Top-Level Design and Simulated Performance of the First Portable CT-MR Scanner

YUTING PENG1, MENGZHOU LI2, (Member, IEEE), JACE GRANDINETTI1, GE WANG2, (Fellow, IEEE), AND XUN JIA1,3

1Innovative Technologies of Radiotherapy Computations and Hardware (iTORCH) Laboratory, Department of Radiation Oncology, University of Texas Southwestern Medical Center, Dallas, 75390 TX, USA
2Biomedical Imaging Center/Cluster, Rensselaer Polytechnic Institute, Troy, 12180 NY, USA
3Department of Radiation Oncology and Molecular Radiation Sciences, Johns Hopkins University, Baltimore, MD 21287, USA

Corresponding authors: Xun Jia (xunjia@jhu.edu) and Ge Wang (wangg6@rpi.edu)

This work was supported in part by the funds from the National Science Foundation (NSF) under Grant 1636933 and Grant 1920920, and in part by the Cancer Prevention and Research Institute of Texas under Grant RP200573.

Yuting Peng and Mengzhou Li contributed equally to this work.

ABSTRACT Multi-modality imaging hardware can be integrated in a single gantry to collect diverse datasets for complementary information and spatiotemporal correlation, with excellent examples including PET-CT and PET-MRI. However, there is no CT-MRI prototype up to today due to technical challenges and associated cost-benefit considerations. Thanks to the rapid development of medical imaging, it becomes feasible now to integrate cost-effective CT and MRI imagers together for portability, popularity, and point of care. In this paper, we present the top-level design of the first portable CT-MRI system and evaluate its imaging performance via realistic numerical simulations. In this CT-MRI system, the magnet made of two neodymium (NdFeB) rings of about 40.0 cm radius forms a magnetic field of about 57 mT at the isocenter and has a gap of 11.3 cm to accommodate the rotating CT gantry. The targeted MR imaging field of view (FOV) is a sphere of ∼15 cm in diameter and that of CT is approximately 20 cm diameter in axial direction and 5 cm in longitudinal direction. Our results show a great potential of such a hybrid system. The proposed CT-MRI system will be valuable in applications such as imaging in underdeveloped countries, disaster scenes and battle fields.

INDEX TERMS Computed tomography (CT), magnetic resonance imaging (MRI), multi-modality imaging.

I. INTRODUCTION Multimodality medical imaging is well known to deliver tremendous utilities in a wide spectrum of clinical scenarios [1]. In fact, each imaging modality produces information in a well-defined scope, as characterized by spatial/temporal resolution, signal-to-noise ratio, structural, functional and molecular features, etc. Although these well-defined scopes may overlap, no single modality could derive all the information another modality offers for a major clinical task, as modern medicine often calls for comprehensive evaluation of the subject from a variety of aspects with different imaging modalities. While it is possible to acquire images of different modalities on separate scanners, thereby achieving multi-modal imaging via post-scan image fusion, this approach inevitably suffers from issues such as required image registration, associated uncertainty, not to mention challenges associated with logistics, burdens and complications of long scan time between scanners caused by moving patients and dynamic changes in them, especially in contrast-enhanced studies and critical care conditions where synchrony and efficiency of imaging procedures is vital.

Over the years, several types of hybrid scanners have been developed for simultaneous multi-modal imaging. PET-CT is now widely available, serving as an indispensable tool for cancer diagnosis, staging, and treatment response assessment [2]. Novel PET-MRI scanners allow characterization of metabolic activities empowered by MRI to delivery structural information in rich soft tissue contrast without ionizing radiation [3]. More importantly, in these situations, imaging data
from different modalities acquired at the same time of the patient permit joint image reconstruction and processing to synergistically take advantages of different images, enhancing image quality and performance in clinical tasks. In contrast to these multi-modality imaging progresses, CT-MRI is the only major remaining multi-modality imaging technology yet to be developed. In our previous Vision 2020 paper [4], we have suggested that such a CT-MRI combination could have profound impacts on several major clinical scenarios such as cardiac diagnosis and contrast-enhanced cancer imaging. Nonetheless, integrating CT and MRI scanners for simultaneous imaging is a rather challenging undertaking. Conventional MRI scanners are often designed with a high-strength (commonly, 1–3 T) homogeneous (of the order of a few parts per million) magnetic field for a high signal to noise ratio (SNR) and imaging performance within a clinically acceptable scan time. This means that the field is susceptible to perturbations from nearby ferromagnetic materials and the imaging performance is easily affected by the x-ray tube and detector assembly, when being integrated with a CT scanner. The weak MR signal is also prone to radio-frequency (RF) signals generated by nearby devices, which deteriorates SNR of MR images. Conversely, the fringe magnetic field over the x-ray tube and detector assembly affects their functions due to deflection of the electron beam and malperformance of electronics in the magnetic field. Last but not the least, bulky MR and CT components have to be integrated in a compact space, further enhancing the challenge of the CT-MRI system design.

