventional Radiology, The University of Texas, MD Anderson Cancer Center, Houston, Texas

Address for correspondence A. Kyle Jones, PhD, MD Anderson Cancer Center, 1400 Pressler St, FCT14.5026, Houston, TX 77030 (e-mail: Kyle.Jones@mdanderson.org).

Semin Intervent Radiol 2019;36:49–62

The minimally invasive image-guided approach to musculoskeletal (MSK) malignancy requires mastery of multiple different imaging modalities. The therapeutic approach can include embolization, thermal ablation, cement consolidation, and percutaneous screw fixation.¹ While some treatments require one specific modality and approach, others might be best performed using multiple modalities or techniques in concert.^{2–5} The challenges to the approach are often due to the wide spectrum of disease presentation related to variable tumor biology, location, size, and vascularity. A firm familiarity with the latest advancements in imaging equipment and software can improve the patient-tailored approach.

Several recent advances in medical imaging can improve minimally invasive treatment options for MSK malignancy. For example, innovation in hardware and software applications has improved the quality of anatomical detail to facilitate visualization of tumor and surrounding critical structures. Needle guidance and fusion capabilities have expanded the treatment potential by expanding the potential combined applications of multiple imaging modalities. The proper application of these technological improvements requires a basic understanding of specific imaging parameters and the underlying medical imaging physics.

This article will review the latest advancements and applications for ultrasound, fluoroscopy, computed tomography (CT), and magnetic resonance imaging (MRI) as applied to the treatment of MSK malignancy. Basic technical information will be reviewed, as will imaging optimization techniques to improve procedural outcomes and safety for both the patient and proceduralist. Lastly, new software technologies and future directions will be presented for each imaging modality that may assist in treatment approach or assessment of immediate procedural effect.

Ultrasound

Ultrasound is an imaging modality that is well suited for MSK procedures owing to the high frame rates (10–70 frames per second [fps]) that allow real-time guidance. While a limited penetration depth prevents visualization of deep structures, superficial non-ossified masses located less than 10 cm deep to the skin are typically well visualized owing to excellent sonographic contrast between fat, muscle, and connective tissues. Doppler and color flow modes improve visualization of vascular structures, which can assist in either intentional puncture or avoidance of these structures. During ablation procedures, ultrasound monitoring may be helpful to assess the superficial ablation margin during radiofrequency and microwave ablation by tracking microbubble formation, and during cryoablation by tracking the hyperechoic superficial border of the therapeutic ice ball.

Ultrasound probes are available in a variety of configurations including linear, curvilinear, sector, and two-dimensional (2D) matrix transducers. Linear arrays are typically higher frequency (e.g., 5–12 MHz), which limits depth of penetration but provides superior resolution for superficial structures. Curvilinear arrays are typically lower frequency (e.g., 1–5 MHz), offering increased depth of penetration and a larger field of view (FOV), but lower resolution. Sector arrays offer a relatively large FOV with a comparatively small footprint that can be used in narrow acoustic windows, for example, in intercostal spaces.

Many modern ultrasound transducers are 1.5 dimensional, containing multiple rows of elements. This allows steering and focusing of the ultrasound beam in the elevational (slice thickness) direction, improving elevational resolution and reducing partial volume effects. Two dimensional transducers are a recent advancement that use many more rows of elements to provide more reliable operation with a smaller footprint. These new 2D transducers also offer three-

dimensional (3D) imaging, otherwise known as volumetric or multiplanar imaging, by beam steering in all directions. This real-time volumetric imaging can reduce or eliminate acoustic shadowing caused by intervening bony structures, such as ribs.

Several basic settings can be optimized on the ultrasound equipment to improve visualization of MSK structures. Modern ultrasound systems are delivered with numerous imaging presets for specific applications, and some systems offer “one click” image optimization. Once the appropriate MSK preset has been selected, common settings to improve tissue contrast include acoustic output, gain, and focal depth setting. Gain changes should be dialed to optimize tissue contrast of the target lesion. Reduced dynamic range offers higher contrast for superficial imaging, while increased dynamic range allows visualization of deeper structures. With broadband probes, higher frequencies (also known as “resolution” setting) should be selected for superficial imaging to maximize resolution, while lower frequencies (also known as “penetration” setting) are needed to image deeper structures. The focal depth setting affects the frame rate to optimize image quality at a specified depth. Alterations in line rate (also known as “resolution” or “speed” setting) will alter image characteristics by its impact on both frame rate and lateral resolution. Spatial compounding and other image processing techniques are routinely used to optimize B mode ultrasound imaging. Finally, beam steering can be useful for maximizing needle visibility during MSK interventions.

Several new ultrasound applications may be applied to different probes to specifically improve guidance during MSK procedures. Tissue harmonic imaging (THI), shear wave elastography (SWE), and contrast-enhanced ultrasound (CEUS) are the most recent advancements. While THI has definite applications to MSK interventions, the use of SWE and CEUS for MSK intervention is still under investigation.

THI is an established technique that can improve contrast and reduce artifacts for specific imaging applications. THI is available on most ultrasound probes, and is accomplished by

tuning the transducer to receive *harmonics* or multiples of the transmitted fundamental frequency. Typically the second harmonic is processed, while echoes returned at the fundamental frequency are cancelled. Harmonic generation is proportional to the intensity of the ultrasound beam at the fundamental frequency, leading to reduced appearance of artifacts from side and grating lobes and improved lateral and elevational resolution. THI may be particularly beneficial for deep locations or in patients with large body habitus due to improve signal-to-noise ratio and contrast for deep tissues. Conversely, imaging of lesions in shallow locations may be optimized by removing harmonics and increasing transmit frequency (→Fig. 1). The relative strength of the harmonic signal increases with depth as the nonlinear distortion of the transmitted ultrasound waveform increases with depth and contamination from shallow echoes is reduced.