Fortunately, over the past years there have been exciting developments towards addressing these challenges. To combine x-ray radiography with MRI scanner, researchers at Stanford group demonstrated the feasibility of operating an x-ray tube and a detector in the magnetic field [5], [6]. In the radiotherapy field, recent advances of MRI-guided radiotherapy achieved by a combination of MRI with medical linear accelerator demonstrated clinical MR imaging functions in close proximity to a powerful x-ray production device [7], [8].

In this study, we report our on-going effort towards prototyping the first portable CT-MRI system employing a non-conventional MR scanner design. Specifically, we propose an ultra-low-field (ULF) MR scanner operating at a 57 mT flux density range together with an inhomogeneous magnetic field design. The medical relevance of ultra-low-field MRI has recently been demonstrated in several examples [9], [10], [11], [12]. We prefer an inhomogeneous field design to reduce the technical challenges associated with building a highly homogeneous magnet. The feasibility of using an inhomogeneous magnetic field for MRI has been previously shown in several systems, such as single-sided MRI scanners. In our design, the low strength and inhomogeneous field make the MRI system more robust to perturbations from the CT components, and the low-fringe field facilitates the integration with the CT subsystem.

II. METHODS
A. TOP-LEVEL SYSTEM DESIGN
1) MRI SUBSYSTEM
The overall design of the CT-MRI scanner is shown in Figure 1. We consider an ULF MR design for the purpose of cost reduction and easy integration with CT subsystem. For this ULF MRI subsystem (Figure 1(a)), the magnet consists of two rings of neodymium (NdFeB) permanent magnets with a gap of 11.33 cm for the x-ray beam of the CT subsystem to pass through. Both rings are configured to orient the permanent magnets in a Halbach format to form the magnetic field uniformly in the y direction. The Halbach array has been widely used in MR-based devices to create highly uniform magnetic fields using magnet bars. Prior studies have also employed novel optimization techniques to design the magnet array [13], [14], [15]. In this study, we design the magnet in a trial-and-error fashion, e.g. by adjusting number of magnets per ring, radii of the rings, and the separation between rings, to yield the configuration meeting our needs. On one ring, 34 bar magnets in grade N52 of 5.08 × 5.08 × 15.24 cm³ are aligned along a circle of radius 40.6 cm, while the second ring consists of 32 bar magnets of 5.08 × 5.08 × 20.32 cm³ on a circle of 40.0 cm radius. These magnets, their sizes, and arrangement define a magnetic field of about 57 mT at the isocenter. The Halbach ring and magnetic flux density distribution on the z = 0 plane is shown in Figure 1(c). The magnetic field is slightly stronger towards the smaller ring, naturally inducing a field gradient which can be utilized for spacial encoding along this direction (z axis) to reduce the required number of gradient coils. The resulting gradient strength is about 25 mT/m. To hold the magnet bars in place, 3D printing technology can be used to create molds based on the designed positions and orientations of magnet bars. Aluminum plates will be used to secure the molds in place. The inner radii the magnet assembly are 36.8 cm and 36.1 cm for the large and small rings respectively.

Two sets of self-shielded gradient coils are designed for spacial encoding along patient lateral directions (x and y directions). The four layers of the two gradient coils are parallel to the inner surface of the magnet, occupying a 2.5 cm space (Figure 1(a)). The target field method is employed to determine the coil patterns [16] to generate a linearly varying field within the planned field of view (FOV) of 15 cm in diameter, while maximally minimizing magnetic field at nearby metal parts to avoid eddy current effect. Figure 1(d) presents the resulting coil pattern for the y-gradient coil. Note that there is a 11.5 cm gap at the center of each coil to avoid blocking the x-ray beam of the CT system.

We plan to use a solenoid transmit/receive RF coil. It follows a standard design for an MR scanner and is not shown here. As for the data acquisition pipeline, standard MRI techniques can be used with a spectrometer for sequence control, RF and gradient amplifiers for signal amplification. Care must be taken to accommodate relatively low SNR due to the low field strength and signal loss.
A cone-beam scanning geometry is adopted for the CT subsystem, as shown in Figure 1(b). The source-to-detector distance is 1,125 mm, and source-to-isocenter distance 500 mm. These values are close to those for a medical CT scanner and large enough to avoid geometrical interference with the MRI subsystem. The medical pulse x-ray source of interest has a focal spot of ~0.5 mm, a half-cone angle of 18°, a kVp range between 100 to 400 kVp, and a current range between 1 to 13 mA. To cover a major part of the field of view, we consider two PaxScan 2520DX flat panel detectors (VAREX Imaging, Salt Lake City, UT) combined to form a larger detector. The x-ray sensor material is amorphous silicon and equipped with a direct deposit CsI conversion screen. The detector has a pixel pitch of 127 µm and supports a resolution 1 lp/mm at >48%. The detector array is of 1536 × 1920 pixels. The energy detection range is 40 to 160 keV. The intrinsic frame rate is 12.5/s but can be increased when working in binning modes. The detector weighs 5.5 lbs. We consider that the detector working in the 4 × 4 binning mode. Due to the obstruction of the MRI assembly, the x-ray beam is confined to a half cone-beam angle 3° in the longitudinal direction to avoid scattering from the MRI components. In this setting, the effective detector array becomes 384 × 960 pixels with pixel pitch of 508 µm. With this setting, the CT FOV is approximately 20 cm diameter in axial direction and 5 cm in longitudinal direction.

The x-ray source and detector are mounted on a rotation gantry and a slip ring design is used for continuous data communication with the rotor, as well as power delivery to the tube and panel. This is a drum type slip ring with a inner diameter of 950 mm and supports a maximum rotation speed of 120 rpm.

B. SIMULATION OF MR DATA ACQUISITION AND IMAGE RECONSTRUCTION

We consider 2D spin-echo type sequences for MR data acquisition, because it helps to alleviate signal dephasing caused by the main magnetic field inhomogeneity. Without loss of generality, for each 2D acquisition, a RF wave of frequency $f$ only excites spins around a curved iso-magnetic field surface defined by the condition $f = \gamma B_0(x, y, z)$, where $B_0(x, y, z)$ is the magnetic flux density field, and $\gamma$ is the gyromagnetic ratio. For a given phantom defined by volumetric images of longitudinal relaxation time $T_1(x, y, z)$, transverse relaxation time $T_2(x, y, z)$, and spin density $\rho(x, y, z)$, let a spin-echo sequence be specified with repetition time $TR$ and echo time $TE$, the integrated signal for a pulse with a slice selection frequency $f$ is

$$S(k_x, k_y) = \sum_{x,y,z} I_w([-f - \gamma B_0(x, y, z)]\rho(x, y, z)B_0(x, y, z)^2$$

$$\times \left[1 - \exp\left(-\frac{TR}{T_1(x, y, z)}\right)\right] \exp\left[-\frac{TE}{T_2(x, y, z)}\right]$$

$$\times \exp[-TE(\delta y \Delta B_0)]\exp[-i(k_x x + k_y y)],$$

(1)

where $I_w[.]$ is an index function with $I_w[|x|] = 1$ if $|x| < w/2$ and 0 otherwise. $w$ is the bandwidth of the RF pulse. The term $\exp[-TE(\delta y \Delta B_0)]$ describes dephasing effect due to the magnetic field gradient, and $\delta$ is a constant obtained by calibrating this model using a MR scanner in our lab.

Noise cannot be neglected in an ULF MRI scan due to the low magnetic field yielding a weak signal and the field inhomogeneity induced dephasing effect, as well as the presence of free electrons in RF coil or hardware vibrations in practice. Hence, we add to $S(k_x, k_y)$ Gaussian white noise with standard deviation $\sigma = C_0 \sqrt{B_0 / \sqrt{T_{ACQ}}} [17]$, where $C_0$ is a calibration constant obtained empirically on an MRI scanner available in our lab, and $T_{ACQ}$ is the total acquisition time. The noise is signal independent.

After collecting $k$ space data, a 2D image $m(x, y)$ is reconstructed using Fourier-transform based reconstruction algorithm. To suppress image noise, we denoise the images using BM3D algorithm [18]. After collecting images for a series of RF frequency, we can obtain 2D images defined on a set of curved iso-surfaces of magnetic flux density, each characterized by $f = \gamma B_0(x, y, z)$, as illustrated in Figure 2. To generate a volumetric image, we resample these 2D images to a Cartesian grid according to the known positions of these iso-surfaces.
C. SIMULATION OF CT DATA ACQUISITION AND IMAGE RECONSTRUCTION

To simulate the CT imaging process, we consider a circular scan mode with 720 projection views per rotation. To simulate x-ray projections, a monochromatic radiation model is utilized for simplicity, and the detected signal is expressed as

\[ p = f_{\text{Poisson}} \left[ N_0 \exp(-P \mu(x, y, z)) \right], \]

where \( P \) stands for the CT projection system matrix, \( \mu(x, y, z) \) the volumetric image of the x-ray attenuation coefficient, \( f_{\text{Poisson}} [-] \) the Poisson process, \( p \) the recorded projections in a vector form, and \( N_0 \) the incoming number of photons, which is set to \( 3 \times 10^6 \) in this study.

With the simulated x-ray data, an image is reconstructed using the standard Fredkamp-Davis-Kress (FDK) algorithm [19]. Since the CT FOV is relatively small compared to the whole human body, even with the combination of two detectors, the CT system is subject to lateral truncation of projections. Hence, an iterative method for interior tomography [20] is used to improve reconstruction results. Specifically, a Simultaneous algebraic reconstruction technique with a total variation (SART-TV) regularization is coded to solve the optimization problem

\[ \arg\min_{\mu} \| P \mu(x, y, z) + \log(p/N_0) \|^2 + \lambda \| \nabla \mu(x, y, z) \|_1, \]

where \( \lambda \) is a parameter balancing the data fidelity term and the sparsity constraint. The ASTRA toolbox is used during the forward projection and volumetric reconstruction [21].