CEUS has mainly been studied for in cardiac imaging and for imaging of visceral organs,^{6,7} but may offer the potential to differentiate MSK malignancy from surrounding normal tissues by imaging differences in vascularity.⁸ Ultrasound contrast agents are nephron-sparing alternatives to CT or MRI contrast agents, as the contained gas is eliminated via the lungs and therefore can be used in patients with renal impairment. The ultrasound contrast agents are composed of gas-filled microbubbles surrounded by a flexible shell that are injected in volume through a peripheral intravenous line and are imaged as they pass through in the capillary beds. Special ultrasound imaging modes use low intensity acoustic energy to detect the greater tendency of the microbubbles to scatter sound waves and affect harmonic resonance. In addition, imaging modes can cause timed destruction of the microbubbles by focused application of higher-intensity acoustic energy, which allowed real-time visualization of vascular differences as the capillaries refill with microbubbles. These special CEUS modes are typically configured to use frame rates sufficient for collecting dynamic information and low acoustic output to maintain a low mechanical index (MI) to prevent premature destruction of the microbubble

Fig. 1 Metastatic melanoma to the right subpectineal soft tissues. Image optimization for visualizing shallow target. Initial settings using low transmit frequency and tissue harmonic imaging suffers from lower resolution and poor needle visibility (a). Disabling tissue harmonic imaging and using higher transmit frequency offer better resolution and improved needle visibility (b).

contrast agent and to minimize harmonic echoes from surrounding tissue. When correctly used, the improved contrast resolution may assist in needle targeting of viable tumor.⁹ Alternatively, vascular changes may be assessed after thermal ablation to evaluate treatment margins.

SWE is a newer technique that applies an intense, focused acoustic pulse from the ultrasound transducer to microscopically displace tissue. The rebounding of the target tissue generates shear waves, which propagate perpendicular to the incident sound wave. The speed of the propagating shear wave is sensed by the transducer and reported as either shear wave speed or tissue elasticity. Different implementations of SWE are available, although all applications are limited to superficial targets within a few centimeters of the dermal surface. The shear wave speed may be reported for a single region of interest (ROI) or as a color map on which multiple ROI measurements can be made. Although a recent study found that SWE added little value for the diagnostic evaluation of soft-tissue masses,¹⁰ this technology may prove to provide some intraprocedural information to discern necrotic tissue from viable tumor during biopsy and ablation.

Advancements in ultrasound guidance software provide improved visualization of the percutaneously advanced needle, project an accurate needle trajectory to safely guide the proceduralist, or register (fuse) the ultrasound image with cross-sectional imaging for enhanced real-time guidance. Simple needle guidance modes allow the operator to plot a track from puncture site to target, which can then be used to manually guide the needle. Other modes allow the real-time display of the needle tip and trajectory on the ultrasound image, allowing the needle to be tracked using an electromagnetic field generator placed near the patient. Sensors are incorporated into the navigation systems, and may be internal or external to both the needle and the ultrasound transducer. The most complex needle guidance modes register previously acquired cross-sectional image data from CT or MRI with the live ultrasound image to allow real-time evaluation of the cross-sectional image in the plane in which the ultrasound probe is positioned. This facilitates targeting of structures better visualized on CT or MRI, while allowing real-time and out-of-plane guidance by the ultrasound. Such systems require a DICOM connection to the CT or MRI system or the hospital PACS for image transfer to the ultrasound. While the majority of recent literature regarding ultrasound needle guidance is in relation to liver ablation,¹¹ the applicability to MSK malignancy is pertinent.

Fluoroscopy and Cone Beam CT

Fluoroscopy is a versatile imaging tool that is commonly applied in the treatment of MSK malignancy. Fluoroscopy provides excellent spatiotemporal information to allow embolization of vascular tumors and facilitate consolidation of pathologic fractures by vertebral augmentation, cementoplasty, or percutaneous screw fixation. Modern fluoroscopy systems offer numerous features that are designed to optimize image quality and patient dose.

All modern fluoroscopy systems incorporate flat panel image receptors. Flat panel detectors (FPDs) are less bulky than X-ray image intensifiers (XRII) and offer better patient access. More importantly, FPDs are less susceptible to image distortions, opening the door to advanced applications such as cone beam computed tomography (CBCT). Most modern FPDs use a cesium iodide conversion layer atop an amorphous silicon transistor array, and such FPDs are available in large formats (e.g., 40 × 30 cm). Recently, FPD based on crystalline silicon have become available in small cardiac equipment, but will soon be available for standard interventional radiology equipment. Lower electronic noise levels and therefore reduced radiation doses.

Optimization of fluoroscopic image acquisition depends on the procedural application. Modern fluoroscopic systems have complex algorithms for adjusting radiation output, radiation quality, and image processing based on imaging preset selection and multiple feedback sources. Fluoroscopic systems are beginning to incorporate task-specific measures of image quality into their feedback logic instead of only using detector signal output as feedback to the X-ray generator. The extent to which the end user has access to the various X-ray and image parameters of a fluoroscopic system depends on the system manufacturer and configuration, and in some cases may require service-level access.

Cone beam CT (CBCT), also known as flat panel CT (FPCT), is a volumetric imaging mode available on many modern angiography systems. In general, CBCT is considered a subset of CT. During a CBCT acquisition, the C-arm rotates rapidly around the patient, covering approximately 200 degrees, while acquiring hundreds of projection images at frame rates exceeding 60 fps. Geometric and X-ray intensity corrections stored in the system are applied to each projection, and filtered back projection algorithms are used to reconstruct multiplanar images from the acquired projection data. Modern CBCT protocols acquire 200 to 400 frames in 3 to 6 seconds. CBCT systems require precise geometric calibration to produce artifact free images. For this reason, CBCT is typically not available on C-arms that use XRII. Future developments in CBCT, including the use of crystalline silicon detectors, will likely result in faster acquisition times with increased view sampling, within gantry speed constraints imposed by mechanical and safety considerations.

Fluoroscopy/CBCT Optimization for Embolization

For embolization procedures, modern fluoroscopy systems use complex, nonstationary spatiotemporal image processing to enhance the appearance of contrast agents, thin guidewires, and low-profile catheters. The configuration settings for these algorithms are not typically accessible to the end user; however, some control may be afforded through the ability to select from a set of preconfigured levels. These settings can strongly affect the appearance of image noise, reduce contrast, and introduce *artificial lag*. While the specifics of image processing are inherent to the

manufacturer, several fluoroscopy settings may be adjusted by the operator.

On all fluoroscopic systems, the operator has the ability to choose from different imaging presets, which may be configured for varied target detector radiation doses, pulse rates, filtration, and image processing settings. Pulsed fluoroscopy is the standard mode of operation, with available pulse rates typically ranging from 0.5 to 30 pulses per second (pps). On most fluoroscopy systems, the operator can override the fluoroscopic pulse rate as needed to deliver the temporal resolution required for the imaging task. Higher pulse rates result in smoother reproduction of motion but also higher radiation doses to the patient and operator, and moderate pulse rates in the neighborhood of 7.5 pps are adequate for most imaging tasks. Another important parameter is the X-ray pulse width, which is set according to the configuration of the imaging preset, and may be affected by other settings, including kV, added filtration, focal spot, and others. Shorter X-ray pulse widths result in sharper images with less motion blurring; however, short pulse widths may involve other tradeoffs such as higher kV or a larger focal spot.

Most fluoroscopic systems offer multiple magnifications modes or FOV. The relationship between detector dose rate, resolution, and FOV is straightforward for fluoroscopes using XR11. The detector dose rate is inversely proportional to the area of the FOV, while the resolution increases linearly as the FOV decreases. This relationship is less straightforward for fluoroscopes with FPD, with most manufacturers increasing the detector dose rate linearly with decreases in FOV. Changes in resolution are more complicated, as detector pixels are often binned at large FOV to facilitate rapid readout for real-time imaging, while smaller FOV uses the native resolution of the FPD. As the FOV is decreased, detector pixel matrices of

different sizes and resolution are interpolated for display on the same size monitor, similar to digital zoom on a camera. Finally, some manufacturers also apply increasing levels of edge enhancement as the FOV is reduced. The result is a nonlinear change in spatial resolution as the FOV is decreased.

Several adjunct imaging modes are available on fluoroscopic systems, including digital acquisition imaging, digital subtraction angiography (DSA), roadmapping, and fluoroscopy reference overlay.¹² Digital acquisition imaging uses high-dose rates to produce a series of radiographic-quality images for diagnosis, while DSA typically uses even higher dose rates to produce background-subtracted images of only contrast-filled vasculature. DSA images are ideally free of anatomical background, and remasking or pixel shifting techniques can be used to correct for motion artifact. The signal in contrast-filled vessels is linearly related to the amount of contrast in the vessel. A single frame of DSA, or temporal maximum (or minimum) intensity projection, also referred to as “peak opacification” or “image stacking,” can be used as a reference image, which can also be inverted and overlaid on the real-time fluoroscopy image to provide a map of arterial anatomy for navigation (►Fig. 2). Roadmap, also known as subtracted fluoroscopy, is an appealing alternative as a lower dose option to create a fluoroscopic map of vascular anatomy. Similar to DSA, a fluoroscopic mask image is subtracted from subsequent fluoroscopy fill images, which are acquired as contrast is injected through a catheter. A peak or maximum opacification image is created, inverted, and overlaid on the live fluoroscopy image to provide a map of vascular anatomy. In stationary anatomy, roadmap is preferred over DSA due to decreased radiation doses; however, acquiring a roadmap of sufficient quality may be difficult in patients with large body habitus.

Fig. 2 Metastatic renal cell carcinoma to the right femoral diaphysis status postresection and surgical stabilization with metallic plate-screw construct. Comparison of reference overlay created from single best DSA frame (a) versus reference overlay created from peak opacification (“image stacking”) processing (b).

Image postprocessing techniques can also be used to optimize fluoroscopic images. Common techniques include edge enhancement, harmonization, and noise reduction. A variety of techniques, including unsharp masking, can be used to enhance edges in fluoroscopic or DSA images. General edge enhancement techniques also increase image noise, and modern algorithms typically use multifrequency processing or edge detection to improve detail with less impact on image noise. Harmonization is used for large-scale contrast optimization, and noise reduction algorithms can be generic, reducing noise but also detail, or adaptive (edge preserving), using advanced processing to reduce noise while preserving edge detail.

While DSA imaging produces images of higher quality with more contrast and less noise than fluoroscopy, the proceduralist should be aware that DSA uses dose rates that are an order of magnitude higher than fluoroscopy. Techniques such as variable frame rate (VFR) DSA may be used to decrease overall radiation dose by reducing the total number of DSA frames acquired. Using VFR, different frame rates can be used for different phases of a DSA acquisition (e.g., 4 fps for the arterial phase of an injection, followed by 2 fps for the venous phase, and 1 fps for the parenchymal phase). Changes in frame rate can be timed or triggered by the operator using a hand switch.