D. PERFORMANCE EVALUATION

We performed numerical simulations with two phantoms to demonstrate the performance of the proposed CT-MRI scanner. The first phantom is a 3D Shepp-Logan phantom. This phantom represents typical brain tissues including scalp, bone, cerebrospinal fluid (CSF), gray and white matters, and tumor. For CT simulation, the attenuation values are chosen in reference to the original specification of the 2D Shepp-Logan phantom [22]. As the Shepp-Logan phantom only defines relative density values to water in the interval \([1.0, 2.0]\), we assign x-ray attenuation based on the density values. For MR simulation, the parameters of proton density \( \rho \), relaxation times \( T_1 \) and \( T_2 \) are set in reference to [23].

To bear more clinical relevance, we employ another phantom generated from the publicly-available, anatomically detailed, 3D CT and MRI database from the Visible Human Project (VHP) of the National Library of Medicine [24]. The male patient images are chosen in this study. X-ray attenuation coefficients are derived from the CT images. In the MRI simulation, since only proton density weighted, \( T_1 \) weighted and \( T_2 \) weighted MR images are provided in the database, we derive the proton density image, longitudinal and transverse relaxation times by solving signal equations associated with the MR sequences.

III. RESULTS

A. MAGNETIC FIELD DISTRIBUTION

Figure 3 presents the magnetic flux density distribution generated by the main magnet. For a targeted spherical imaging field of view (FOV) of 15 cm in diameter, the flux density strength is \( 55 \sim 68 \) mT. As shown in Figure 3(c), the magnetic flux density generally decays monotonically along the \( z \) direction, creating a gradient about \( 25 \) mT/m in the FOV, which offers the opportunity for spatial encoding along this direction without a gradient coil. In Figure 3(a) and (b), the iso-surfaces of various magnetic fields are plotted. These surfaces are planned for data acquisition and image reconstruction by performing 2D scans on these surfaces. The separation between neighboring surfaces is \( \sim 0.4 \) cm. It is clearly shown that these surfaces are not planar, which hence requires image domain interpolation to generate a volumetric image defined on the Cartesian grid.

The magnetic flux density strength drops to \( \sim 20 \) mT outside the magnet at a radius of 50 cm from iso-center, for the positions of the x-ray tube, and \( \sim 3 \) mT at the CT detector position with a radium of 62.5 cm from the iso-center. These low magnetic field is expected to be not interfering the normal functions of these CT components [25].

B. IMAGING PERFORMANCE

We first present simulation results for the Shepp-Logan phantom in Figure 4. For the MR simulation, the relaxation time
FIGURE 4. Orthogonal views of (a) the proton density of the 3D Shepp-logan phantom, (b) simulated MR images (TR = 1,000 ms, TE = 80 ms, and NEX = 16), and (c) simulated CT images. The dashed red circles indicate the targeted MR FOV of 15 cm in diameter.

TR and TE are 1,000 ms and 80 ms, respectively. The slice thickness is chosen as 4 mm, and the voxel size on the axial planes is $1 \times 1$ mm$^2$. To reduce noise, we repeated data acquisition with the number of excitations (NEX) being set to 16. The images in Figure 4(a) display proton density of the phantom in three orthogonal views to show the structure of the phantom. Figure 4(b) shows the resulting MR image. The CT simulation results of the Shepp-Logan phantom are in Figure 4(c). The image voxel size is 0.5 mm$^3$.

Due to the narrow x-ray beam through the gap between the two magnet rings, the longitudinal coverage of the CT scan is $\sim 5$ cm.

Figure 5 presents the axial view of the phantom in CT and MR images with different reconstruction and processing methods. For CT, on the lateral direction, the limited FOV introduces truncation artifacts in the conventional FDK reconstruction result, including the bright distorted parts in the image and the cupping effect in the vertical profile. Also, the three small structures close to the FOV boundary can be hardly discerned from the background due to the low contrast between the tumor and the gray matter. Those issues are successfully addressed with the interior tomography technique as demonstrated in the SART-TV result ($\lambda = 0.0003$). The tumors with only 10 HU contrast from the background (pointed by the red arrow) are clearly revealed thanks to the noise suppression capability of the technique. The cupping and truncation artifacts are also successfully removed from the reconstruction, and the profiles align well with the ground truth counterparts.

As for the MR images, the image reconstructed by Fourier algorithm had a large amount of noise due to the low magnetic field strength and field inhomogeneity-induced signal dephasing. Using the BM3D algorithm effectively suppressed image noises.

Figure 6 presents the results to demonstrate image resolution performance of the imaging systems. As such, we embed in the phantom bar patterns along the three major directions, with the bar width ranging from 0.5 mm to 5 mm. It is observed that the smallest bar at width 0.5 mm can be still observed, although the contrast reduces for small bars. To illustrate the performance of the proposed system with more clinical relevance, we present in Figure 7-9 results for different body sites using the VHP data. In the brain...
case shown in Figure 7, we remark that the relatively low resolution along superior-inferior direction in the MR image is intrinsically caused by the phantom itself. In fact, in the VHP MR image dataset the slides spacing is 3 mm, which limited the image resolution in the simulation results.