Lastly, the application of CBCT may provide added procedural benefit. The preembolization tumoral vascular supply and the postembolization residual tumor blush may be monitored by CBCT to assess treatment effect and guide further treatment decisions. In addition, an arterial 3D map may be obtained by CBCT acquisition during selective angiography. This technique may be useful when performing embolization of spinal tumors, as 3D and multiplanar imaging may identify collateral supply to the anterior spinal artery of Adamkiewicz before embolization. In addition, selective CBCT angiography can then be used to identify aberrant or occult vascular supply and provide navigational potential through the application of navigation software. While this technique has been predominantly applied during liver tumor embolization, the application for MSK tumors can prove advantageous. The technique is typically performed with power injection of undiluted contrast through a 5-Fr catheter or 2.4- to 2.8-Fr microcatheter at 2 to 3 mL/second for a total of 20 to 40 mL, with an acquisition delay of 6 to 8 seconds.

Fluoroscopy/CT Optimization for Osseous Stabilization

Fluoroscopy is a valuable tool during minimally invasive consolidation of pathological fractures or treatment of unstable osseous structures due to malignancy. Fluoroscopic guidance can be employed during cement needle advancement and percutaneous screw placements. Furthermore, the spatial imaging benefits of fluoroscopy are particularly conducive to monitor cement flow and distribution during percutaneous cement injection with real-time monitoring.

Several fluoroscopy settings may be adjusted to best delineate and sharpen bone borders, as well as decrease radiation dose. Higher levels of edge enhancement and narrowing of window width in the gray scale processing may be adjusted to enhance fine detail in bony structures (►Fig. 3). Radiation dose may be decreased without substantially degrading image quality by reducing fluoroscopic pulse rates (2–4 pps) and increasing kV and added filtration. If available, the use of dedicated bone algorithms can improve visibility of bony structures and cryoablation ice balls in CBCT images (►Fig. 4).

Precise needle guidance navigation systems are widely available that use 3D acquisition from CBCT acquisition to facilitate needle and screw advancement (►Fig. 4). The patient should remain immobile on the procedural table during needle guidance, else realignment with a repeat CBCT will be required. Typically, a 5- or 6-second CBCT acquisition is sufficient. Shorter CBCT acquisition protocols reduce imaging time and radiation dose by acquiring fewer projections, at the expense of an increased prevalence of streak and metal artifact. The use of higher kV or added copper filtration may slightly reduce the magnitude of metal artifact.

Multidetector Computed Tomography

Conventional CT, more appropriately referred to as multidetector CT (MDCT), is a valuable image guidance modality with excellent spatial and contrast resolution. CT guidance can be employed during needle placement for biopsy, ablation, and osseous stabilization procedures. In addition, CT may be used to monitor cement flow and distribution for vertebral augmentation and cementoplasty, although real-time imaging capability is limited. Furthermore, CT may be used to monitor ablation margins, which is particularly instrumental to visualize the cryoablation therapeutic ice ball extent and to assess postablation margins with contrast-enhanced imaging.

Modern MDCT offers iterative reconstruction algorithms to reduce image noise and improve detectability, while maintaining or slightly improving spatial resolution.^{13,14} This improvement in image noise is often traded for reduced radiation dose when using iterative reconstruction algorithms.¹⁵ General optimization principles for CT imaging include adapting technical factors to patient size and imaging task.¹⁶ The image reconstruction settings may be configured for the specific imaging task. For MSK interventions, appropriate selection of image thickness, kV, CT reconstruction algorithm (kernel), and window settings are important for optimal visualization. These settings should be adapted based on the required imaging task, with different settings required for seeing bone detail compared with accurately tracking the interface between the cryoablation ice ball and surrounding tissue. Lower kV settings provide better iodine contrast and can improve ice ball visualization, while higher kV settings can reduce metal artifact. Sharp kernels that enhance bony detail and relatively narrow windows are useful when placing needles and screws (►Fig. 5), while moderate kernels are more useful for tracking the ice ball (►Fig. 4). Thinner

Fig. 3 Edge enhancement processing using unsharp masking. The original image (a) is blurred using a Gaussian filter (b). The resultant filtered image is subtracted from the original, which creates an image containing only high spatial frequency (i.e., edge) information (c). The edge image is then combined with the original image (d) to produce the edge-enhanced image (e). Notice that noise is also enhanced in the processed image.

Fig. 4 Painful metastatic renal cell carcinoma to the right iliac wing (a—arrow, hypervascular lesion on contrast-enhanced CT). Treated on a fluoroscopy equipment with embolization (b) followed by cryoablation. Cryoablation probes placed using both ultrasound (c) and cone beam CT needle guidance software (d). Fluoroscopy demonstrates final needle position using narrow windows to facilitate needle placement (e). Cryoablation ice ball margins (arrows) monitored with cone beam CT for deep muscular components and intra-abdominal components near bowel (f) and with US for superficial component near skin (g). Post procedure CT, optimized for soft tissue contrast, demonstrates devascularized zone (arrows) corresponding with ablation zone (h). with improved contrast by wider window and moderate kernels (h).

Fig. 5 Basic CT image optimization for MSK intervention. The use of a “soft” CT kernel limits reproduction of bone detail (a) while the use of a “hard” or “sharp” CT kernel provides cortical detail (b). The reconstruction kernel can be changed without rescanning the patient by performing retrospective image reconstruction.

images are often preferable for visualizing bony detail due to improved spatial resolution, at the expense of increased image noise. During cryoablation, somewhat thicker images may be better for visualizing the therapeutic ice ball due to decreased image noise and improved soft-tissue density contrast.

Several CT hardware and software options exist that may improve MSK interventions. These include CT intervention packages, metal artifact reduction algorithms, and dual-energy CT (DECT). CT intervention packages include techniques such as needle guidance and CT fluoroscopy. Needle guidance software can facilitate accurate targeting of tumors, while metal artifact reduction and DECT may decrease metal artifact and improve tumor visualization during ablation treatments. Recently, several postprocessing applications, including multiplanar reconstruction and virtual monoenergetic DECT, have been integrated into the basic image reconstruction workflow on the CT scanner itself, simplifying clinical workflow.

Intervention packages available on some modern CT systems offer tools that are useful for the planning and navigation of MSK interventions. Basic functionality often includes sequential (axial), spiral (helical), and real-time (CT fluoroscopy) image guidance modes. Spiral image guidance modes image a limited region of the patient using helical CT, and produce multiplanar reformatted images. Such image guidance modes may also incorporate basic needle guidance, allowing the operator to mark the target and entry points and use them to plot a needle trajectory. Some software also reconstructs images in the plane of the marked path, improving needle visualization. Other tools to optimize image guidance may be available, including techniques that automatically introduce a small amount of gantry tilt based on

needle gauge to minimize the appearance of metal artifact from needles and probes. More sophisticated standalone instrument guidance systems can be used with CT. These systems add fiducial markers that are attached to the patient during CT imaging, utilize tip-tracked needles, and provide real-time feedback to the operator regarding the actual needle trajectory and position compared with the planned trajectory. Such systems require a DICOM connection to the CT or hospital PACS to transfer images from the CT to the guidance system.

Metal artifacts are ubiquitous during CT-guided MSK interventions. In some procedures, such as simple biopsies, such artifacts may be neutral or somewhat beneficial, as they can be used for passive tracking of needle trajectory. However, in more complex interventions, they are often a hindrance, obscuring important details such as tumors or ablation margins. Recent advances in metal artifact reduction include metal reduction algorithms and DECT. Beam hardening and photon starvation contribute to metal artifact, and are caused by X-ray beam paths that traverse substantial amounts of metal. Early strategies for reducing metal artifact included increasing the kV to reduce beam hardening and tilting the gantry to reduce the amount of metal in the beam path. While these techniques do confer benefit, there are drawbacks. Gantry tilting alters the visualization of patient anatomy, and increasing the kV often reduces contrast resolution and may increase radiation dose. Metal artifact reduction algorithms typically use segmentation of reconstructed images to identify data in the CT sinogram that is compromised by metal artifact (—Fig. 6). These data can then be replaced via interpolation from surrounding data through an iterative correction process. DECT has also been explored for its potential to reduce metal

Fig. 6 Metastatic thyroid cancer to the right acetabulum status post cementoplasty treated with cryoablation for progressive disease along the anterior acetabulum resulting in worsening pain. Cryoablation ice ball margins were difficult to visualize on CT due to artifact from the prior cementoplasty and cryoablation probe (a). Postprocessing with metal artifact reduction software improved visualization of the ice ball (b).

artifact, typically by reconstructing virtual monoenergetic images at a high kV to reduce the presence of metal artifact (→Fig. 7). The effectiveness of this approach depends on the amount of metal in the beam path—for small to moderate devices such as screws and large gauge needles, virtual monoenergetic images often result in reduced metal artifact, while for large amounts of metal (e.g., a hip prosthesis), virtual monoenergetic images offer no improvement.¹⁷

While CT guidance provides cross-sectional imaging capability and superior contrast resolution compared with fluoroscopy, it does have a few drawbacks. Real-time imaging in CT (i.e., CT fluoroscopy) is limited in terms of both frame rate and patient coverage, and exposes both the patient and operator to much higher radiation dose rates than fluoroscopy. The limited coverage along the long axis of the patient (z-axis) presents challenges for out-of-plane needle placement and monitoring cement injection. While spiral CT guidance provides an arbitrary amount of z-axis coverage, repeated acquisition can be cumbersome with higher radiation doses to the patient when long scans are repeated multiple times. A recent best practices guidelines from the Society of Interventional Radiology discusses optimization of CT-guided interventions in detail.¹⁸

Comparison of Conventional CT to Cone Beam CT

While the technologies underlying conventional CT (MDCT) and CBCT are similar in some ways, there are several basic imaging physics differences that are important to understanding the limitations of each modality.

First, there is a marked difference in projection density (view sampling), which affects the presence and severity of artifacts as well as contrast resolution. During a CBCT acquisition, 200 to 400 frames are acquired as the C-arm rotates approximately 200 degrees around the patient in 3 to 6 sec-

onds. Compare this to MDCT, where thousands of projection samples are acquired in less than a second each time the gantry rotates around the patient. The higher projection density of MDCT, along with projection data acquisition from a complete 360-degree rotation around the patient, reduces the frequency and severity of streak artifacts around prostheses, IR instruments, and exogenous contrast (→Fig. 4f vs. 4h). In general, the higher the projection density, the lower the incidence of such artifacts. The projection density may vary depending on gantry rotation time in MDCT, and often varies with acquisition speed (rotation time) in CBCT.