We present in Figures 8 and 9 the results from the neck and chest sites. In each figure, in addition to presenting the simulation results of the CT and MR images, we also generated a blended view by displaying CT and MRI images in different squares. The capability of integrated CT-MRI imaging in the same spatial coordinate system is expected to offer advantages in a variety of clinical tasks, such as disease diagnosis and therapy planning.

IV. DISCUSSIONS

In this paper, we have proposed a top-level design for an integrated CT-MRI system. The design is featured by an unconventional MRI subsystem with an ultra-low inhomogeneous main magnetic field and a slim CT subsystem. The advantages of taking this approach are multiple folds. The first and most profound advantage is to ensure system integration and compatibility between CT and MR components. Reducing the strength of the main magnetic field makes the field at CT components, i.e. x-ray tube and flat panels, work without any significant electromagnetic interference. If additional shielding is needed to reduce the field strength further, it is straightforward to use shielding materials, such as mu metals, to isolate the CT and MRI subsystems bi-directionally.

Employing an inhomogeneous field design naturally increased tolerability to uncertainties in the system engineering, as compared to the traditional homogeneous field design. Realizing the homogeneous field usually requires complex approaches, such as active and passive shimming, to achieve the targeted homogeneity level of a few parts per million. In contrast, with the inhomogeneous field, our design does not require a specific level of homogeneity, and the imaging and reconstruction process is performed with the magnetic field distribution as prior information. This conceptual change is expected to substantially reduce the engineering challenge in the design of the magnetic field distribution. In the actual hardware system, even if the achieved field...
deviated from the design, as long as the deviation is small, e.g. still satisfying the condition of monotonicity along the z-axis, the field is still usable for MRI.

As demonstrated in our study, the magnetic main field varying along the z-axis can naturally serve as the spatial encoding field along this direction, hence eliminating the need for one of the three spatial gradient coils in classical MR scanners. This helps reduce the complexity of the system integration within a compact space, lowering the system cost and simplifying the system maintenance.

Our design is also featured by a relatively large bore size of 67 ~ 68 cm in diameter at the inferior and the superior sides. This is larger than commonly used MR scanners. This large bore allows flexible patient positioning to place the anatomical part of interest in the imaging FOV. In principle, a smaller bore size would help increase field strength, being beneficial from SNR perspective. For this ultra-low-field design, a slight increase in field strength due to reduced bore diameter would not significantly translate to a significant SNR improvement. Hence, we elected a large bore size design to give freedom of patient positioning. This design should be favorable for patients with claustrophobia.

The current design aims at a low-cost solution for a CT-MRI system. The magnet was uniquely designed for this system, which uses >200 NdFeB cubic magnets arranged in two rings with a gap to accommodate a CT subsystem. Together with aluminum plates and other supporting structures, the cost of the magnet is ~$20k. The MR gradient coils, RF coils, and spectrometer follow the standard design for conventional MRI, and price is expected to be ~$50k. Together with the CT subsystem with an expected hardware cost of ~$100k, the total cost of the proposed system is estimated to be under ~$250k including other accessories, which is very attractive in a good number of use cases. These numbers are bulk estimates and may subject to variation. If the system is for commercial production, additional factors have to be included, such as development costs. This low-cost nature makes the proposed CT-MRI system particularly attractive for underdeveloped countries. We also note that we chose a permanent magnet design for cost consideration in this on-going research project. Increasing field strength using superconducting magnet is certainly possible, although at an increased cost and complexity of the magnet design to ensure compatibility with CT. On the other hand, it is also possible to generate a suitable low-strength magnetic field using conducting coils.

The current study has several limitations. First and foremost, being a simulation study, it can only demonstrate the performance of the proposed system roughly, whereas the overall feasibility and performance will inevitably require actual construction of the system, which is taking our major effort now. Several aspects of the system have not been included in the current study, including CT-induced RF interference to MRI, mechanical stability of CT rotation and calibration, etc., which cannot be well simulated at this stage. However, to our best knowledge, this is the first design of a combined CT-MR scanner in the ULF setting. Simulation results have demonstrated its promising performance. We will first build the ULF MR scanner as a standalone system to evaluate the MRI performance. The prototype MRI subsystem will then serve as a basis for the evaluation of the compatibility with the CT subsystem. Down the road, we will proceed towards the complete portable CT-MRI system construction, demonstration and translation.

Regarding image evaluation tasks in this study, we simulated CT and MR images and assessed the image quality visually or based on standard metrics on image artefacts and resolution. While these approaches are of technical interest, there is a need for evaluating images from a clinically more relevant perspective. For instance, in [26], the authors proposed a radiomics-based methodology to assess images acquired after PET tracer injection and identify important features. Yet, different from the experimental study that can acquire real images over time, the simulation study in this manuscript does not allow to reliably simulate temporal evolution of CT-MR images with respect to disease progression or treatment response. Hence, the radiomics-based image evaluation method cannot be immediately performed in the current setting. Once we have a hardware platform constructed in the near future, these evaluation matrices of more clinical relevance would become feasible and will be generated for system performance characterization.