Second, there are important differences in FOV and imaging size capabilities between CBCT and MDCT. The FOV of CBCT is limited by the size of the flat panel image receptor and thus is smaller than MDCT, although this smaller FOV is adequate for most MSK interventions. In CBCT, the transverse FOV is approximately 25 cm at isocenter with a z-axis (long axis of the patient) coverage of approximately 20 cm. However, implementations of CBCT on systems using robotic C-arms are capable of performing two offset rotations to extend transverse coverage close to that available in MDCT. In comparison, the FOV in MDCT is typically 50 cm at isocenter with arbitrary z-axis coverage, determined by the scan length. The only limitation of z-axis coverage is during sequential or CT fluoroscopy guidance modes, which are typically limited to 1 to 2 cm. While the FOV may be smaller in CBCT, the ability to image large patients is aided by the geometry of CBCT. An angiographic C-arm with CBCT capability can image large patients, as the source-to-image distance for these systems is around 120 cm. Modern MDCT scanners offer bore sizes ranging from 70 to 90 cm, with usable FOV ranging from 50 cm to 70 cm.

Third, scatter radiation from CBCT is higher than MDCT owing to differences in radiation beam widths and size of image receptors. While typical MDCT systems have radiation beam widths ranging from 10 to 65 mm at isocenter, the

Fig. 7 Metastatic prostate cancer to the posterior acetabulum treated by cryoablation for pain control. Dual-energy CT images demonstrate utility of optimal selection of monoenergetic images on soft-tissue contrast and metal artifact. Monoenergetic images reconstructed at 40 keV offer the best soft-tissue contrast but suffer from metal artifact (a). Progressive increase in monoenergetic keV demonstrates decreasing soft-tissue contrast, but also reduction in metal artifact (b–d).

radiation beam width for CBCT is much wider, approximately 200 mm. This leads to a higher scatter-to-primary ratio in CBCT compared with MDCT, which negatively affects contrast and low contrast detectability¹⁹ (► Fig. 8).

Lastly, motion artifact is more pronounced in CBCT for the same amount of patient movement. MDCT acquires data from a limited volume of the patient during each subsecond rotation, while CBCT samples the entire imaged volume during a rotation of 3 to 6 seconds. Patient motion that occurs during an MDCT scan contaminates only a small subset of images, while motion that occurs during a CBCT scan contaminates the entire imaged volume.

Direct comparisons of radiation dose between MDCT and CBCT are difficult, but have been performed using anthropomorphic phantoms and specific imaging tasks,^{20–22} with studies identifying either CBCT or MDCT as using the lower radiation dose. A recent study found that when factory default imaging presets were used and image noise levels

were matched between CBCT and MDCT, the radiation dose difference was minimal.²³

Magnetic Resonance Imaging

Magnetic resonance imaging guidance has great applicability for the treatment of MSK malignancy during biopsy and ablation procedures.^{24–26} Biopsy of lesions otherwise difficult to visualize with other modalities may be possible with MR guidance due to the unparalleled soft-tissue contrast of MRI. While heat-based ablation equipment, including radio-frequency and microwave, are often incompatible with MRI, some cryoablation systems are compatible with MRI. During cryoablation, the therapeutic ice ball can be monitored owing to its absence of MR signal. In addition, MRI is still the most flexible modality for image orientation, offering direct imaging in any plane. While modern interventional CT systems now offer direct multiplanar reformats (axial,

Fig. 8 Metastatic melanoma to the T10 vertebral body resulted in pathologic compression fracture and prompted palliative treatment with kyphoplasty. Pretreatment MDCT with mild streak artifacts associated with spine hardware (a - arrow). Intraprocedural CBCT (b) allows spatial definition, with decreased contrast and resolution, but increased streak artifact (arrows) owing to decreased view sampling and limited angle data acquisition.

sagittal, and coronal), MRI provides the off plane imaging potential at acquisition and does not require reformatting.

Early work on MR-guided MSK interventions focused on open MRI systems that facilitated easy patient access, including “double donut” and biplanar system designs. These systems suffered from low field strength and reduced magnetic field homogeneity compared with cylindrical MR systems. Modern high-field MRI systems (e.g., 1.5 T) are available in short, wide bore configurations that facilitate patient access while offering high-resolution imaging.

Most MRI-guided MSK interventions rely on fast imaging sequences such as short TR gradient echo or single-shot fast-spin echo techniques. Gradient echo techniques are prone to susceptibility and chemical shift artifacts; however, susceptibility artifact from devices and instruments is often used for passive tracking of such devices (→ Fig. 9). The use of high field strength MRI systems (1.5 T and higher) allows for high temporal resolution with acceptable spatial resolution during MR-guided interventions. Parallel imaging techniques also have promise for moving MRI-guided interventions closer to real time. These techniques, however, require additional hardware and software for implementation.

MRI is also used in MR-guided high-intensity-focused ultrasound (MRgHIFU), a noninvasive ablation technique that uses an ultrasound transducer to thermally ablate tissue in the patient under MRI guidance. MRgHIFU is used in sites not susceptible to motion, with most use currently in uterine fibroid treatment, but ablation of breast, prostate, brain, and even bone lesions have also been explored.²⁷ The periosteal region of benign or malignant bone tumors can be treated for pain palliation, and there may be the possibility of a tumor therapeutic effect depending on the location of the tumor and the thickness of the cortex.²⁸ MRgHIFU treatments are time consuming owing to the small area of the focused ultrasound beam.