Second, the novel design in this study also means unique challenges. Historically, ultra-low-field MRI has been neglected due to low SNR. MR scans are often uncomfortable due to long scan time. Reducing field strength would in general further prolong scan time. In recent years, interest in this direction has been renewed to build MRI scanners for its favorable features such as low cost and high mobility. Recent advances in image processing, especially deep learning-based imaging [27], [28], have offered a potential solution to overcome the low SNR challenge in the ultra-low-field regime, ensuring their values in clinical applications. In our study, imaging performance was reported using the Fourier transform-based reconstruction method together with a denoising step. To increase SNR, we employed a simple protocol with a relatively large number of repetitions, hence causing the long data acquisition time. Applying advanced deep reconstruction and processing techniques to improve image quality is beyond the scope of this paper, and will be our future work. We expect that it will be highly rewarding to employ deep learning techniques to mitigate noise and together with super-resolution techniques to maintain image resolution.

Another major direction to improve the proposed system performance is joint CT-MRI reconstruction. With the proposed system, CT and MR images are naturally aligned in the same coordinate system, sharing many similar features, such as edges. Taking advantages of this explicit correlation, many implicit corrective relationships will synergize CT and MRI reconstructions. Previous studies demonstrated feasibility and potential advantages of this joint reconstruction.
approach over conventional image domain regularization [29]. With recent advances in deep learning, it should be possible and desirable training a unified model to learn a joint distributions of CT and MR images, which can serve as a strong prior-information to guide the reconstruction process and improve the resultant image quality.

Another limitation of the proposed system is related to the inhomogeneous magnetic field. Compared to conventional homogeneous field MRI systems, the inhomogeneous field causes intra-voxel dephasing and hence signal loss. We need to use spin-echo MR sequences for refocusing dephased spins during data acquisition. Nonetheless, this inevitably increases data acquisition time compared to rapid gradient based sequences. While some sequences, such as turbo spin echo, are expected to accelerate data acquisition, the inherent signal loss, coupled with the already low SNR in this low-field setup, calls for advanced imaging reconstruction and processing techniques to optimize the image quality.

Comparing to MRI, the CT subsystem basically follows a standard CT scanner design. Because of the relative maturity of the CT technology, we expect less hurdles in developing the CT subsystem. Again, simulation studies were used as a proof-of-principle study to establish the feasibility of our overall idea. More thorough investigation will be performed in the next step to analyze all factors that affect the imaging performance. For example, Monte Carlo radiation transport simulation may be used to evaluate the x-ray scatter effect.

V. CONCLUSION

In this paper, we presented the top-level design of the first portable CT-MRI system and evaluated its imaging performance via realistic numerical simulation studies. In this CT-MRI system, the magnet made of two NdFeB rings of about 40.0 cm radius forms a magnetic field of about 57 mT at the iso-center. It has a gap of 11.33 cm to accommodate the x-ray beam from the CT subsystem. The targeted MR imaging field of view is a sphere of ~15 cm in diameter and that of CT is approximately 20 cm diameter in axial direction and 5 cm in longitudinal direction. Our numerical simulations results demonstrated a great potential of such a hybrid system. We expect the proposed CT-MRI system will be valuable in applications such as imaging in underdeveloped countries, disaster scenes and battle fields.

REFERENCES

[1] D. Townsend, “Multimodality imaging of structure and function,” Phys. Med. Biol., vol. 53, no. 4, pp. R1–R39, Feb. 2008.
[2] T. M. Blodgett, C. C. Metzler, and D. W. Townsend, “PET/CT: Form and function,” Radiology, vol. 242, no. 2, pp. 360–385, Feb. 2007.
[3] M. S. Judenhofer, H. F. Wehrli, D. F. Newport, C. Catana, S. B. Siegel, M. Becker, A. Thielbuer, M. Kneilling, M. P. Lichy, M. Eichner, and K. Klinger, “Simultaneous PET-MRI: A new approach for functional and morphological imaging,” Nature Med., vol. 14, no. 4, pp. 459–465, Apr. 2008.
[4] G. Wang, M. Kalra, V. Murugan, Y. Xi, L. Gjestebry, M. Getzin, Q. Yang, W. Cong, and M. Vannier, “Vision 20/20: Simultaneous CT-MRI—Next chapter of multimodality imaging,” Med. Phys., vol. 42, no. 10, pp. S879–S889, Oct. 2015.
[5] R. Fahrig, K. Butts, J. A. Rowlands, R. Saunders, J. Stanton, G. M. Stevens, B. L. Daniel, Z. Wen, D. L. Eggen, and N. J. Pelc, “A truly hybrid interventional MR/X-ray system: Feasibility demonstration,” J. Magn. Reson. Imaging., vol. 13, no. 2, pp. 294–305, Feb. 2001.
[6] R. Fahrig, Z. Wen, A. Ganguly, G. DeCrescenzo, J. Rowlands, G. Stevens, R. Saunders, and N. Pelc, “Performance of a static-anode/flat-panel X-ray fluoroscopy system in a diagnostic strength magnetic field: A truly hybrid X-ray/MRI imaging system,” Med. Phys., vol. 32, pp. 1775–1784, Jun. 2005.
[7] J. J. Lagendijk, B. W. Raaymakers, and M. Van Vulpel, “The magnetic resonance imaging-linac system,” Seminars Radiat. Oncol., vol. 24, no. 3, pp. 207–209, Jul. 2014.
[8] B. Raaymakers, I. Jürgenliemk-Schulz, G. Bol, M. Glitner, A. Kotte, B. Van Asselen, J. De Boer, J. Bluemink, S. Hackett, M. Moerland, and S. J. Woodings, “First patients treated with a 1.5 T MRI-Linac: Clinical proof of concept of a high-precision, high-field MRI guided radiotherapy treatment,” Phys. Med. Biol., vol. 62, no. 23, pp. L41–L50, Nov. 2017.
[9] J. P. Marques, F. F. Simonis, and A. G. Webb, “Low-field MRI: An MR physics perspective,” J. Magn. Reson. Imaging., vol. 49, no. 6, pp. 1528–1542, Jun. 2019.
[10] P. C. McDaniel, C. Z. Cooley, J. P. Stammkeand, and L. W. Wald, “The MRI map: A single-sided MRI system designed for potential point-of-care limited field-of-view brain imaging,” Magn. Reson. Med., vol. 82, no. 5, pp. 1946–1960, Nov. 2019.
[11] C. Z. Cooley, P. C. McDaniel, J. P. Stammkeand, S. A. Srinivas, S. F. Cauley, M. Śliwiak, C. R. Sappo, C. F. Vaughn, B. Guerin, M. S. Rosen, and M. H. Lev, “A portable scanner for magnetic resonance imaging of the brain,” Nature Biomed. Eng., vol. 5, no. 3, pp. 229–239, Nov. 2021.
[12] Y. Liu, C. T. Leong, Y. Zhao, J. Xiao, H. K. Mak, A. C. O. Tsang, G. K. Lau, G. K. Leung, and E. X. Wu, “A low-cost and shielding-free ultra-low-field brain MRI scanner,” Nature Commun., vol. 12, no. 1, pp. 1–14, Dec. 2021.
[13] M. Meribout and S. Sonowan, “Optimal halbach magnet array design for portable NMR targeting multiphase flow metering applications,” IEEE Trans. Magn., vol. 55, no. 1, pp. 1–7, Nov. 2018.
[14] M. Meribout, “Optimal design for a portable NMR and MRI-based multiphase flow meter,” IEEE Trans. Ind. Electron., vol. 66, no. 8, pp. 6354–6361, Oct. 2018.
[15] M. Meribout and M. Kalra, “A portable system for two dimensional magnetic particle imaging,” Measurement, vol. 152, Feb. 2020, Art. no. 107281.
[16] R. Turner, “Gradient coil design: A review of methods,” Magn. Reson. Imag., vol. 11, no. 7, pp. 903–920, Apr. 1993.
[17] A. Macovski, “Noise in MRI,” Magn. Reson. Med., vol. 36, no. 3, pp. 494–497, Sep. 1996.
[18] M. Maggioni, V. Katkovnik, K. Egiazarian, and A. Foi, “Nonlocal transform-domain filter for volumetric data denoising and reconstruction,” IEEE Trans. Image Process., vol. 22, no. 1, pp. 119–133, Jul. 2012.
[19] L. A. Feldkamp, L. C. Davis, and J. W. Kress, “Practical cone-beam algorithm,” J. Opt. Soc. Am. A, Opt. Image Sci., vol. 1, no. 6, pp. 612–619, Jun. 1984.
[20] H. Yu and G. Wang, “Compressed sensing based interior tomography,” Phys. Med. Biol., vol. 54, no. 9, pp. 2791–2805, Apr. 2009.
[21] W. Van Aarle, W. J. Palenstijn, J. De Beenhouwer, T. Altantzis, S. Bals, K. J. Batenburg, and J. Sijbers, “The ASTRA toolbox: A platform for advanced algorithm development in electron tomography,” Ultramicroscopy, vol. 157, pp. 35–47, Oct. 2015.
[22] L. A. Shepp and B. F. Logan, “The fourier reconstruction of a head section,” IEEE Trans. Nucl. Sci., vol. NS-21, no. 3, pp. 21–43, Jun. 1974.
[23] H. M. Gach, C. Tanase, and F. Boda, “2D & 3D Shepp-Logan phantom standards for MRI,” in Proc. 19th Int. Conf. Syst. Eng., Aug. 2008, pp. 521–526.
[24] M. J. Ackerman, “The visible human project,” Proc. IEEE, vol. 86, no. 3, pp. 504–511, Mar. 1998.
[25] Z. Wen, R. Fahrig, and N. J. Pelc, “Robust X-ray tubes for use within magnetic fields of MR scanners,” Med. Phys., vol. 32, pp. 2327–2336, Jul. 2005.
[26] V. Benfante, A. Stefano, A. Comelli, P. Giacone, F. P. Cammarata, S. Richiuse, F. Scopelliti, M. Pometti, F. Ciccar, S. Cosentino, and M. Lunardon, “A new preclinical decision support system based on PET radiomics: A preliminary study on the evaluation of an innovative 64-Cu-labeled chelator in mouse models,” J. Imag., vol. 8, no. 4, pp. 92, Mar. 2022.
YUETING PENG was born in Henan, China. She received the B.Sc. and master’s degrees in physics from Henan Normal University, in 2012 and 2015, respectively, and the Ph.D. degree in physics and applied physics from the University of Texas at Arlington, in 2020. She started to work in the medical physics field, from 2021. From 2012 to 2020, she was involved with the computational simulations on the optoelectronic and thermodynamic stability of low-dimensional semiconductor materials and published more than 30 journal articles in Small, Physical Review B, Acta Materialia, Physical Chemistry Chemical Physics, Journal of the European Ceramic Society, and Journal of the American Ceramic Society. Her current research interests include numerical simulations of MR imaging system, imaging process, and image reconstruction. She received multiple awards and scholarships, including the National Graduate Fellowship of China, Carrizo Graduate Fellowship, the James L. Horwitz Physics Scholarship, Scharff Physics Award, the Science Dean’s Excellence Scholarship, and Dr. John L. Fry Endowment for the Department of Physics.