There are numerous mechanisms by which contrast can be generated in MRI, and many of these mechanisms can be used to assess either relative or absolute temperature changes in tissue, with different sensitivity, spatial, and temporal resolution.^{28,29} No other imaging modality is as sensitive to temperature change as is MRI, and MRI is currently used for thermometry to monitor temperatures during thermal ablation,²⁸ most commonly for laser-induced thermal therapy and MRgHIFU, but also for radiofrequency and microwave ablation. MR thermometry for cryoablation is currently being explored in research laboratories.^{30,31}

Even in the absence of ionizing radiation, there are potential safety hazards in the MRI environment, including ferromagnetic objects become dangerous projectiles, device interference, device displacement, and heating of tissue and conductors from radiofrequency energy. Practices performing MRI-guided interventions should have MRI safety policies and procedures in place.³²

Integrated Multimodality Imaging Units

Growing interest exists for integrated multimodality imaging suites that are located in a single room or in separate rooms divided by a retractable barrier. Traditionally, integrated multimodality units have been custom built for specific interests or visions. Given fast-paced technological advancements that decrease cost and enhance integration of different imaging modalities, these units have gained increasing worldwide interest. The potential advantages of multimodality integration have been suggested as increasing procedural capabilities, improving patient flow, and decreasing potential utilization costs (→ Fig. 10).

The most common hybrid unit is the CT-fluoroscopy combination, while MR-fluoroscopy and PET-CT combinations

Fig. 9 Metastatic fibrosarcoma to the right posterior pleura treated on a hybrid MRI–fluoroscopy unit (a—preprocedure CT). Initial treatment by embolization of subcostal arteries to minimize bleeding risk (b), followed immediately by cryoablation (c—initial procedure MRI TRUF1, d—needle placements MRI TRUF1). Note difference in soft tissue and bone prominence when assessing ice ball extension into adjacent tissues on MRI TRUF1 sequence (e) and MRI fast spin-echo sequence (f), with increase lung detail in the former and improved bone detail in the latter.

Fig. 10 Metastatic non-small cell lung cancer to the left scapula resulted in severe pain (a—plain radiograph; b—contrast-enhanced MRI). Treatment in a combination CT–fluoroscopy unit for embolization of the hypervascular lesion (c—preembolization angiogram, d—postembolization angiogram) followed immediately by cryoablation (e).

have also been described in the procedural suite (►Fig. 9). Modern MRI–fluoroscopy hybrid systems locate the angiography system outside the 5 Gauss line of the MRI system and to avoid interference of one modality with the other. Build in ultrasound integration into fluoroscopy and CT units has also emerged as a potentially beneficial union.

Mobile Imaging

High end mobile fluoroscopes and mobile CT systems have become attractive options for performing MSK-guided interventions owing to their reduced cost and mobility. The mobility allows shared use in multiple procedural rooms. Furthermore, the mobility may allow the proceduralist to position a mobile CT in a fluoroscopy procedure room, or vice versa, to obtain functionality similar to a combined CT–fluoroscopy unit. Such systems have X-ray generators and tubes that are less powerful than installed systems, offer fewer features, and often have reduced image quality compared with installed systems. This may limit the patient population that can be treated using mobile systems (e.g., the image quality when imaging very large patients may not be adequate) and the types of procedures that can be performed. National Council on Radiation Protection and Measurements Report 168 states that potentially high-dose procedures should be performed using fluoroscopic systems that incorporate modern dose-reduction and dose-measurement technology.^{33,34} These features may not be available on mobile fluoroscopes. Radiation protection is a particular concern when using mobile X-ray imaging, and special consideration should be paid to ensuring that all personnel in the room during X-ray production have necessary protective equipment.

Conclusion

Advancement in the interventional oncology treatment of MSK malignancy benefits from a firm application of multiple imaging modalities. New imaging software capabilities continue to provide the opportunities to explore new treatment possibilities. We have herein briefly described the most prevalent imaging modalities and new imaging technologies to assist in the treatment of MSK malignancy.

Conflict of Interest
None.