MENGZHOU LI (Member, IEEE) received the B.S. degree in optoelectronic information engineering, and the M.S. degree in instrument science and technology from the Harbin Institute of Technology, in 2014 and 2016, respectively. He is currently pursuing the Ph.D. degree with the Department of Biomedical Engineering, Rensselaer Polytechnic Institute, Troy, NY, USA. His research interests include X-ray tomographic imaging, super-resolution light microscopy, and machine learning. He received awards and honors include the National Scholarship (Ministry of Education of China), the China Instrument and Control Society Scholarship, in 2014, and the First-Class Science and Technology Progress Award (Chinese Society for Measurement), in 2018.

JACE GRANDINETTI was born in Pensacola, FL, in 1992. He received the B.S. degree in physics from the University of Texas at Arlington (UT), in 2015 and the M.S. degree in medical physics from Vanderbilt University, in 2018. He is currently pursuing the Ph.D. degree in medical physics, UT Southwestern. His current research interest includes hardware development for low-field MR systems and deep-learning reconstruction methods for MRI.

GE WANG (Fellow, IEEE) is the Clark & Crossan Chair Professor and the Director of Biomedical Imaging Center, Rensselaer Polytechnic Institute, Troy, NY, USA. He pioneered the spiral cone-beam/multislice method in the early 1990s and many follow-up papers in this area. There are 200 million medical CT scans yearly, a majority of which are performed in the spiral cone-beam mode. He published the first perspective on AI-empowered tomographic imaging, in 2016, and a series of papers on diverse deep learning-based imaging topics. He wrote more than 550 journal articles in PNAS, Nature, Nature Machine Intelligence, Nature Communications, and other well-known journals. He gave many seminars, keynotes and plenaries including NIH AI Imaging Presentations, in 2018 and SPIE O+P Plenary, in 2021. He is a fellow of SPIE, AAPM, OSA, AIMBE, AAAS, and the National Academy of Inventors (NAI). He received various awards, including IEEE EMBS Academic Career Achievement Award, in 2021, IEEE R1 Outstanding Teaching Award, in 2021, and SPIE Aden and Marjorie Meinel Technology Achievement Award, in 2022.

XUN JIA was born in Heilongjiang, China. He received the master’s degree in applied mathematics, in 2007, and the Ph.D. degree in physics, in 2009, from the University of California at Los Angeles. He is currently a Professor and the Chief of the Medical Physics Division, Department of Radiation Oncology and Molecular Radiation Sciences, Johns Hopkins University. He has conducted productive research in several areas including low-dose cone beam CT reconstruction, GPU-based Monte Carlo radiation transport simulation, deep-learning for image processing and radiotherapy treatment planning, preclinical small animal irradiation technologies, and automated high-dose-rate brachytherapy. He has published over 150 peer-reviewed manuscripts. His research has been funded by NIH, state, industrial, and charitable funding agencies. He is currently serves as an Executive Editorial Board Member for Physics in Medicine and Biology and Associate Editor of Medical Physics, textscIEEE Transaction on Radiation and Plasma Medical Sciences and a few other journals. He is the recipient of the John Laughlin Young Scientist Award by the American Association of Physicians in medicine, in 2017.

**References**

[27] G. Wang, J. C. Ye, and B. De Man, “Deep learning for tomographic image reconstruction,” Nature Mach. Intell., vol. 2, no. 12, pp. 737–748, Dec. 2020.

[28] C. Shen, D. Nguyen, Z. Zhou, S. B. Jiang, B. Dong, and X. Jia, “An introduction to deep learning in medical physics: advantages, potential, and challenges,” Phys. Med. Biol., vol. 65, no. 5, Mar. 2020, Art. no. 05TR01.

[29] F. Knoll, M. Holler, T. Koesters, R. Otazo, K. Bredies, and D. K. Sodickson, “Joint MR-PET reconstruction using a multi-channel image regularizer,” IEEE Trans. Med. Imaging, vol. 36, no. 1, pp. 1–16, Jan. 2016.