References

- Kurup AN, Callstrom MR. Expanding role of percutaneous ablative and consolidative treatments for musculoskeletal tumours. *Clin Radiol* 2017;72(08):645–656
- Wallace AN, Greenwood TJ, Jennings JW. Radiofrequency ablation and vertebral augmentation for palliation of painful spinal metastases. *J Neurooncol* 2015;124(01):111–118
- Nakatsuka A, Yamakado K, Maeda M, et al. Radiofrequency ablation combined with bone cement injection for the treatment of bone malignancies. *J Vasc Interv Radiol* 2004;15(07):707–712
- Munk PL, Rashid F, Heran MK, et al. Combined cementoplasty and radiofrequency ablation in the treatment of painful neoplastic lesions of bone. *J Vasc Interv Radiol* 2009;20(07):903–911
- Hartung MP, Tutton SM, Hohenwarter EJ, King DM, Neilson JC. Safety and efficacy of minimally invasive acetabular stabilization for preacetabular metastatic disease with thermal ablation and augmented screw fixation. *J Vasc Interv Radiol* 2016;27(05):682–688.e1
- Claudon M, Dietrich CF, Choi BI, et al. Guidelines and good clinical practice recommendations for contrast enhanced ultrasound (CEUS) in the liver—update 2012: a WFUMB-EFSUMB initiative in cooperation with representatives of AFSUMB, AIUM, ASUM, FLAUS and ICUS. *Ultraschall Med* 2013;34(01):11–29
- Dietrich CF, Averkiou M, Nielsen MB, et al. How to perform contrast-enhanced ultrasound (CEUS). *Ultrasound Int Open* 2018;4(01):E2–E15
- Cosgrove D. Microbubble enhancement of tumor neovascularity. *Eur Radiol Suppl* 1999;3:S413–S414
- Bang N, Bachmann Nielsen M, Vejborg I, Mellon Mogensen A. Clinical report: contrast enhancement of tumor perfusion as a guidance for biopsy. *Eur J Ultrasound* 2000;12(02):159–161
- Pass B, Jafari M, Rowbotham E, Hensor EMA, Gupta H, Robinson P. Do quantitative and qualitative shear wave elastography have a role in evaluating musculoskeletal soft tissue masses? *Eur Radiol* 2017;27(02):723–731
- Hakime A, Yevich S, Tselikas L, Deschamps F, Petrover D, De Baere T. Percutaneous thermal ablation with ultrasound guidance. Fusion imaging guidance to improve conspicuity of liver metastases. *Cardiovasc Intervent Radiol* 2017;40(05):721–727
- Jones AK, Balter S, Rauch P, Wagner LK. Medical imaging using ionizing radiation: optimization of dose and image quality in fluoroscopy. *Med Phys* 2014;41(01):014301
- Euler A, Solomon J, Marin D, Nelson RC, Samei E. A third-generation adaptive statistical iterative reconstruction technique: phantom study of image noise, spatial resolution, lesion detectability, and dose reduction potential. *AJR Am J Roentgenol* 2018;210(06):1301–1308
- Chen B, Ramirez Giraldo JC, Solomon J, Samei E. Evaluating iterative reconstruction performance in computed tomography. *Med Phys* 2014;41(12):121913
- Solomon J, Mileto A, Ramirez-Giraldo JC, Samei E. Diagnostic performance of an advanced modeled iterative reconstruction algorithm for low-contrast detectability with a third-generation dual-source multidetector CT scanner: potential for radiation dose reduction in a multireader study. *Radiology* 2015;275(03):735–745
- Tam AL, Ensor JE, Zvavanjanja RC, et al. Standardizing CT-guided biopsy procedures: patient dose and image noise. *AJR Am J Roentgenol* 2015;205(04):W390–W399
- Yu L, Leng S, McCollough CH. Dual-energy CT-based monochromatic imaging. *AJR Am J Roentgenol* 2012;199(5, Suppl):S9–S15
- Jones AK, Dixon RG, Collins JD, Walser EM, Nikolic B; on behalf of the Society of Interventional Radiology Health and Safety Committee. Best practices guidelines for CT-guided interventional procedures. *J Vasc Interv Radiol* 2018;29(04):518–519
- Siewerdsen JH, Jaffray DA. Cone-beam computed tomography with a flat-panel imager: magnitude and effects of x-ray scatter. *Med Phys* 2001;28(02):220–231
- Steuwe A, Geisbüsch P, Schulz CJ, Böckler D, Kauczor HU, Stiller W. Comparison of radiation exposure associated with intraoperative cone-beam computed tomography and follow-up multidetector computer tomography angiography for evaluating endovascular aneurysm repairs. *J Endovasc Ther* 2016;23(04):583–592
- Bai M, Liu B, Mu H, Liu X, Jiang Y. The comparison of radiation dose between C-arm flat-detector CT (DynaCT) and multi-slice CT (MSCT): a phantom study. *Eur J Radiol* 2012;81(11):3577–3580
- Kwok YM, Irani FG, Tay KH, Yang CC, Padre CG, Tan BS. Effective dose estimates for cone beam computed tomography in interventional radiology. *Eur Radiol* 2013;23(11):3197–3204
- Jones AK. Abstract 1017: An Apples to Apples Comparison of Radiation Dose and Image Quality between Flat Panel CT and

- Multidetector CT. Society of Interventional Radiology Annual Meeting, Los Angeles, CA; 2018
- 24 Smith KA, Carrino JA. MRI-guided interventions of the musculoskeletal system. *J Magn Reson Imaging* 2008;27(02):339–346
 - 25 Sequeiros RB, Ojala R, Kariniemi J, et al. MR-guided interventional procedures: a review. *Acta Radiol* 2005;46(06):576–586
 - 26 Ahrar K, Sabir SH, Yevich SM, et al. MRI-guided interventions in musculoskeletal system. *Top Magn Reson Imaging* 2018;27(03):129–139
 - 27 Napoli A, Anzidei M, Ciolina F, et al. MR-guided high-intensity focused ultrasound: current status of an emerging technology. *Cardiovasc Intervent Radiol* 2013;36(05):1190–1203
 - 28 de Senneville BD, Mougenot C, Quesson B, Dragonu I, Grenier N, Moonen CTW. MR thermometry for monitoring tumor ablation. *Eur Radiol* 2007;17(09):2401–2410
 - 29 Zhu M, Sun Z, Ng CK. Image-guided thermal ablation with MR-based thermometry. *Quant Imaging Med Surg* 2017;7(03):356–368
 - 30 Wansapura JP, Daniel BL, Vigen KK, Butts K. In vivo MR thermometry of frozen tissue using $R2^*$ and signal intensity. *Acad Radiol* 2005;12(09):1080–1084
 - 31 Overduin CG, Fütterer JJ, Scheenen TW. 3D MR thermometry of frozen tissue: feasibility and accuracy during cryoablation at 3T. *J Magn Reson Imaging* 2016;44(06):1572–1579
 - 32 Kanal E, Barkovich AJ, Bell C, et al; Expert Panel on MR Safety. ACR guidance document on MR safe practices: 2013. *J Magn Reson Imaging* 2013;37(03):501–530
 - 33 National Council on Radiation Protection and Measurements. Radiation Dose Management for Fluoroscopically-Guided Interventional Medical Procedures. NCRP Report 168. Bethesda, MD: NCRP; 2011
 - 34 IEC. International Electrotechnical Commission. Medical Electrical Equipment—Part 2–43: Particular Requirements for the Basic Safety and Essential Performance of X-Ray Equipment for Interventional Procedures, IEC 60601–2–43 ed 2.0. Geneva: International Electrotechnical Commission; 2010